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Proceedings of the Twelfth Congress of the

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Montreal, Quebec, Canada

Editors

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FOREWORD

The first International Society of Electrophysiology and Kinesiology (ISEK) Congress was held in Montreal, at the Queen Elizabeth Hotel, in August of 1968. This first international meeting of the Society was organized by a local committee working under the guidance of Dr. John Basmajian.

The XII ISEK Congress is also held in Montreal, at the Queen Elizabeth Hotel, this time under the Honorary Presidency of Dr. Basmajian, who is professor emeritus of medicine and anatomy at McMaster University, Hamilton, Ontario, Canada. It is the belief of the organizing committee that this Congress is the occasion to recognize the progress made in the last thirty years in our fields of research and teaching. It is also the occasion to acknowledge the vitality of the Society with regards to its aims and objectives.

The members of the Organizing Committee are : A. Bertrand Arsenault, Ph.D., (Chairperson), Universite de Montreal, Canada ; Patricia McKinley, Ph.D. (co-chairperson), McGill University, Canada ; Yazunobu Handa, M.D., Tohoku University Graduate School of Medicine, Japan ; Keith C. Hayes, Ph.D., The University of Western Ontario, Canada ; Bernard Maton, Ph.D., Universite de Paris-Sud, France ; Bradford McFadyen, Ph.D., Universite Laval, Canada ; Serge Roy, Ph.D., Boston University, USA.

Our Keynote Speakers are : Carlo DeLuca, Ph.D., (The John Basmajian Lecture) Boston University, USA ; Anatol Feldman, Ph.D., Universite de Montreal, Canada ; Marco Knaflitz, Ph.D., Politecnico di Torino, Italy ; Toshio Moritani, Ph.D., Kyoto University, Japan ; Alan McComas, M.D., McMaster University, Canada ; Jaynie Yang, Ph.D., University of Alberta, Canada.

For the first time, the John V. Basmajian Student Investigator Award will be offered. This award is designed to promote the presentation of work from student researchers at the Congress. This contest is under the responsibility of Dr. Anne Mannion (Switzerland).

We are pleased that you could join us in Montreal.

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ABSTRACTS

Motor Units Alive - Understanding Them One Pulse at a Time

C. DeLuca
Boston University, Boston, Mass., USA

It is a great honor for me to deliver the Basmajian Lecture at the XII th ISEK Congress which celebrates the 30 th Anniversary of ISEK in the great city of Montreal, the place it all began in September 1968. This date also has an important meaning for me personally. It was on this very same month of this very same year that I first met and became a student of Dr. Basmajian. From that moment on, the direction of my research career was fixed. It was in his laboratory and under his tutelage that I began a long trek into the world of motor unit control. It is, therefore, fitting that I speak on this topic as a contribution to Dr. Basmajian's legacy.

The work that I am about to describe has had contributions from numerous associates and students. They are too numerous to mention, but their contributions are well documented in the literature.

From the beginning, we set out to study the recruitment and firing rate behavior of simultaneously active motor units. A technique, now called "Precision Decomposition" was developed (LeFever et al., 1982; LeFever and De Luca, 1982; Mambrito and De Luca, 1984; Mambrito and De Luca, 1983; De Luca, 1993). It enabled us to separate the individual action potentials from a needle EMG signal detected during isometric contractions. This technique is capable of decomposing with 100 % accuracy the EMG signal obtained during contractions which reach the maximal voluntary contraction (MVC) level, but only for isometric contractions and not for all contractions. We have decomposed signals containing up to 9 motor units with considerable effort; decomposition of 4 motor units is more typical. We developed a test for verifying the accuracy of the technique. This was a non-trivial problem because it is not obvious that the decomposition of a set of superimposed pseudo random action potential trains will have a unique solution. The technique was originally developed using a special needle electrode which detects three independent channels of EMG signal from three pairs of differential 50 pm diameter electrode surfaces located orthogonally. Recently it has been modified to work with wire electrodes consisting of three 50 pm diameter wires glued together and cut to expose only the cross section of the wires. The signals are conditioned to accentuate the differences in the wave shapes and are sampled at 50 k/s. Rule based algorithms identify action potentials using template matching and probability of firing, resolve superpositions, and allocate the action potentiaIs to motor units. User interactive editing algorithms are used to check the accuracy and make modifications according to well established rules. Having the precise timing of all the action potentials of several concurrently active motor units allows us to study the recruitment, firing rate, and synchronization interaction.

The first observation was the Firing Rate Decay (De Luca and Forrest, 1973; De Luca, 1985; De Luca et al., 1996). We reported that during isotonic isometric contractions, the firing rate of the motor units decreased as a function of time. As the firing rate decreased, we never saw a new motor unit being recruited during the first 20 s of a contraction. We suggested (De Luca, 1979)and later (De Luca et al., 1996) we interpreted the firing rate decrease during sustained voluntary contractions to be a manifestation of two phenomena: a) The intrinsic property of the motoneuron to exhibit a firing rate decay over time when stimulated with a DC current which was first described in an animal preparation by Kernel1 (1965). b) A reduced need to fire a motor unit due to the increase in amplitude and duration of its force twitch upon repeated discharge, commonly referred to as twitch Potentiation.

The second observation was the phenomenon of Common Drive (De Luca et ***al.,*** 1982a; 1982b). We found that the firing rates of motor units fluctuated in unison with essentially no time delay between them. This was seen by cross-correlating the firing rates of pairs of concurrently active motor units. We saw this ***behavior*** in all muscles tested, ranging from small distal muscles to large proximal muscles. Even motor units belonging to different motoneuron pools exhibited common firing rate fluctuations when controlled as one functional unit; this we observed during antagonist muscle co-activation (De Luca and Mambrito, 1987). The existence of the common drive has been verified by independent investigators (Miles, 1987; Stashuk and de Bruin, 1988;

Guiheneuc, 1992; Iyer et al., 1994; SemmIer et ***al.,*** 1997). This finding that the motor units are not controlled independently contradicted the previously proposed concept of individual control of motor units. It indicates that the CNS has evolved a relatively simple strategy for controlling motor units. Additionally we found that the irregular nature of the firing rates and the common drive phenomenon imply that muscles cannot produce smooth constant forces. We verified this fact by cross-correlating the firing rates and the force output of the muscle and found a significant correlation with a latency due to the mechanical delay in force build up of the muscle fibers and force transmission through the muscle and tendon tissue.

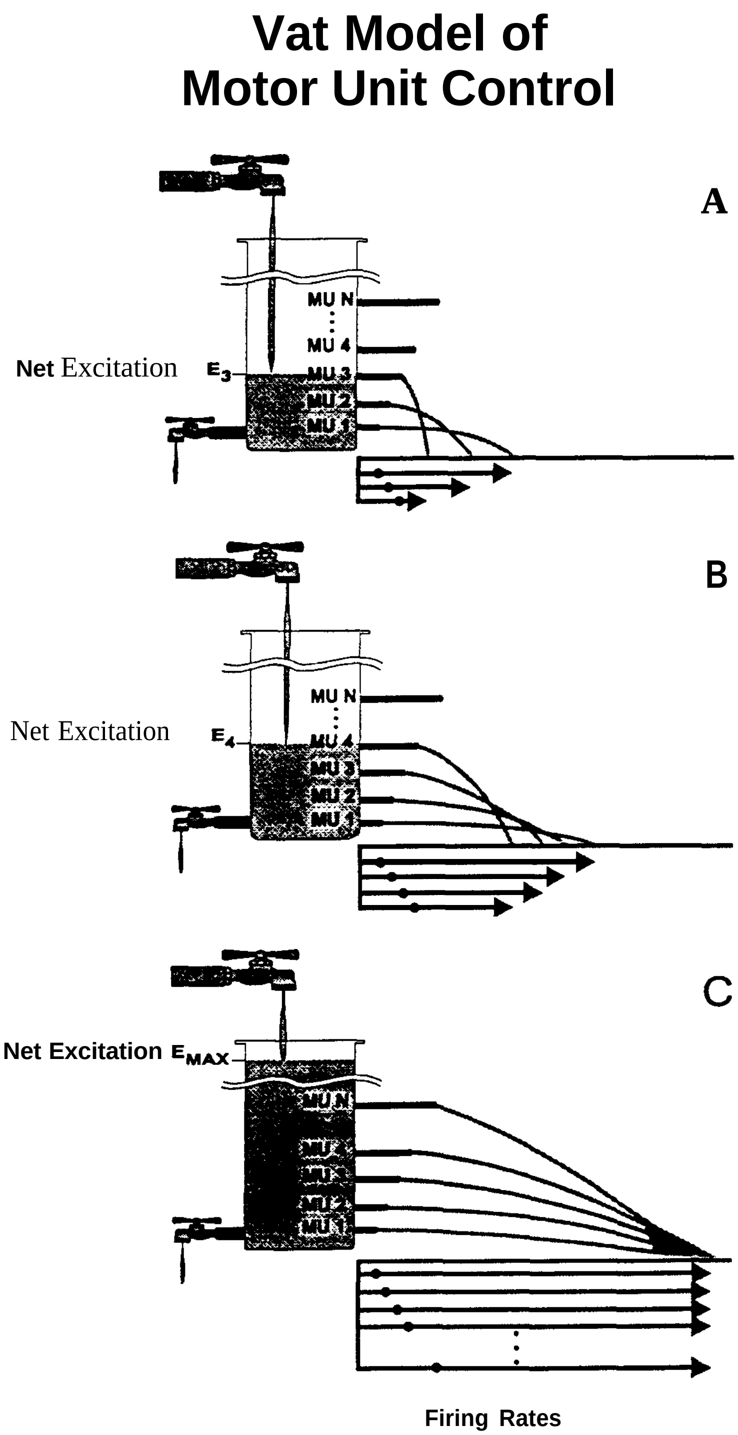
The existence of a high degree of cross-correlation between the firing rates of motor units does not imply that the individual firings of the motor units are synchronized. By <u>Svnchronization</u> it is meant that motor units fire at a fixed time latency with respect to each other. Synchronization occurs in two modalities: short-term and long-term. A study of motor unit pairs detected during isometric isotonic contractions in six muscles revealed that an average of 8 % of the firings were short-term synchronized and only 1 % long-term synchronized (De Luca et al., 1993). Short-term synchronized firings occurred at sporadic intervals and in bursts of typically one or two consecutive firings which had no apparent effect on the force produced by the muscle. We concluded that synchronization of motor unit firings is an epi-phenomenon with no physiological design of its own.

The third observation was the <u>Onion Skin</u> phenomenon. Along with Person and Kudina (1972) as well as Tanji and Kato (1973), we (De Luca and Forrest, 1973) were among the first to report that during isometric contractions lasting less than 20 s, the earlier recruited motor units always fired at greater average rates than later recruited motor units. (When the firing rates are plotted as a function of time, the hierarchal values of the motor unit firing rates form overlapping layers resembling the structure of the skin of an onion.) Subsequently (De Luca et al., 1982a) we documented this phenomenon in detail. Independent verifications followed by Hoffer et al. (1987) and Stashuk and de Bruin (1988). Thus, the later recruited, more glycolytic, faster-twitch motor units which require a greater firing rate than the earlier recruited, more oxidative, slower-twitch motor units to fuse would be less likely to tetanize. Even during high level contractions in the neighborhood of 80 to 100 % MVC, the firing rates of the high threshold motor units is in the range of 20 - 30 pulses per second (pps), an amount likely to be insufficient for complete tetanization. This finding ran counter to the previously held belief that higher threshold motor units would be expected to fire faster. The onion skin phenomenon begs the question as to why motor unit control developed so as to not maximize the force generating capacity of a muscle. In fact, there was a previous widely held belief that the higher threshold motor units would fire with greater firing rates so as to produce more force. After all, if the purpose of a muscle was to generate force, it was reasonable to speculate that the motor unit control would be organized to make the most of available mechanical capacity within the muscle. Why should muscles evolve to have an apparent <u>Reserve Canacitv</u> not commonly accessible during voluntary contractions? This an intriguing question. One possible explanation is that the higher threshold motor units, which are faster fatiguing, would become exhausted quickly if they fired fast. A control system so organized would not provide sustained contractions at high force levels which would be necessary to cope with life threatening situations and ensure the survival of the species. It appears that the motor unit control developed to maximize a combination of contraction force and contraction time rather than only the contraction force. The available reserve capacity for generating force over brief periods of time may explain the occurrence of exceptional feats of strength that are reported to occur during life threatening situations.

Two corollary observations were also made for the behavior of the firing rates: a) The latter recruited motor units had greater initial firing rates as previously indicated by Clamann (1970). b) The firing rates of all units converged to a near common value during maximal contractions (De Luca and Erim, 1994; Erim et al., 1996).

All the above findings indicate that he control signals (net excitation) act on the motoneuron pool as a unit. As proposed by Henneman and colleagues (Henneman et al., 1965a; 1965b) the individual properties of the motoneurons determine the recruitment hierarchy in response to the net excitation. To that enlightening observation we now add that the firing rate of the individual motor units respond to the net excitation communally and simultaneously, and that the average value of the firing rates is also hierarchically organized with an inverse relationship to the recruitment threshold.

The fourth observation was the Diversification of the control properties (De Luca et al., 1982a). The motor units of smaller, distal muscles such as the first dorsal interosseous tend to be recruited in the force range up to 50 % MVC and have mean firing rates which reach relatively high values (approx 40 pps) at 80 % MVC. **Whereas,** those from larger, proximal muscles such as the deltoid and the trapezius recruit their motor units in a force range up to 80 % MVC and have firing rates which reach relatively lower values (approx 30 pps). A similar observation in the adductor Pollicis and biceps brachii muscles was reported independently by Kukulka and Clamann (1981). The reduced dynamic range of the larger more proximal muscles may be due to the increased recurrent inhibition of the Renshaw system which is more prominent in these muscles as shown by Rossi and Mazzachio (1991). These diverse control properties are useful in at least two ways. First, they allow for a smoother force. Smaller muscles have less motor units, therefore, force gradation due to recruitment would be coarser throughout the full range than in larger muscles which have many more (an order of magnitude or more) motor units. Second, the larger more proximal muscles tend to be more postural and are required to produce sustained contractions more often. The lower firing rates in these muscles delay the progression of fatigue.



Many of the above observations can be conveniently summarized by a simple hydraulic analog which we call the Vat Model (De Luca and Erim, 1994). The water flowing into the tank corresponds to the excitation reaching the motoneuron pool, and the water flowing out of the tank corresponds to the inhibition. The level of the water in the tank corresponds to the net excitation received by the motoneuron pool. The broken line indicates that the height of the tank is greater than depicted. The spouts correspond to the motor units, with their location on the side of the tank corresponding to their recruitment threshold. The length of the spout corresponds to the initial firing rate and the distance traveled by the water stream corresponds to the firing rate of the individual motor unit. The common drive is represented by the fact that all the motor units respond to a fluctuation in the level of the tank. (A) Demonstrates the case when the net excitation is sufficient to activate only three motor units. (B) As a new motor unit is recruited, the firing rate of previously activated motor units increases. (C) As the excitation reaches the maximal value, (presumably) all the motor units have been recruited and the firing rates tend to converge to the same value.

We found that long-term Exercise appears to induce modifications in the motor unit control properties (Adam et al. , in press). We compared the motor unit control parameters of the first dorsal interosseous muscles of the dominant and non-dominant hands performing isometric, isotonic contractions at the same MVC level. We found that when compared to the non-dominant hand, motor units of the dominant side had lower firing rates for the same level of contraction, and a larger number of motor units were found to be recruited at lower force levels. This finding is consistent with previously known facts that the dominant hand has slower-twitch muscle fibers, probably due to the life-long preferential use. The twitches of slower fibers fuse at lower firing rates allowing for a reduced excitation and decreased firing rates in the dominant hand without a reduction in force output.

My colleague, Dr. Z. Erim, and I have recently found that Aging causes alterations in the motor unit control properties (Erim *et al.,* 1997). In our study in the first dorsal interosseous muscle of elderly subjects above 65 years of age, we found that the firing rate and the recruitment threshold of motor units became modified in the same manner as that induced by exercise. This observation was not surprising because it is well known that aged muscles contain a greater percentage of slow twitch Type I fibers, as is the case in exercised muscles, although the cause for the increased percentage of Type I fibers appears to be different. When we studied the common drive in the elderly, we found that in approximately one-half of them the cross-correlation between pairs of motor units was severely reduced and in some cases apparently non existent. Also, in the elderly, the onion skin phenomenon was disrupted. When plotted as a function of contraction time, the firing rates of numerous motor units crossed over those of earlier recruited ones and the behavior of the firing rates was not orderly in a hierarchial sense; some decreased while others increased during an isometric, isotonic contraction. We surmise that this dissociation among the firing rates of motor units leads to an inefficient force generation scheme.

All the above observations were made on relatively short-duration (less than 20 s) isometric, isotonic contractions. They may not fully describe the behavior of the control properties during sustained contractions of limb muscles or postural muscles which are commonly required to contract for prolonged periods of time. My colleague Dr. R. Westgaard and I have seen cases where the onion skin property is disturbed during short-term (20 s or less) contractions of normal healthy trapezius muscle and during long-duration (150 s or more) contractions in normal healthy first dorsal interosseous muscles. These are relatively new observations. We suspect that the cross over of the firing rates is due to at least two factors which cause the firing rates of earlier recruited motor units to decrease below the value of the newly recruited motor units: a) The Renshaw recurrent inhibition of earlier recruited motor units which is more dominant in proximal muscles such as the trapezius, hence the disturbed onion skin during short-duration contractions. b) The motoneuron adaptation process reported by Kernel1 (1965) which decreases the firing rates of motor units during sustained activation causing the discharge rates of earlier recruited motor units to decrease below that of later recruited motor units.

While studying long-duration contractions in the range of 5 min to 60 min, Westgaard and I also observed definite examples of Motor Unit Substitution. These are cases where a motor unit stopped firing during a sustained contraction when the activity level decreased slightly, and in response to a subsequent slight increase in the force output, a new motor unit was recruited in place of the one which was derecruited. We believe this phenomenon is the result of adaptation of the recruitment threshold of active motoneurons. The recruitment threshold of a motor unit which had been active for some time would have become greater than the recruitment threshold of the next one in the hierarchy. In this fashion the next motor unit becomes recruited in response to an increase in the net excitation to the motoneuron pool.

We continue to pursue this line of investigation and we anticipate additional interesting observations.

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Control Body Configuration During Load Lifting: A Simulation Study

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Introduction. During trunk extension, the weight of the trunk and the external load are distributed between different tissues : bones, ligaments and muscles. The nervous system controls the activity of muscles and thus may influence the amount of stress applied to passive tissues. Direct, in vivo, measurements of forces acting at the level of muscles and passive tissues may be difficult. To overcome this problem, we combined the results of experimental studies with mathematical modeling and computer simulation of appropriate movements. We addressed the problem of the control of the body configuration during load lifting using a version of the equilibrium point hypothesis (the λ model)[1,2].

Model and Results. The trunk is considered an inverted pendulum with two degrees of freedom (trunk inclination in sagittal plane and the lumbar spine curvature, lumbar lordosis, Figure 1).

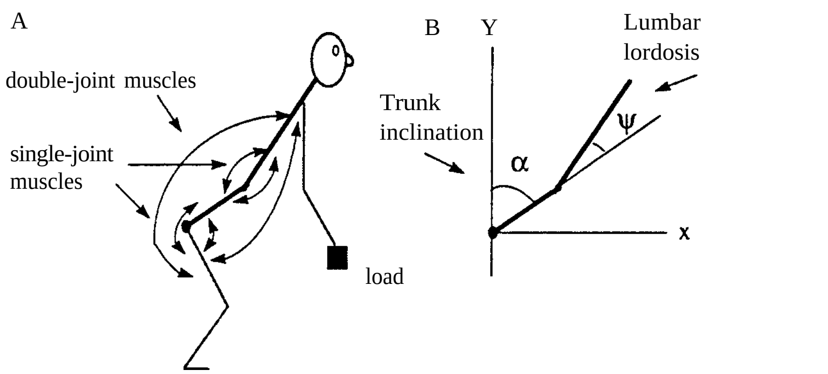

Figure 1. Two joint representation of the body (A) and definition of angles (B) : α - trunk inclination angle and ψ - lumbar lordosis characterising the lumbar curvature.

The dynamical equations of weight lifting include gravitational, active (muscle) and passive forces. The latter are represented by nonlinear elastic elements. Both single and double-joint muscles are considered in the model. Muscle forces are generated depending on muscle length, velocity and control variables : the reciprocal (R) command, co-activation (C) command and a damping factor, (μ). The R command is the set of joint angles at which the transition from agonist to antagonist muscle torque or vice versa occurs. The C command is the set of angular zones in which agonist and antagonist muscles may be active simultaneously. The time dimensional command (μ) is the ratio of the dynamic to static sensitivity of muscle spindle afferents.

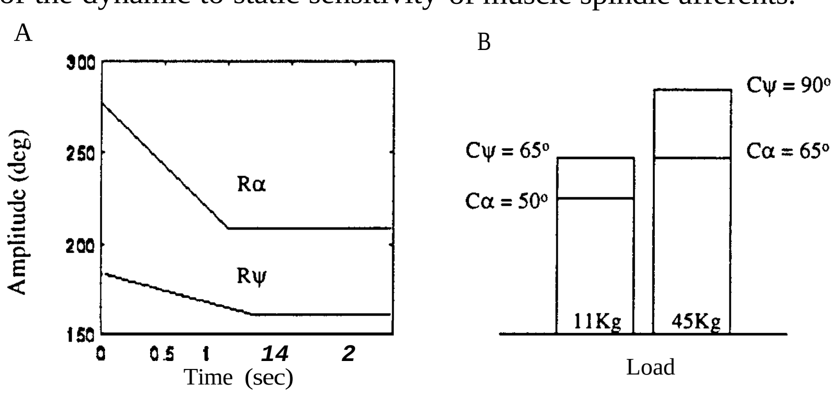

Figure 2. Patterns of control variables (A) : Initial and final values of R commands (deg) correspond to the initial and final configurations of the trunk (B) : constant C commands (deg) were increased for the greater load, μ = 300 msec.

Shifts in the equilibrium position of the trunk are produced by a ramp-shaped pattern of changes in the R command for each degree of freedom. The C and μ commands remained constant during lifting a given load (Figure 2). However, the values of C commands were increased with the increasing load. Such a control strategy produces the necessary stiffness of the body, prevents oscillations and provides stability of the final vertical posture. The model reproduced the coordination between trunk flexion/extension angle and lumbar lordosis during lifting of different loads[3,4] as well as EMG patterns characteristic of individual subjects. The EMG patterns in the model (Figure 3) corresponded to well known phenomenon' of silence in trunk extensor muscles in the horizontal position of the trunk when the spine is fully flexed.

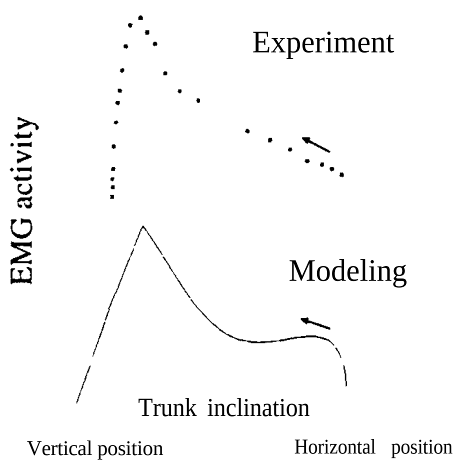

Figure 3. Observed EMG pattern (upper curve) compared to the simulated EMG (lower curve). The arrows indicate the direction of the trunk movement from the horizontal to the vertical position..

Conclusions. This study demonstrates that comparatively complex kinematic and EMG patterns can be simulated using direct dynamical solutions with simple control inputs. This approach may be helpful in the analysis of altered biomechanics and muscle activation patterns in patients with low back pain.

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Implementation of Software Adaptation for 3-D Cervical Kinematics by Means of a Magnetic Tracking Device

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INTRODUCTION: AIMS OF THE SOFWARE ADAPTATION

In order to realise users friendly registrations of joint kinematics for clinical use and research applications, software was written to control, to treat and to organise the output stream of kinematic results, obtained with a commercial available Flock of Birds (FOB) magnetic tracking device. In particular, the use of the magnetic tracking system with two receivers, operating in the master-slave configuration, was considered.

The software adaptation was created in order to obtain:
1. registrations of discrete *motion sequences,*
2. *relative* kinematic data, resulting from changes of orientation and position between the master and slave receivers (estimating the changes of orientation and position of the body segments to which they are attached),
3. zero *values* for the orientation of the reference systems of both receivers *in the starting position* (via software based correction of reference frames),
4. *simultaneous graphical display of Eulerian angles* (in order to verify coupled motion) and *combined graphical display of the translation components,*
5. data storage formatted in a way spreadsheets can access them.

RESULTS AND DISCUSSION

Control of data registration was aided by the introduction of a number of users friendly menus. The main menu provides a “start” and “stop” function to determine a distinct period of kinematic observation. Relative kinematic data are calculated from the absolute output data, received from the FOB in “stream mode”. Starting from zero values is obtained by mathematically pushing the orientation of the reference system of the master receiver to that of the slave receiver at the moment of the “start” instruction. Via the “graph” menu, simultaneous graphical display of Eulerian angles can be realised. The possibility exists of opening a second window, showing the three translation components. Additional possibilities allow further handling of the graphical results and adding personal information and comments. Via the “file” menu results are accessible for modern spreadsheets.

A posteriori software based adaptation of the orientation of the references frames of the receivers has been shown useful to overcome the differences between the orientation of technical and anatomical reference frames and to avoid singularities in the calculation of angles.

With the described software adaptations, axial rotations of the cervical spine, performed in flexion, in extension and in neutral position, were studied in 40 male and 43 female volunteers.

Examples are given in Fig. 1-4. Preliminary results showed classical patterns of coupled motion in the cervical spine, however with **a** number of inter individual differences that deserve special attention in further analysis of these data.

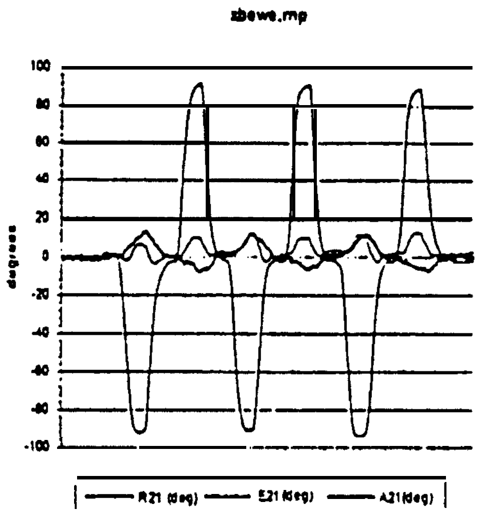

Fig. 1. observation of coupled motion during subsequent axial rotations of the cervical spine subject xbewe(*)
- cervical spine in the neutral position -

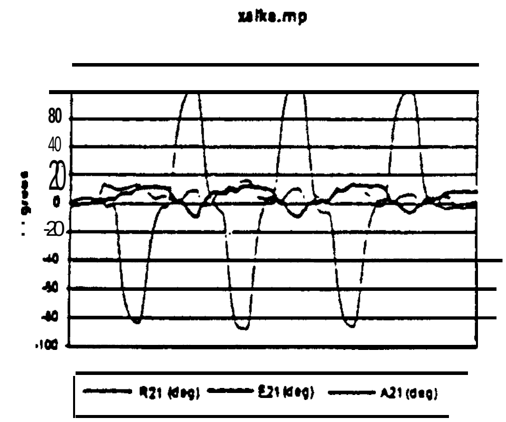

Fig. 2. observation of coupled motion during subsequent axial rotations of the cervical spine subject xalka(*)
- cervical spine in the neutral position -

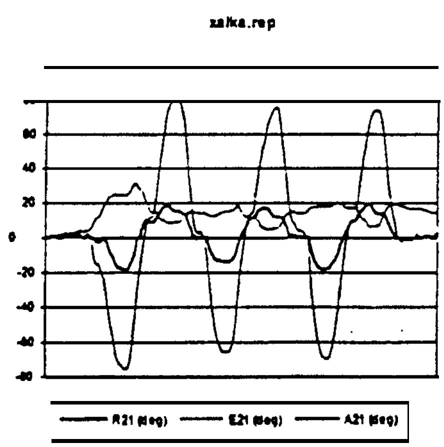

Fig. 3. observation of coupled motion during subsequent axial rotations of the cervical spine subject xalka (*)
- cervical spine in the extension -

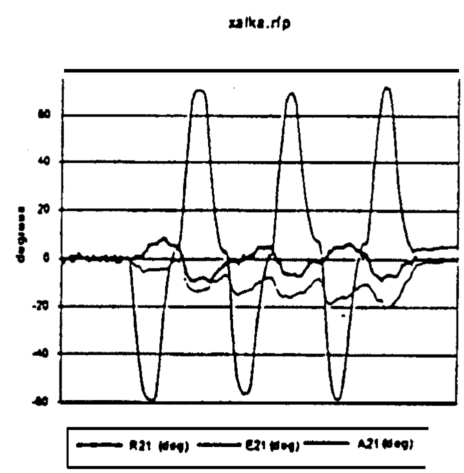

Fig. 4. observation of coupled motion during subsequent axial rotations of the cervical spine subject xalka(*)
- cervical spine in flexion -

(*) R21 left-right lateroflexion
E2 1 flexion-extension
A21 left-right axial rotation

A Physiologically Based Simulation of the Electromyographic Signal

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INTRODUCTION. The surface electromyogram is a noisy interference pattern, shaped by many factors and subject to high inter-subject variability. A single recording alone, can thus yield little quantitative information about the muscle behaviour. However, information about motor unit firing and recruitment patterns, fibre conduction velocities and distance from the electrodes, is embedded within the signal. The question remains as to how much of this data has been obscured, how much remains to be extracted and how this can best be done. The potential benefits of an accurate physiologically based model that has the ability to simulate gross electromyography, (EMG), under a variety of conditions, should be substantial. The ability to alter parameters such as muscle fibre type and distribution, fibre conduction velocity and central nervous system control mechanisms, offers the possibility to explore the influence of such factors on the signal in a manner which would clearly not be possible *in vivo. This* study presents such a model, compares the simulated signal with real EMG data and explores the information contained in the frequency spectrum during a sustained fatiguing contraction.

METHODS

Model: Much work has been done in modelling single Fibre and motor unit action potentials based on modified versions of traditional volume conductor models, (Andreassen and Rosenfalck,l98 1 ;Nandedkar and Stalberg,1983;Griep et al.,1982). Few attempts, however, have been made to extend these models to simulate gross voluntary EMG. Models of the surface electromyogram have instead been based mainly, on a standard linear dynamic system, (Agarwal and Gottlieb, 1975). Although signals are produced which appear similar to raw EMG, and which may yield theoretical information such as about the effect of firing statistics on the spectrum, (Lago and Jones, 1977), their application as accurate simulations of physiological EMG is clearly limited. The fundamental building block of the model presented here is the action potential recorded at the electrodes when a single fibre is stimulated. Action potentials are calculated individually for each fibre in every motor unit, taking account of fibre diameter, length and conduction velocity, the random nature of the end-plate locations and the electrode configuration. Action potentials are calculated in the time domain using the line source model and the algebraic expression for the transmembrane potential described by Rosenfalck(1969), modified by Nandedkar and Stalberg(198 1). Motor units size can be varied in size and can be located randomly through the muscle cross section, or assigned a particular distribution pattern. Fibres are located randomly within their motor unit territory. Each motor unit is assigned a mean interpulse interval, with Gaussian distribution. The potential recorded by the electrodes is obtained by summing the contributions of each motor unit. External noise and filtering can be applied at this stage.

Recordings: EMG signals were recorded with bipolar electrodes from the vastus lateralis and biceps brachii of 10 subjects during an 25sec isometric maximum voluntary contraction. The spectrum of the signal at the beginning and end of the contractions were obtained, and the normalised cumulative power at each frequency was calculated for both. The shift in each percentile frequency was then examined (i.e. at 0%..50%..100% of total cumulative power), enabling the frequency shift to be examined across the spectrum and the change in conduction velocity to be estimated (Lo Conte and Merletti, 1996).

RESULTS and DISCUSSION. Theoretical and experimental observations described in the literature are apparent in the model simulation. These include the low pass filtering effect of muscle tissue and the subsequent relationship between Fibre-electrode distance and MUAP amplitude and duration, the effect of fibre conduction velocity on these and on the frequency spectrum, the influence of the recording configuration on the spectrum, and of the motor unit firing rates on the low frequency content. EMG signals were simulated varying the conduction velocity. When ratio of the new frequency to the original for each percentile in the spectrum was examined, the model produced a graph remarkably similar to that of the experimental data. All subjects illustrated a spectral compression by an approximately constant factor throughout the mid-frequency range, suggestive of a decrease in fibte conduction velocities (Stullen and De Luca,1981). This is supported by the observed increase in the EMG amplitude, (recruitment being complete above ~75%MVC). In both the simulated and experimental data, the ratio deviated from the expected value at the lower and higher ends of the spectrum, see Fig.1. At the higher end, the ratio approached unity as the MUAP frequency content lay predominantly below this region. In the model, the firing rates were simulated to remain constant throughout the contraction, therefore the power in the spectrum due to the firing rates remained stationary, while that due to the MUAP shape altered as the conduction velocity changed. This resulted in clearly visible peaks at the mean firing rate and its harmonics. In the experimental data, periodic low frequency peaks were observed in most subjects, indicating that the mean firing rate remained relatively constant during the contraction, and enabling an estimate of this value to be made. Dips at approx. 1OOHz are due to the bipolar configuration (fdip=cv/(interelectrode dist.)). Although some recordings of surface electromyography have been reported where peaks corresponding to the firing rates are visible in the raw spectrum (Van Boxtel and Schomaker, 1983),this was not the case here.

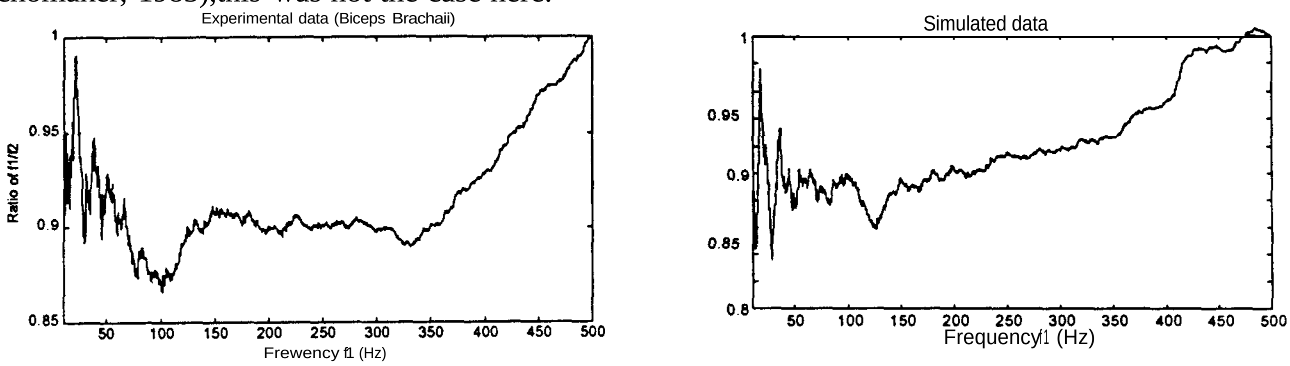

Figure 1: Experimental and simulated data, showing the ratio of f, to f2 through the range of the freq. spectrum. In the simulated data the conduction velocity decreased to 90% of its original value, mean firing rate is 2OHz. The difference in slope at the higher end of the spectrum is due to the presence of high freq. noise in the real data.

CONCLUSION. A model of the gross EMG signal has been developed to simulate the events taking place during voluntary contraction. It allows different muscle architectures and nervous system control mechanisms to be explored. The model indicates that examining the cumulative power in the mid-freq. region, i.e. between 100 and 350Hz for the data above, yields a more accurate estimate of the conduction velocity than the cumulative power between 0.05 to 0.95, recommended by Lo Conte and Merletti (1996). The model also indicates that it may be possible to observe peaks in the ratio of f_2/f_1 at the mean firing rate and its harmonics, even when these are not visible in the raw spectrum. This is supported by the experimental data gathered. Any filtering process, including the recording electrode configuration, will have a significant influence on these estimates. The model as designed offers the possibility of examining many other aspects of EMG analysis, in a manner similar to that applied here to fatiguing contractions.

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Can Peripheral Motor Drive be Estimated from Surface EMG Recordings ? An Experimental Simulation Study

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Introduction

The EMG signal results from the summation of trains of motor unit action potentials (MUAP) of motor units (MU) with territories in the vicinity of the recording electrode. Thus the features of the EMG are determined by both peripheral parameters (MUAP shape and size) and by central parameters (MU firing statistics). However, during voluntary contractions the relative contributions of these parameters to EMG characteristics cannot be identified from EMG recordings alone. In particular, it can not be determined whether and how reliably peripheral motor drive, here defined as the ensemble of the firing rates of all the MUs of a muscle, can be estimated from EMG recordings. The matter is complicated by the fact that different MUs contribute to the EMG signal with MUAPs of widely ranging size, that the relationship between MU size (defined as twitch or tetanic force) and MUAP size need not be the same for all muscles&at different strategies of MU recruitment and **firing** rate modulation may operate in different muscles (1, 2), and that signal cancellation (between MUAP components of opposite polarity) is bound to depend on the level of motor drive. For any measure of EMG to be a useful indicator of peripheral motor drive, it would have to meet at least two criteria: first, EMG signal magnitude would have to increase monotonically with the ensemble activation rate (EAR) of the pool of MUs and, secondly, the relationship between EMG magnitude and EAR should not significantly depend on the particular recruitment and rate modulation strategy that is used for graded activation of a muscle.

The question of the invariance of the EMG-EAR relation was investigated using an experimental EMG simulation technique, which is based on independently controlled stimulation of large numbers of single MUs in a cat nerve-muscle preparation.

Met hods

The experiments were performed in 14 cats maintained under general anaesthesia. The soleus muscle was prepared with intact nerve supply. The remainder of the limb was denervated. Following laminectomy, the L7 and S1 ventral roots were micro-dissected to isolate functionally single soleus motor axons. Single axons were stimulated supramaximally, either alone (to record their characteristic MUAP shapes; 4 animals) or in combination with other axons (to record multi-channel EMG signals; 10 animals), in either case at constant muscle length. Multi-channel activation was performed with sets of 10 single MUs, with each axon being operated by a separate stimulator.

During combined stimulation, each axon was activated with its own stimulation profile (time course of instantaneous stimulation rate). Ensembles of such stimulation profiles formed multichannel stimulation patterns that were designed to imitate different, physiologically likely MU pool activation strategies: for a pure recruitment strategy, graded muscle activation was achieved by staggered recruitment of MUs at a fixed mean stimulation rate (typically 21/s); a pure rate modulation strategy involved near-simultaneous activation of all MUs at a low rate (7/s), followed by a gradual increase in stimulation rate; combined activation strategies featured narrowly or widely staggered recruitment (at 7/s) combined with rate modulation at varying degrees of augmentation of stimulation rate. All stimulation profiles were defined as pm-edited computer-generated pulse trains, with added Gaussian noise imitating physiological firing variability.

In order to simulate activation of 40 single MUs, four different sets of 10 MUs were stimulated successively, each with its own lo-channel stimulation pattern. The four lo-channel EMG signals were then summed algebraically to obtain estimates of 40-channel EMGs. This summation technique rests on separate experimental simulations, where it was shown that EMG signals generated by activation of up to 40 MUs are the result of strict linear summation of the constituent MUAP trains (Day & Hulliger; submitted).

EMG was recorded with a surface patch electrode stitched to the epimysium. Signal magnitude calculated as a moving window average of the rectified EMG signal (AEMG), using a window duration of 100 ms.

Results

The main fading was that for the cat soleus muscle the relationship between AEMG and EAR was nearly identical for 5/6 MU pool activation strategies that were simulated, including the physiologically unlikely pure recruitment strategy. The only significant deviation was observed for the extreme case of pure rate modulation. For the four - physiologically more realistic - strategies that featured both recruitment and rate modulation in various combinations, the AEMG-EAR relations were indistinguishable.

In all cases the AEMG-EAR relation was monotonic, but revealed significant gain compression (decelerating relation). This gain compression was due to signal cancellation, which increased as peripheral motor drive (EAR) increased. This was shown by the absence of gain compression in AEMG-EAR relations that were calculated as algebraic sums of action potential trains that were recorded during stimulation of individual MUs. These single-MU action potential trains were rectified prior to summation to eliminate signal cancellation. The stimulation profiles and motor units, that were used for single MU stimulation and individual MUAP train recordings, were identical with those used for combined activation of 40 MUs.

During the multi-channel stimulation experiments it was repeatedly noticed that for soleus MUs there appeared to be no strong correlation between MU force and MUAP size. This observation was pursued in experiments specifically designed to isolate and record from as many individual MU as possible. The qualitative observation was confirmed in that for a sample of 250 M from 4 preparations the slope of the relation between MUAP area (AEMG) and isometric tetanic force (at 30/s) was not significantly different from zero.

Discussion

The main advantage of the experimental simulation technique is that it allows EMGs to be recorded under conditions where the peripheral motor command is completely known, as it is defined by the investigator. This made it possible to analyze the relationship between estimates of EMG magnitude and MU pool activation rate experimentally, using recorded MUAP waveforms, rather than theoretically, using artificial waveforms.

The present results show for the cat soleus muscle a simple measure of EMG magnitude is a surprisingly reliable, if non-linear, estimator of peripheral motor drive, regardless of the relative contribution of recruitment and firing rate modulation to the gradation of force. One possible reason for this invariance of the AEMG-EAR relation is the equally surprising lack of correlation between MUAP size and MU size that was found for a large and unbiased sample of soleus motor units. Thus, late recruited large MUs will, statistically, contribute equal amounts to the composite EMG signal as early recruited smaller MUs. For other muscles positive correlations and linear (3) or square root (4) relationships between MUAP and MU size have been reported. It remains to be determined to what extent the AEMG-EAR relation is strategy-dependent in muscles where MUAP and MU size are more strongly correlated.

Differences in AEMG-force relations have previously been attributed to differences in activation strategies (1, 2). In the present series of experiments it was confirmed that the precise shape of the AEMG-force relation is indeed influenced by the MU pool activation strategy, and it was shown that, for the cat soleus, these effects are exclusively due strategy dependent differences in force generation. Thus, the force-EAR, but not the AEMG-EAR relation is sensitive to differences among central activation strategies.

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Acknowledgements

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A Multi function Myoelectric Control Strategy Using an Array of Surface Electrodes

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INTRODUCTION:

Two limitations of current multifunction myoelectric control systems are: (1) the limited information content in the available control signals; and (2) the large amount of data required to choose the desired function. A typical myoelectric control system for an above elbow amputee uses feature information extracted from approximately 250 ms of data from the myoelectric signals (MES) of the biceps and triceps. As a consequence, a maximum of only three degrees-of-freedom can be controlled reliably and there is a delay between function selection which results in a non-continuous and sequential type of control[1][2].

A new approach is introduced which uses the correlation coefficients of a multi-channel array of surface electrodes to define an information vector which represents user intent. The new information is encoded in the crosstalk between electrodes in the array. Preliminary results of this approach based on simulated data are presented.

METHODS:

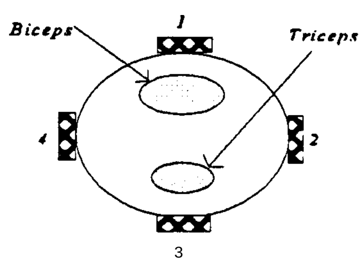

Fig. 1 Electrode Array

Two sets of Gaussian random signals with zero mean and unity variance were generated. These random signals were band limited to dc-1000 Hz and considered to be the simulated MES of the biceps and triceps. Co-contraction signals were generated from the combination of the two signals. Four 4th order Butterworth low-pass filters with different bandwidths were constructed to simulate the aspect of tissue filtering that exists between each muscle and each recording electrode. A set of data for each contraction type measured at each electrode was generated from 1000 realizations of the filtered random signals.

A correlation vector for the m^{th} contraction type is defined as:

$$P(m) = [\Omega(1,2)\ \Omega(1,3)\ \Omega(1,4)\ \Omega(2,3)\ \Omega(2,4)\ \Omega(3,4)\ \omega(1)\ \omega(2)\ \omega(3)\ \omega(4)] \qquad m = 1,2,3$$

where the correlation coefficients are defined as:

$$\Omega(i,j) = \frac{R_{ij}(\tau)_{\mathrm{peak}}}{\sqrt{R_{ii}(\tau)}_{peak} * \sqrt{R_{jj}(\tau)}_{peak}} \qquad ij = 1,\ldots,4,\ i*j; \quad and \quad \omega(i) = R_{ii}(\tau)_{peak} \quad i = 1,\ldots,4$$

and where $R_{ij}(\tau)_{peak}$ is the peak value of the cross correlation between electrodes i, j and $R_{ii}(\tau)_{peak}$ *is the* peak value of the autocorrelation of the ith signal. The P(m) forms the basis vectors for the classification input.

Four techniques were employed to classify the signals. For the rank order classifier the values in the basis vector for each of the 1000 realizations were computed and ranked in ascending order. The index of the structure forms a unique pattern where the signals are clustered into different classes. In this paper, only the first five coefficients of the ordered vector were taken as the class specific pattern

The second classifier tested was the minimum distance classifier with uniform weighting elements. The first 500 realizations were used as the training set and the next 500 realizations were used as the test set. For the fuzzy c-means classifier, the matrix of the initial membership function was chosen based on the apriori knowledge of the pattern exhibited by the basis vectors. Since three distinct classes were to be identified, the partition value was chosen to be 3 and the fuzziness index was chosen to be 2 [3]. Based on the established parameters, the first 500 realizations were used as the training set to obtain the membership function and the cluster center. The next 500 realizations were used to test the classifier. The fourth approach used the Bayes classifier. Again the first 500 realizations were used as the training set to find the standard deviation and the mean and the next 500 realizations were used as the test set.

RESULTS:

The classification rate (in percentage) of different techniques for different lengths of data is summarized in the table below. All strategies classified the simulated signals into three different classes with 100% accuracy if the sampled data was sufficiently long (i.e. n > 200). The rank order classifier gave the poorest result for reasonable data length (i.e. n < 200). The Bayes classifier, the minimum distance scheme, and fuzzy c-means all yielded 100% accuracy using 200 ms of simulated signals and gave reasonable classification rates with as little as 30 ms of data..

# of sample points (n)	5	10	20	30	40	50	100	200	500	1000
Rank Order	32.8	39.0	50.4	55.9	60.0	60.6	75.7	90.4	99.5	100
Min. distance	77.0	80.1	88.9	93.4	95.7	97.3	99.7	100	100	100
Fuzzy c-Means	78.7	85.1	93.7	97.1	98.4	99.1	99.6	100	100	100
Bayes Classifier	79.3	83.1	90.1	96.3	99.3	99.7	100	100	100	100

Table **1. Classification Rate** (in percentage) of different strategic for different lengths of data.

DISCUSSION:

Unlike Doerschuk et al.[4] who used the autocorrelation and cross-correlation approach to reduce the crosstalk between channels before the MES signals were classified, in this new approach the crosstalk information is used to define the basis feature vector. Preliminary results from the simulated data show that the technique can achieve a reasonable classification rate with a shorter amount of data (i.e. 30 ms). If the same result can be shown for real data a more continuous type of multifunction control strategy can be defined.

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Naive Modeling of Control Stategies for Single-joint Movement

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By comparing the behavior of a model system with that of a physical system, we can test our ideas about a physical system's structure and operating mechanisms. This becomes a problematic approach when the input to the physical system is not directly observable. We are then forced to rely on indirect measurements which themselves rely on the model system in order to speculate on what the input might be. In such cases, the input becomes part of the model system itself. In the field of movement control, this has lead to two schools of belief, one which assumes that centrally generated inputs are closely related to the muscle forces that drive a movement while the other assumes those inputs to be more closely related to the movement's trajectory. The former describes planned muscle activation (PMA) hypotheses and the latter, equilibrium-point (EP) hypotheses. Although seemingly different, it may be impossible to distinguish between the two approaches. This presentation describes three models, one PMA and two EP, and how they might control three different movements. One movement is the commonly studied, self-terminated, point-to-point movements (SM). The second is a movement between the same two points but terminated by forceful impact on a mechanical barrier (IM). The third movement goes to the same target and immediately returns (RM) to the starting position.

One of the inherent difficulties in planning voluntary movements is the ill-posed nature of the tasks. Even for a movement between fixed endpoints with a single degree of freedom, the trajectory, the joint torque, the muscle activation patterns and the choice of muscles are all subject to unknown selection criteria which the experimenter cannot control. Some have argued that this can be resolved by optimization rules. We have pursued the approach that designates some of the potentially variable features as specified by rules that for want of a better word, we will call arbitrary. We further assume that the neural control system is highly conservative in the sense that unless there is an obvious reason to change the rules, they will be applied as widely as possible. “Obvious” reasons include instructions given the subject by the experimenter or conformity with Newtonian mechanics.

Applying aconservative PMA hypothesis predicts that joint torques will be similar during acceleration of all three types of movements. In comparison with SM, the acceleration torque of IM will increase to higher peaks at later times since the antagonist muscle is not required generate torque to brake the movement. By similar logic, the acceleration torques of SM and RM will be similar but the deceleration torque o fRM will be larger than that of SM. Applying a conservative rule to one version of the EP hypothesis (EPs, which posits a simple pattern for one of its control variables called R), we expect that the three equilibrium trajectories will be similar until they approach the target. Upon reaching the target RSM stops, RIM continues and RPM returns to the starting position. We make similar conservative speculations abou R of the other form of EP hypothesis (EPc which proposes a more complex R).

We describe experiments in which we measured the kinematics, joint torques and EMG patterns associated with the three movement types. We then use these kinematic and torque variables with simple models that try to predict the EMG patterns. Based upon the assumptions we make (which may not be ones that those who subscribe to EP hypotheses would choose), the control signals of the PMA hypothesis can best be inferred from physical principles. The control signals predicted by EPs are either posited (R) or determined by simulation trial and error (C et. al.). The control signals predicted by EPc use the same logic followed by the PMA hypothesis and may be indistinguishable from PMA. Choosing among hypotheses requires consideration of what kinds of assumptions one wishes to make about the system and what kinds of intuition one brings to bear on the problem. We find that all models are compatible with some features of the EMG patterns and none are compatible with all. That is to say, in order to produce some significant features of the EMG patterns, we are required to modify either our assumed input signals or the structure of the model.

EMG Time-frequency Analysis of Constant and Variable-Force Isometric Contractions

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INTRODUCTION

Recent signal processing developments utilizing time-frequency (TF) analysis have provided a new capability of computing spectral estimates from non-stationary EMG signals recorded during dynamic fatiguing contractions [l]. However, in order to effectively utilize these methods to extract spectral estimates of localized fatigue, it is necessary to study the influence of confounding factors related to changes in contractile force during the dynamic task. Therefore, this study investigated the influences of force magnitude and rate of force production on EMG spectral estimates computed by time-frequency analysis during constant and force-varying contractions in the first dorsal interosseus (FDI) muscle.

METHODS

Seven male subjects [right-hand dominant; age 31 (7.4), mean (SD)] with no history of neurological or orthopedic disorder volunteered for this study after providing written informed consent. The right hand was constrained using a custom device that isolated index finger abduction and measured isometric force via a non-compliant load cell. A custom-designed, flexible, S-channel bar-electrode array was positioned over the muscle belly of the FDI. A monitor provided the subjects with force feedback relative to target force trajectories. Following measurement of a maximal voluntary contraction (100 % MVC), the subjects performed randomized, isometric, constant-force contractions (5 s duration) between 20-80% MVC at 10% MVC intervals. The subjects then performed isometric ramp contractions from O-90% MVC with force slopes of lo%, 20% and 40% MVC/s, also in a randomized order. Each contraction attempt \vas separated by two minutes of rest and the whole protocol was repeated three consecutive times. The signals were digitized with 1024 Hz and recorded on a computer. A moving time-frequency analysis was performed on one electrode channel selected on the basis of location, innervation zone position, and conduction velocity during constant-force contractions. The median frequency was estimated for 125 ms intervals and 50% overlapping for each recording of the selected EMG channel during constant and force-varying contractions. ANOVAs were computed to investigate the influences of contraction force level force slope, and repeatability on the median frequency estimations. Additionally, a time-frequency estimation of constant-force contractions was compared with a classic power spectrum density (PSD) estimation of 500 ms window length and 50% overlapping.

RESULTS

The analysis of variance for both constant-force and force-varying contractions resulted in significant ‘force-level’ effects and interactions. The following effects were statistically significant: force level (p<O.O392), force slope & force level (p<O.O452), force level & repetition (p<O.O054), and force slope & force level & repetition p<0.0114). Main effects of ‘force slope’ and ‘repetition’ were not significant. All data consistently showed an increase in MDF as a function of force level for contractions up to 50% MVC, followed by a decrease in MDF for contractions above 50% MVC (FIGURE 1). These results were the same for both static and dynamic contractions. For contractions reaching force levels above 50% MVC, the MDF for the third repetition was consistently lower than during the first repetition.

A comparison of median frequencies derived from the time-frequency distribution and the PSD during the constant-force contractions revealed different results. The time-frequency analysis pointed to a significant influence of both the force level ($p<O.O012$) and the number of repetitions ($p<0.03\ 10$) whereas the MDF estimations derived from PSDs only indicated a significant influence of the force level ($p<O.O082$).

DISCUSSION

The results demonstrate that changes in force level have a consistent effect on median frequency estimates, regardless of whether the force is increased linearly or via a series of successively higher constant-force contractions. The increase in MDF observed for increases in force up to 50% MVC likely reflects the effects of recruiting higher proportions of type II motor units. These units can raise the MDF value because of their higher muscle cross-sectional area and greater rates of depolarization/repolarization at the muscle membrane [2]. Prior studies have shown that the FDI reverts to rate coding rather than spatial coding as a strategy for increasing force production beyond 50% MVC [3]. The decrease in MDF observed beyond the 50% MVC threshold is most likely a result of reduced conduction velocity and twitch Potentiation that occurs with fatigue. The fact that these findings were consistent for constant and variable-force procedures indicates that confounding factors related to changes in muscle length and electrode position were either not present or of similar magnitude. The procedures, however, differed in that the constant-force protocol resulted in a significant reduction in MDF for repeated trials, indicating that a rest period of two minutes is not sufficient for such a protocol. Interestingly, MDF estimates derived by time-frequency analysis were highly sensitive to this fatigue-effect whereas PSD parameters were not. The superior capability of TF analysis, particularly when combined with electrode array detection systems, is evident from these findings.

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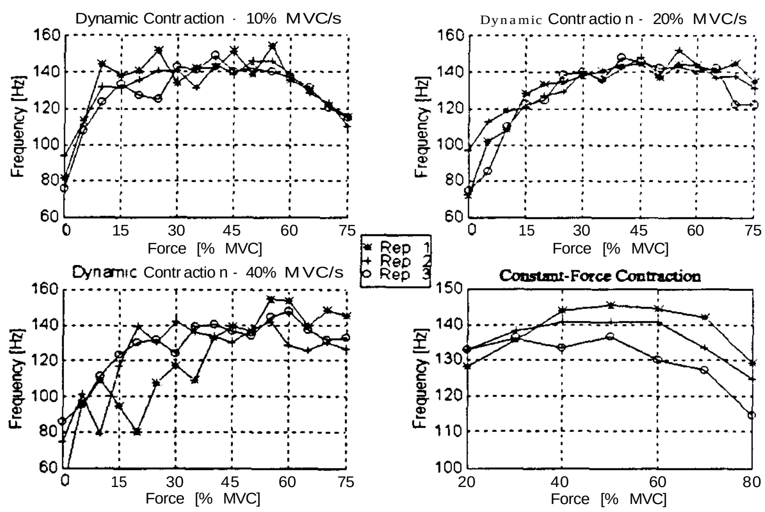

FIGURE 1: The influence of contraction velocity and repetition on the MDF during dynamic (variable-force) contractions at lo%, 20% and 40% MVC/s [plots l-3] and constant force contractions [plot 4].

Time-Frequency Methods for Classification of the Transient Myoelectric Signal

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INTRODUCTION

The myoelectric signal, measured at the surface of the skin, is influenced by an imposing ensemble of factors. The nature of a recorded waveform is determined by the neuromuscular recruitment scheme and firing patterns, the spatial geometry of the active motor units, any mechanical disturbances that may occur during contraction, the properties of the volume conductor, and the effects of the instrumentation. In a given recording environment, it is reasonable to assume that the two latter factors may be assumed constant during different types of muscular effort. If two or more classes of muscular effort are to be distinguished, the information encoded in the former (recruitment scheme, firing patterns, spatial geometry, and mechanical disturbances) may contribute to the distinction. Unfortunately, it is virtually impossible to measure these phenomena independently; we can only gain vague insight via the superimposed activity of the surface MES.

The *steady-state* MES (that produced during constant effort) may be characterized by statistical measures designed to quantify its amplitude (variance, mean-absolute value) or its frequency characteristics (Fourier spectrum, mean/median frequency). This is often not enough to distinguish between classes of muscular effort. The steady-state MES has very little temporal structure due to the active modification of recruitment and firing patterns needed to sustain a contraction. This is due to the establishment of feedback paths, both intrinsic (afferent neuromuscular pathways) and extrinsic (the visual system). In a departure from conventional steady-state analysis, Hudgins [l] investigated the information content in the *transient* burst of myoelectric activity accompanying the onset of sudden muscular effort. It was found that significant temporal structure exists in these transient MES bursts. This temporal structure encodes information important for pattern discrimination; transient MES patterns have demonstrated greater capacity for classification than steady-state signals [l]. Hudgins devised a control system for powered upper-limb prostheses using time-domain features (zero crossings, mean absolute value, and trace length) and a simple multilayer perceptron artificial neural network as a classifier. This controller identified four types of muscular contraction using signals measured from the biceps and triceps. This classifier performs well, but improved classification performance would benefit the functionality and, ultimately, the acceptance of artificial limbs controlled by the MES.

In the quest to improve classification accuracy, one has the choice of improving the classifier or the means of signal representation (the feature set). Although some classifiers demonstrate obvious advantages over others, it is the signal representation that is crucial to robust classification, and this is the focus here. Given that transient MES patterns have significant structure in both time and frequency, it is suggested that the signal energy which would discriminate amongst contraction types would be best concentrated in a dual representation. This work explores the efficacy of feature sets derived from time-frequency representations.

METHODS

Time-frequency representations (TFRs) have received considerable attention in such diverse fields as speech recognition, and the classification of radar, underwater acoustic and geoacoustic signals. Those that have shown greatest utility are the short-time Fourier transform (STFT), the wavelet transform (WT), and the wavelet packet transform (WPT). At the risk of over-simplifying the distinction between linear TFRs, the fundamental difference is in the manner in which they partition the time-frequency plane. The STFT has a fixed tiling, and each cell has an identical aspect ratio. The tiling of the wavelet transform is variable – the aspect ratio of the cells varies such that the frequency resolution is proportional to the centre frequency. This tiling has been shown to be more appropriate for many physical signals, but the partition is nonetheless still fixed. The WPT provides an *adaptive* tiling – an overcomplete set of tilings are provided as alternatives, and the best for a given application is selected. In this application, the "best" tiling was determined as that which maximizes a class separability index [2]. The TFRs used here produce a large number of coefficients, sometimes as large (or larger) than the number of points in the original waveform. This necessitates a scheme of dimensionality reduction; the feature set must be concentrated into a manageable dimension so as not to overwhelm the classifier. The *Kurhunen-Loeve* transform was used to project the TFRs onto their principal components. A small number of the most significant principal component features were then presented to the classifier. Four classes of myoelectric signal patterns were collected from the biceps and triceps; the classes correspond to flexion and extension of the elbow, and pronation and supination of the forearm. Each pattern consists of n = 256 points, sampled at 1000Hz. See [l] for details on the experimental protocol. The data were divided into a training set and a test set, with 80 patterns in each set. To emphasize the effects due to the chosen *feature set,* a simple classifier – a linear discriminant – was used.

RESULTS

Figure 1 depicts the test set classification error upon datasets from ten subjects: seven normally-limbed and three limb-deficient individuals. The results are shown for Hudgins' time-domain (TD) set, and features derived from the STFT, the WT and the WPT. A distinct improvement is visible when progressing from TD to STFT to WT-based feature sets, for both normally-limbed and limb-deficient subjects. The WPT does appear to offer improvement over the WT in the small sample of limb-deficient subjects, but no improvement is apparent with the normally-limbed individuals. This might suggest that the WT offers a time-frequency tiling that is close to optimal for many datasets, but clearly, there are not enough subject data available at this time to indicate statistical significance amongst the feature sets.

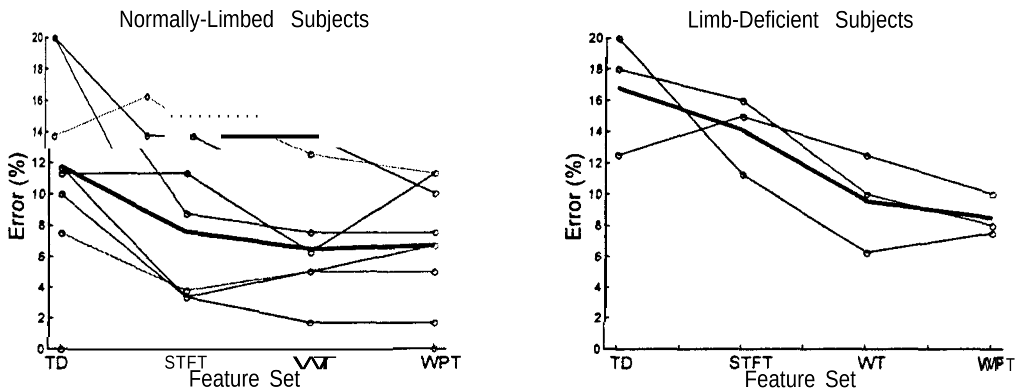

Figure 1 - The test set classification error for normally-limbed and limb-deficient subjects, using TD, STFT, WT, and WPT-based feature sets. The heavy line represents the mean across subjects.

CONCLUSIONS

These results offer insight into a work in progress; the initial findings demonstrate the promise of time-frequency based features, particularly those derived from the WT and WPT. Even with this small sample, it seems that the best feature set is strongly subject-dependent, suggesting the use of a hybrid set of features, or a committee of classifiers – each dedicated to a particular feature set. This, and the task of acquiring a more extensive subject database, is the subject of current work.

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Acknowledgements

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Non-invasive Detection of the Single Motor Unit Action Potential (MUAP)

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Introduction:

The information about the single motor unit action potential (MUAP) is of high relevance, since the motor unit (MU) forms the smallest unit of voluntary muscle activation. The most direct information about the MUAP is expected to be found in monopolar recordings. The problem herewith is to separate the single MU activity from simultaneously active adjacent ones. Commonly, the monopolar recording of the single MUAP can only be achieved by invasive methodologies in which a needle electrode is located close to the MU of interest. Recently, different new recording techniques have been developed which allow the detection of the single MU activity on the skin surface. One approach is based on the combination of a surface electrode with an indwelling single fibre needle-electrode [2,3]. The single MUAP on the skin surface can be detected by averaging the surface-EMG signal in which the trigger is taken from the single fibre activity. However, due to the indwelling trigger-electrode this technique is still an invasive method. The non-invasive separation of the single MU activity becomes possible, if the method of spatial filtering is used [1,4,5]. But, low spatial frequencies which contain important information about the MUAP are suppressed by the used spatial filter. In this study, a procedure is presented which allows the completely non-invasive detection of the single MUAP by combining both, the triggered averaging technique and the spatial filter methodology.

Methods:

An electrode array consisting of two-dimensionally arranged pin-electrodes (diameter 0.5 mm, interelectrode distance 5 mm) has been used to detect the spatial potential distribution generated on the skin surface by an isometric, voluntary contraction of the biceps muscle. Form each electrode the surface-EMG signal has been recorded monopolarly. One spatially filtered channel has been build by a weighted summation of five crosswisely arranged monopolar leads. Such a filter configuration is sufficient to separate the activity of a single MU from the activity of simultaneous active ones, even during maximum voluntary contraction of the muscle [1,4,5]. In the spatially filtered channel the activities of different MUs are represented by isolated peaks. The moment when a particular MU was active has been identified in the spatially filtered channel by means of the maximum peak amplitude, the shape of the peak and the probability for a new activation of the MU. By averaging the different timeslots of one monopolar recorded surface EMG signal, in which the trigger was related to the moment when the MU of interest was active in the spatially filtered channel, the monopolarly recorded MUAP has been extracted. Timeslots of 200 ms were averaged with 100 ms before and after the triggering point. The surface EMG has been recorded over a period of 30 seconds voluntary contraction. Depending on the contraction level of the muscle 15 to 200 peaks of one MU could be identified in the spatially filtered signal and used for triggering the monopolar data.

Results:

Figure 1 shows the MUAP of a single MU recorded monopolarly and completely non-invasively by triggering the surface-EMG signal on a spatially filtered channel. 197 trigger averages have been used to extract the MUAP. Represented are five monopolar channels arranged parallel to the muscle fibres. Additionally, four bipolar channels are shown, which have been recorded simultaneously to the monopolar ones (Fig.1). The MUAP propagates from the proximal end of the electrode array (channel 1) to the distal end of the array (channel 5) with a conduction velocity of 4 m/s.

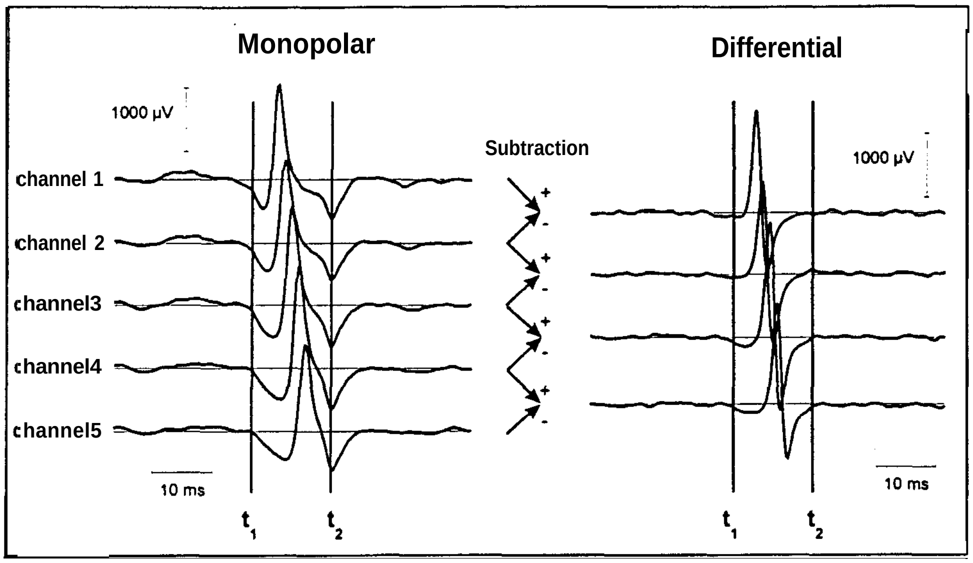

Fig. 1:
Comparison of the monopolar and bipolar recorded MUAP detected on the skin surface of the Biceps muscle. The signals result from 197 averaged timeslots in which the trigger signal has been taken from a non-invasively recorded spatially filtered channel.

In the monopolar signals, beside the propagating component, two stationary components can be identified which are not represented in the bipolar leads. From the needle triggered method and different simulations it is already known, that the first stationary part (t_1) results from the insertion of the MUAP at the motor point and the second stationary part (t_2) from the extinction of the excitation at the tendon insertion.

Discussion:

Triggering the surface-EMG signal on a spatially filtered channel opens access to a monopolar, but nevertheless complete non-invasive detection of the single MUAP during voluntary contraction of the muscle. The methodology supplies in principle the same information as invasive techniques. It allows the localisation of the endplate region and the tendon insertion as well as the calculation of the conduction velocity. Since the single MUAP can be recorded monopolarly, the most direct information about the MUAP can be received. In this way, it overcomes some problems of the spatial filtering technique. However, due to the spatial filter needed for triggering the monopolar signal, the method is limited to superficial MUs, even though it can be used up to maximum voluntary contraction.

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Relationship Between the EMG Mean Power Frequency and the Rate of Isometric Torque Development in Voluntary Contractions

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INTRODUCTION

Central tendency statistics of the electromyographic (EMG) power spectrum (median and mean power frequency (MPF)) have been shown to be sensitive to muscle fiber type composition and muscle fiber cross sectional area (Kupa et al., 1995; Gerdle et al., 1991). Muscles with a higher type II fiber proportion present higher MPF or median frequency values. Because the rate of torque development is also associated with fiber type composition (Hakkenin, 1985), an association between MPF and rate of torque development should be demonstrable.

The purpose of the present study was to examine the relationship between the mean power frequency of the elbow flexors and the rate of isometric torque development.

METHODS

Fourteen healthy subjects between the ages of 23 - 36 years old participated in the study. Subjects were seated with the right forearm resting horizontally in a testing apparatus and the elbow flexed to 90 degrees. The elbow joint axis was approximated at the lateral epicondyle and aligned with a torque transducer. After skin preparation, bipolar, preamplified surface electrodes were placed over the biceps brachii short head (BBS), biceps brachii long head (BBL), brachioradialis (BR) and triceps brachii. Three maximum voluntary isometric contractions (MVC) of elbow flexion were then performed. Target torques of 20,40 and 60% were calculated from the highest torque recorded. These target levels were displayed on an oscilloscope to provide visual feedback. Subjects ***were*** instructed to reach the target torque levels as fast as possible. The torque produced by the subject had to match the target torque within 10%, three times, for each target. In addition, subjects were asked to produce isometric elbow flexion efforts as hard and as fast as they could. The MPF was calculated from 64 ms windows taken from the onset of the EMG signal for each muscle. Rate of torque development was calculated as the slope of the torque curve from 30 - 70 % of a given peak torque (i.e. 20% MVC, 40% MVC, 60% MVC, and maximal efforts). Correlation coefficients were calculated between MPF and rate of torque development expressed as a % MVC / s for each muscle at each target torque.

RESULTS

The rate of torque development gradually increased at different target torque levels, ranging from 194.0% MVC / s to 855.8% MVC / s at the 20% MVC target and maximum efforts, respectively. Significant correlations were found between MPF and rate of torque development for the different target torque levels in the muscles tested (Figure 1 & Table 1). BBL and BBS showed stronger association at lower target torques. Conversely, BR showed higher correlations at higher torque levels. Generally, the relationships were moderate and consistent across conditions. The rate of torque development explained up to 40% of the variability in MPF values. Triceps EMG was minimal during all rapid isometric contractions.

Table 1. Correlations between the MPF and rate of torque development expressed as % MVC / s.

	20%		40%		60%		MAX	
Muscle	r-value	p -value	r-value	p -value	r-value	p -value	r-value	p -value
BBL	0.62	0.02*	0.54	0.04'	0.52	0.06	0.12	0.68
BBS	0.55	0.04*	0.63	0.01 •	0.43	0.12	0.37	0.2
BR	0.08	0.78	0.49	0.07	0.59	0.03*	0.51	0.06*

*Denotes significance at $p < 0.05$

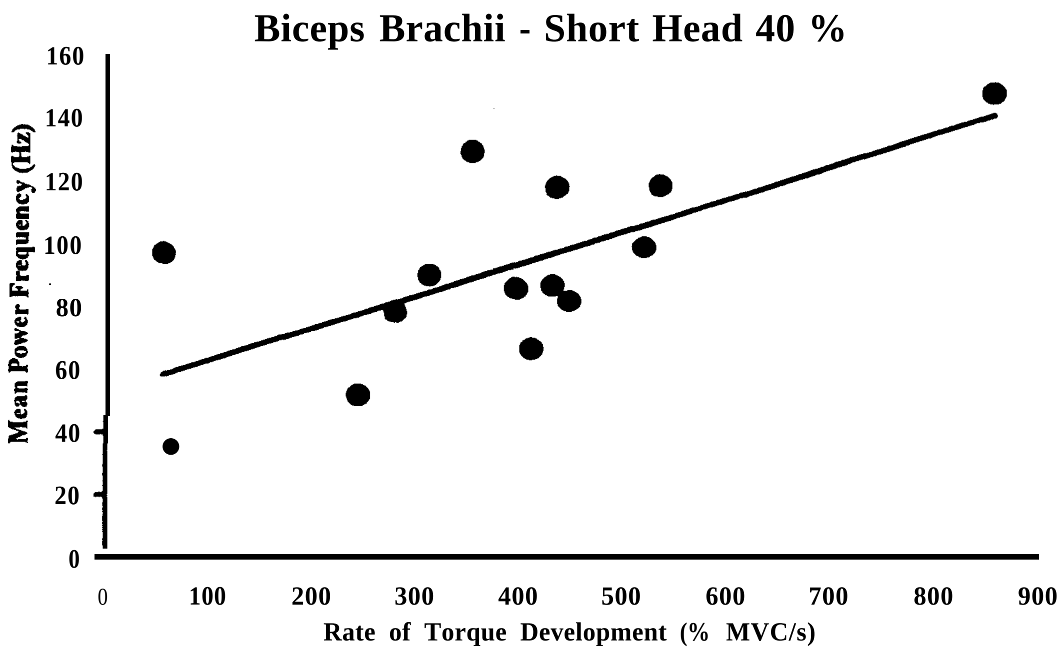

Figure 1. Correlation between MPF and rate of torque development was positive for the biceps brachii short head at the 40% MVC target torque level.

DISCUSSION

The results of this study suggest a moderate relationship between MPF and rate of isometric torque development. Previous studies have shown a strong influence of muscle fiber proportion and size on EMG spectral parameters (Kupa, 1995; Gerdle, 1991). Higher type II muscle fiber proportions and greater muscle fiber cross sectional area are associated with higher EMG spectral parameters. Higher rates of torque development have also been associated with higher type II muscle fiber proportion (Hakkinen, 1985). Therefore, the observed positive correlation between rate of torque development and MPF was expected. However, the moderate strength of the relationship suggests that factors other than muscle fiber type/size will have a significant effect on the rate of isometric torque development.

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Characterization of Surface Electromyograms by Means of Recurrence Quantification Analysis: A Comparison with Frequency Domain Analysis

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INTRODUCTION

Frequency domain analysis of surface electromyograms (sEMG) obtained during voluntary and electrically elicited contractions in humans is a well esperimented technique [l]. It provides information regarding the rate of spectral compression over time during an isometric contraction at a preset percentage of Maximal Voluntary Contraction (MVC) as well as during stimulated contractions at a given stimulation rate [2]. It has also been proposed that from the time evolution of those SEMG power spectrum parameters able to describe the phenomenon of the well known localized muscular fatigue, such as the median frequency (MDF) of the power density spectrum, the modalities of motor unit recruitment can be inierred [3]. It has been often claimed that the analysis in the frequency domain is not appropriate mainly due to the non stationary nature of the signal itself. Moreover, the introduction of transients and discontinuities in the signal strongly often limits the use of the Fourier analysis. In recent years, the application of non linear tools to the analysis of SEMG has been proposed, one such tool is the so called Recurrence Quantification Analysis (RQA) [4]. This approach is appealing because it does not require large sets of stationary data and makes no assumption about the statistical properties of the input data. In the present paper we further explore this topic, with the aim: a) to investigate the possible analogies and /or differences between one of the parameters extracted from RQA, namely the percentage of determinism (%DET) and the MDF; b) to evaluate the effectiveness of %DET to describe SEMG behavior during some isometric tasks. In particular, we tested the sensitivity of %DET to different levels of voluntary muscle activation and the capability of ?/bDET to detect transients hidden in myoelectric signal.

MATERIALS AND METHODS

The biceps brachii (BB) muscle and tibialis anterior (TA) muscle were studied in isometric force constant conditions. First, we evaluated the muscular fatigue in two functionally different muscles, a monoarticular muscle (tibialis anterior) and a biarticular one (biceps brachii), fire different levels of %MVC were considered for the biceps brachii 20%, 40%, 60%, 80%, 100%) and two levels for the tibialis anterior 60%, 80%, A set of three contractions was performed at any given force level. Eight subjects participated in the BB esperiments, while six subjects enrolled into the TA esperiment. The surface EMG signal was recorded by a pair of 1Omm bar electrodes with interelectrode distance of 20mm placed on the skin, respectively 1Ocm far from the elbow joint and 20cm from the ankle joint. The recorded single differential signal was divided in epoch of 1 second. In each epoch the RQA and the Fourier analysis mere applied and then the %DET and the MDF were calculated (as described in [4, 5] respectively) and represented as a function of time. Absolute intercepts and slopes of MDF and %DET were computed from linear and exponential regression over the entire duration of the contraction. Second, in order to study the effects oi an isometric non constant force contraction on surface EMG, the subject had to track one torque target that changed from 2O%MVC to 100%MVC. This task obviously introduced transients in the myoelectric signal. A 10th degree polynomial interpolation was applied to the experimental points of MDF and %DET and the instant in which this polynomial leaves out the confidence interval was detected. This instant was considered as the detected onset time of transient in the force level (4).

RESULTS

In Fig 1 are presented pooled MDF and %DET data (8 subjects, three contractions each) from the linear regression model obtained on BB muscle. It is evident the mirroring trend between MDF and %DET for either slopes and intercepts. The associated standard deviations are always lower for %DET with respect to

MDF. In Fig. 2 are reported MDF and %DET trends when the subject attempted to track the target force. %DET detected the first force transient 3.5 s in advance with respect to MDF.

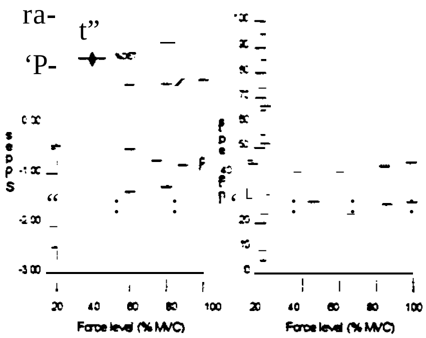

Fig 1 Left: average slope data ±S.D .for MDF (Hz/s) and ?/oDET on biceps bra&ii at five diffacnt force levels (8 subjects), lefl arm; Right: avrage intercept data =S.D. for MDF (Hz) and %DET ca biceps brachii at five different force levels (8 subjects) left arm

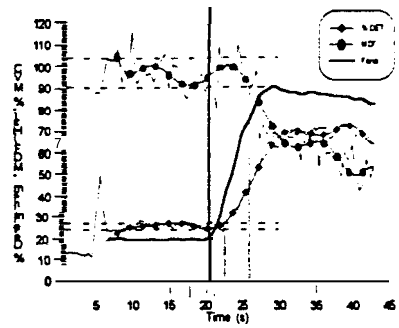

Fig2 %DET (squara) and MDF (circles) during linearly varying isometric contraction (no symbols). Dashed lines represent the 95% confidence interval of the 10th degree polynomial. Vertical lines indicate the time to status change detection.

DISCUSSION

%DET is sensitive to the force level requested to the subject, in the same fashion as is MDF both in term of slopes and intercepts. MDF presents a larger standard deviation with respect to the mean power frequency (MNF) value. Nevertheless, it is interesting that %DET showed a lower variance with respect to MDF. Some preliminary tests not reported in this paper indicates that this is true also when %DET is compared with MNF Also if Webber et al. (4) compared %DET with MNF reporting the same results as those presented here, this result needs a more careful evaluation. From the evaluation of our second esperiment, it may be observed a) that %DET detects a brisk change in muscle mechanical output earlier than MDF and b) that %DET increases until muscle force increases, while MDF significantly varies along time. The first phenomenon was also observed by Webber and coworkers (4) using a different experimental protocol. The lack of variation of %DET during the attempted constant MVC phase (or almost MVC), which contrast with the classical MDF behavior, may be attributed to an independence of the %DET from muscle fiber conduction velocity and/or motor unit firing rates. It is also tempting to intepret the plateau phase of %DET as a consequence of the fact that, as the muscle is approaching his maximal mechanical capability, thus limiting the possibility of alternative solutions, the EMG diminishes its random characteristics, presenting a more deterministic behavior. In conclusion, our results, although a more critical discussion of non linear analysis is needed, confirm that the RQA performed on surface EMG may represent a complimentary technique useful to obtain a better description of the complicated time and force course phenomena occurring during voluntary muscle activation.

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Accuracy of Reconstructing Motor Commands by Means of Pulse Density Demodulation Processing of Electromyograms

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In this paper, a strategy for assessing the accuracy of reconstructing motor commands by means of pulse density demodulation processing of electromyograms (PDD-EMG) is proposed.

RECONSTRUCTING MOTOR COMMANDS BY MEANS OF PDD-EMG

Processes of generation of neural pulses and observation of electromyograms (EMG) are schematized in Figure 1, with special regard to PDD-EMG. Tiie pattern of motor commands which controls degree of contraction of a whole muscle is generated in the motor area, and is transmitted to the muscle through motoneurons as neural pulse trains. The change of the pulse density in each of the motoneurons, if superimposed over a whole muscle, corresponds to the time pattern of the motor commands smoothed by transmission characteristics in the neural passage.

When muscle fibers connected to each motoneuron contract, the action potentials propagate along the muscle fibers. The action potentials are detected with electrodes as the EMG waveform, and the time pattern of the motor commands is reconstructed by measuring time change in pulse density (pulse density demodulation), after correcting the difference in amplitude among pulse trains caused by the attenuation due to the distance from the motoneurons to the electrodes (pulse shaping).

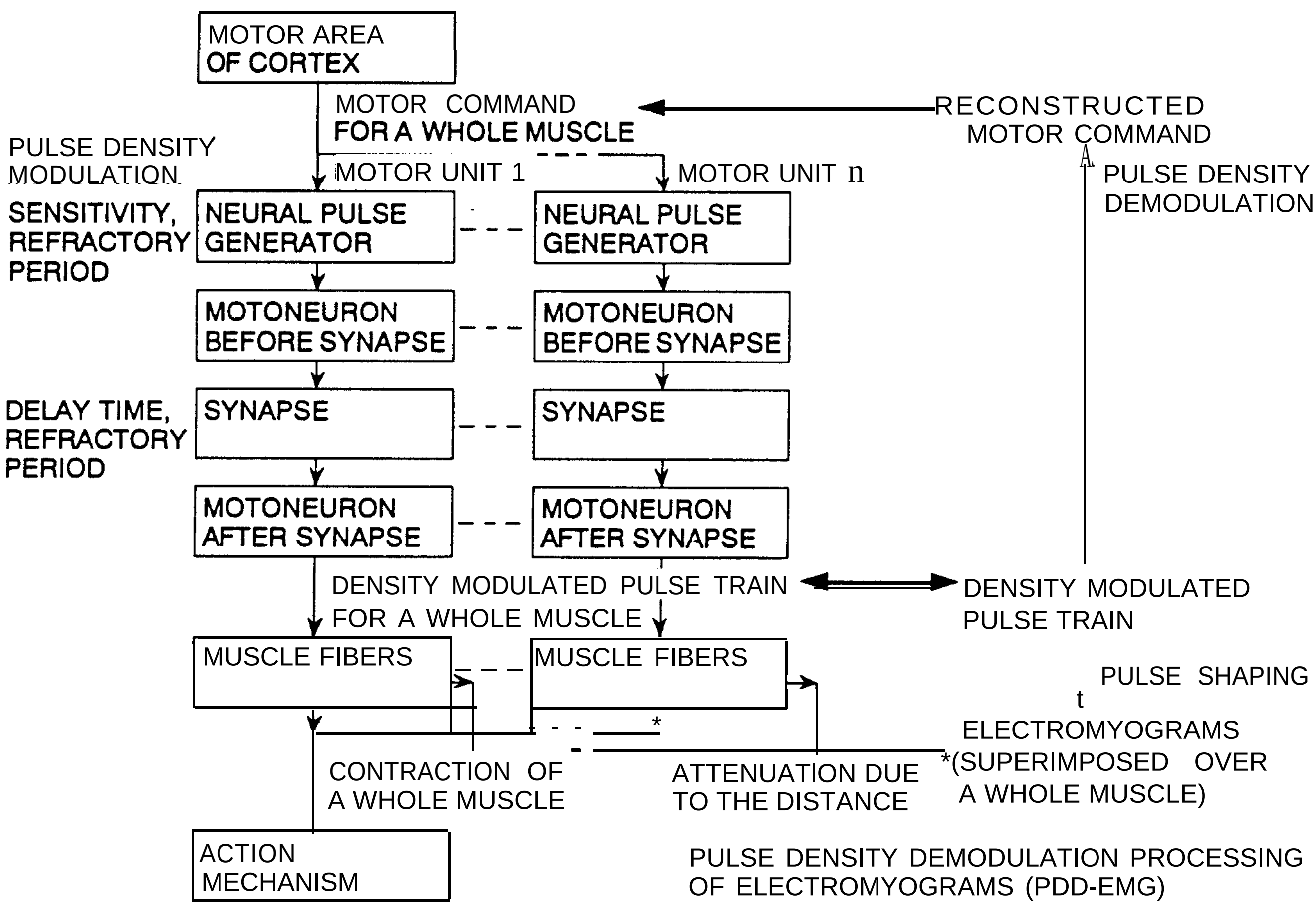

Figure 1. Processes of generation of neural pulses and observation of EMG.

A STRATEGY FOR ASSESSING ACCURACY OF PDD-EMG

The basic idea of PDD-EMG was first realized in the clinical use by an electronic circuitry processing for hooked wire electrode EMG of speech organs (Reference 1). In recent years, it has been attempted to apply it to the computer processing of surface electrode EMG of extremity muscles (References 2, 3, 4).

A preliminary result of PDD processing for surface electrode EMG of lower leg muscles was presented at the previous congress (References 5, 6). The issue raised at that stage was on whether the result of reconstructing the motor command by the processing holds reasonable accuracy, even when the contraction of a muscle is very strong and the pulse density is very high, or not.

In order to assess the accuracy of the processing, in this paper, a strategy using a computer simulation of PDD-EMG is proposed. Between the processes of generation of neural pulses and observation of EMG, there are two signals, namely, the motor commands and density modulated pulse trains for a whole muscle, which correspond each other (bold arrows in Figure 1).

When real EMG were to used as the material, it is not possible to observe explicitly the time pattern of the motor commands and the density modulated pulse trains. However, if all the steps involved in the processes of generation of the neural pulses in the motoneurons and observation of EMG are simulated in a computer, by introducing the neurophysiological properties as real as possible, the known generated motor commands and neural pulses serve as the reference for the detailed analysis of the nature of errors in the result obtained by the processing.

This provides a direct approach to assessing the accuracy of PDD-EMG. In addition, the process of computer simulation can be modified in any complex way, if necessary for further analysis.

A computer simulation proposed in this strategy has been executed preliminary, and a promising result is reported by the present authors in this congress.

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Electromyography of the Paraspinal Muscles in Response to Mechanical and Electrical Stimulation of the Supraspinal Ligaments in Humans and Cats

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INTRODUCTION

It is hypothesized that a permanent source of low back pain arises from prolonged periods of incorrect posture which deforms the posterior spinal ligaments. Mechanoreceptors in the ligaments, in turn, become active, and initiate contraction of the paraspinal muscles in order to correct the deformation. The muscles, over time, fatigue, occlude circulation, develop spasms and pain. Rest was recently shown to resolve low back pain of this etiology.

It is the objective of this paper to test the hypothesis that a ligamento-muscular spinal reflex exists and could initiate activity in the paraspinal musculature.

METHODS

Six human subjects, candidates for spinal surgery, and twelve adult cats, both anaesthetized with non-paralyzing agents were used. After the initial superficial skin incision over the lumbar spine, the connective tissues were cleaned gently to fully expose the lumbodoral fascia, which was cleaned with saline.

In humans, needle electrodes were inserted bilaterally into the paraspinal muscles one and two levels above the level of injury. In cats, intramuscular wire electrodes were inserted bilaterally in the paraspinal muscles from T-l to T-7 (cat has 7 lumbar segments). Initially, a bipolar stimulating electrode probe was applied over the Supraspinal ligament delivering supramaximal stimuli at 10 pulses per second. The probe was applied over each ligament segment separately while EMG recorded from the muscles.

Than, a ligature was inserted around the ligament, and attached to a force transducer. The ligature/transducer were pulled up to deform the ligament while EMG and pull force were recorded.

RESULTS

Electromyographic activity was recorded in response to the ligament stimulation in both paradigms. M-wave trains corresponding to the stimulation frequency was recorded with the highest peak to peak amplitude at the level of stimulation, and with lower magnitudes as far as two spinal levels above and below the stimulated level. Mechanical deformation of the ligament in humans and cats resulted in EMG discharge from the paraspinal muscles also with highest magnitude in the level of stimulation, and gradually lower magnitude in the two levels above and below. The peak to peak and mean absolute value of the EMG was related to the deformation force applied. The EMG activity persisted as long as ligament deformation was present.

CONCLUSION

Mechanical deformation of mechanoreceptors in spinal ligaments seems to excite, via spinal neural loops the paraspinal muscles in attempt to oppose and correct the deformity. Such reflexive activation is the strongest in the muscles corresponding to the level of deformation and weaker as far as two levels above and below. Prolonged deformation resulted in prolonged activity in the muscles, indicating a robust neurological and mechanical synergy between the muscles and their corresponding ligaments in the lumbar level. The possibility of development of low back pain due to prolonged activation of the paraspinal muscles by the ligaments is evident.

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Repeatability of EMG Timing Parameters During a Dynamic Activity

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INTRODUCTION

A prerequisite of any experiment is that the data collected must be reproducible. One of the disadvantages of Electromyography (EMG) is repeatability. Various factors exist including electrode position, sweat during testing, subcutaneous fat tissue, movement of the muscle under the skin to name a few which affect the EMG signal and make it difficult to reproduce the same signal twice. Some of the associated problems can be controlled, such as electrode position, though according to Basamajan and De Luca (1985) when wire electrodes are implemented a diversion of only 1 00μm can decrease the peak to peak amplitude of the output signal by 75%. More recently Veiersted (1991) reported a Coefficient of Variation of 16.8% during submaximal exercise, when electrode position was altered from their predefined zero position to 3 mm either side of this mark.

Repeatability studies have been performed in the past by Komi and Buskirk (1970) who investigated both surface and inserted wire electrodes. They reported that repeatability of IEMG was better between recordings made during the same testing session than when the tests were spaced temporally. Similarly, though more recently Gollhofer et. al. (1990) elaborated on past research by using stretch shortening type of contractions. They reported reliability coefficients exceeding 0.85 for the amplitude of EMG. In addition they reported the pattern of EMG activity to be stable, though neglected to quantify these results.

The aim of this study was to investigate the repeatability of EMG timing patterns of seven leg muscles during a highly dynamic task when tests are spaced temporally. A knowledge of the reliability of repeated tests of EMG temporal characteristics would aid in preventing any false or misleading interpretation of the origin of movement patterns.

METHODS AND PROCEDURES

The ten subjects who participated in this study had a mean age of 27±2 years. Their average weight and height were 80±4 kg and 176±4 cm respectively. All had participated in some form of sprinting before though at the time of testing none were training regularly.

The study was performed over a period of five weeks, with 6 testing sessions performed by each subject. After an intensive warm-up, the subjects were asked to perform six maximal sprints with a five to 10 minute rest period between trials.

EMG recordings, simultaneously with acceleration data to determine foot strike, were taken of seven major leg muscles. These included medial hamstings (m. semitendinosus and m. semimembranosus), m. biceps femoris, m. rectus femoris, m. vastus lateralis, m. vastus medialis, m. gastrocnemius and m. tibialis anterior. Prior to electrode application, the skin was shaved, cleaned with alcohol and sandpapered if necessary, to achieve an impedance below 5kR. Bipolar electrodes (blue sensor disposable electrodes N-00-S) were placed according to Winter (1980) with an interelectrode distance between 2 and 3cm. Blue sterile ink was implemented for tattooing the skin at the electrode sites. The sampling frequency was set at 1000Hz. To ensure that movement artifacts were kept at a minimum the electrodes were taped to the skin with elastic surgitape and the cables held in place by skin fitting tights worn by the subjects.

Opto-electric timing gates were implemented to determine the velocity of the subjects, and only those trials where horizontal velocity was within a 2% standard deviation were used for further analysis. A minimum of three trials were analysed for each subject.

RESULTS

The average speed attained by the subjects was 8.42 ● 0.32m/s with an average stride time of 0.49 ± 0.03ms. Results of the EMG activation pattern across tests within one test day revealed no significant differences for all seven leg muscles investigated. Similarly no significant differences could be found when data was compared within subjects across testing days. Figure one displays typical EMG timing patterns of one leg muscle across six testing sessions for subject 3.

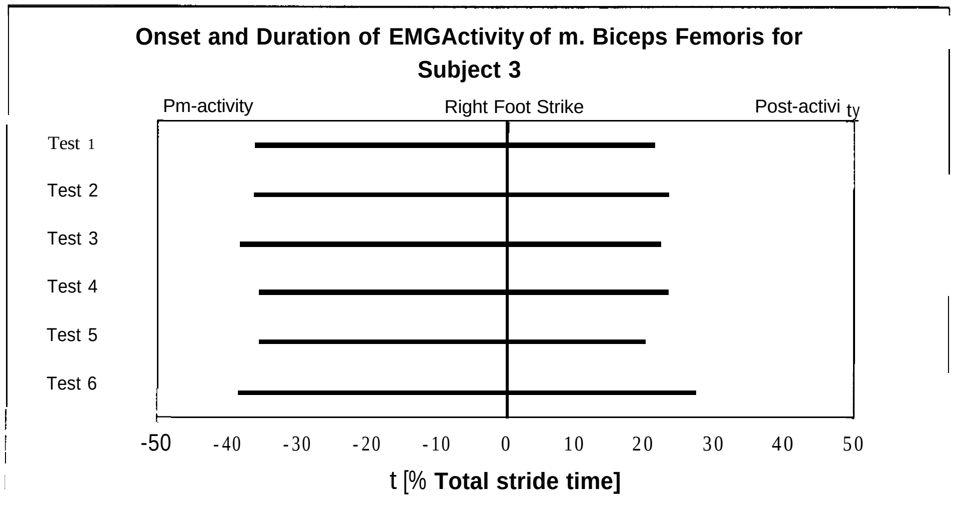

Figure1 : Typical EMG timing pattern for m. Biceps Femoris for Subject 3.

DISCUSSION

This paper investigated the reliability of EMG activation patterns for seven lower limb muscles during a dynamic sprinting activity. No significant variability was found between individual tests on all testing days or between different testing sessions. It is therefore concluded that surface EMG is an appropriate method for testing subjects when timing patterns are examined.

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Raising Eyebrows: The Relationship Between Muscle Activity and Displacement

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INTRODUCTION. Although physical therapists often use integrated surface electromyography (SEMG) as biofeedback to help patients with facial paresis to regain motion, the relationship between the SEMG and the motion of the face during expressions is incompletely described. One study, by Burres, described a linear relationship (r=0.97) between the log of SEMG at midface and the change in distance divided by the rest distance (percent displacement), for three positions (nose wrinkle, smile and eyes closed tight) in thirty normal subjects.' However, since these measurements were taken in the mid cheek area, multiple muscles acting in concert may have influenced the final motions seen. In addition, electrical "cross talk" of the muscles may have influenced the SEMG recorded.

We sought to determine the relationship between facial muscle electrical activity and extent of skin displacement under circumstances in which the actions of a single muscle could be studied. The action of the frontalis muscle raising the brow was chosen because it is a site where both electrical cross talk and unrelated surface movements are minimal. We chose to measure skin displacement as a consequence of a series of designated levels of muscle electrical activity to determine if a correlation could be characterized.

MATERIALS AND METHODS. Seven volunteer subjects (average age 26.1 years) without a history of facial nerve disorder or other major medical problems participated. Bipolar, 1cm diameter, pregelled Ag-AgCl surface electrodes were used to measure frontalis muscle activity. Quantification of the SEMG recording of muscle activity during voluntary contractions was determined with Neuro Educator II®[a] which uses full wave rectification, real time processing and lOO-msec time constant integration averaged over 0.3 seconds. A video-based variation of the Maximal Static Response Assay[2] measured displacement by subtracting the position of brow markers during rest, from those during brow elevation. Splicing the images of both the Neuro Educator II screen and a video of the subject's face using Videonics Digital Video Mixer®[b] allowed selection of frames of the video to correspond to exact times of the SEMG.

While looking at the visual SEMG feedback, the subject raised his/her eyebrows for a five second maximal voluntary contraction. The level of maximal SEMG was approximated from the visual recording; and three quarters, one half, and one quarter of the maximum were calculated. For each of these four levels, the subject raised his/her eyebrows for a five second voluntary contraction at that level of SEMG. Three trials were recorded, after a practice attempt, for each level with a ten second rest between trials.

In order to standardize the range of SEMG and displacement between subjects, the percent SEMG and percent displacement were also calculated for each subject. Percent SEMG equals the SEMG divided by the subject's maximal SEMG; and percent displacement equals the displacement divided by the subject's maximal displacement.

RESULTS. The results of the average of the seven subject's SEMG and displacement data are shown in table 1. In order to find the strongest relationship between SEMG and displacement, Pearson correlation coefficients [r] were determined for the relationships shown in table 2. The plot of percent displacement versus percent SEMG is shown in figure 1. The line of best fit has a slope of 0.8975, a x intercept of 0.06 and a y intercept of -0.06.

Table 1.

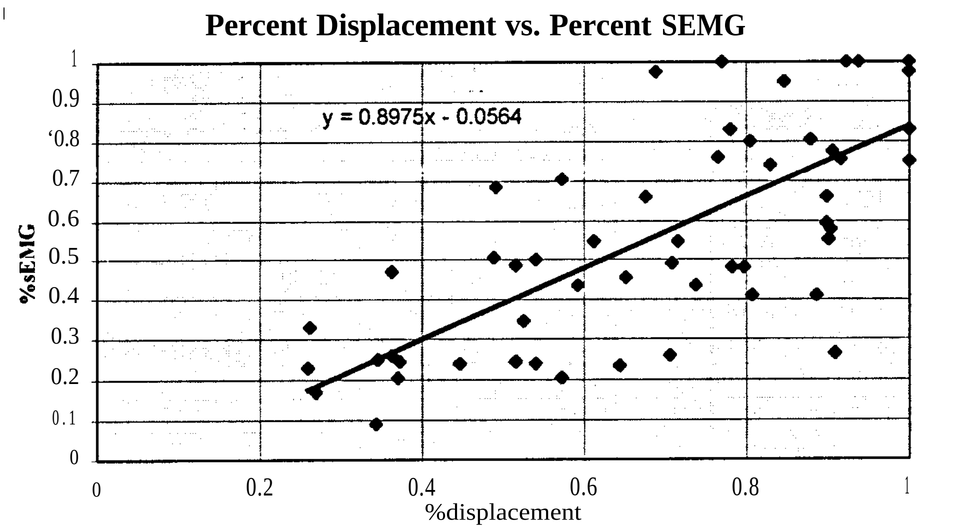
sEMG in microvolts	average	range	SD	C.V.	% diff. r&l
Maximum	56.4	35.9-87.6	5.8	0.1	16.6
Three quarters	43.5	19.8-70.0	5.1	0.12	19
One Half	29	14.8-44.4	4.1	0.14	16.4
One Quarter	15	5.6-23.6	1.8	0.12	22.2
displacement in centimeters					
	average	range	SD	C.V.	% diff. r&l
Maximum	0.98	0.44-1.61	0.07	0.07	16.1
Three quarters	0.87	0.36-1.30	0.05	0.06	16.3
One Half	0.65	0.35-0.99	0.05	0.08	21.5
One Quarter	0.46	0.29-1.13	0.07	0.17	30.2

Table 2.

Pearson Correlations		
x-axis	y-axis	r value
% displacement	% sEMG	0.738
% displacement	% logsEMG	0.737
% displacement	logsEMG	0.668
% displacement	sEMG	0.607
displacement	% sEMG	0.534
displacement	% logsEMG	0.531
displacement	logsEMG	0.429
displacement	sEMG	0.343

Table 1. The average and range of SEMG and displacement for the seven subjects are shown. The standard deviation (SD) is the average of the SDS for each subject. The coefficient of variation (CV) is the SD divided by the average. The percent difference between the right and left sides (% diff. r&l) is the average of the % diff. r&l of each subject.
Table 2 The Pearson correlation coefficient are ranked from highest to lowest.
Figure 1. The plot of percent displacement vs. percent sEMG has 56 points derived from 7 (subjects) x 2 (right and left) x 4 (levels of sEMG).

Figure 1.



DISCUSSION. The subjects were able to maintain a relatively constant SEMG over the five second intervals, as evidenced by the CV of 0.14 or less at all levels of SEMG While the percent difference of SEMG between sides appears to be roughly the same at low or high levels of SEMG, there appears to be a larger percent difference of displacement between sides at lower levels of SEMG. The Pearson correlation coefficients show that the best correlation is between the percent displacement and percent SEMG, and the least correlation is between distance and SEMG, implying that there is a linear relationship between the amount of muscle electrical activity and displacement of the eyebrows, but that the relationship varies from person to person.

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SUPPLIERS. a Neuro Educator II; Therapeutic Alliances, Inc., 333 N. Broad St., Fairborn, OH. b Videonics, Inc., 1370 Dell Avenue, Campbell, CA 95008 USA

Analysis of Muscle Coordination and Function During Cutting Movements

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Introduction

Ankle sprains are arguably the most common injury in sport (Maehlum and Daljord, 1984) and can lead to significant impairment characterized by functional instability of the ankle joint (e.g. Yeung et al., 1994). Ankle sprain injuries often occur during cutting-type movements (e.g. Garrick, 1977), which to date, have received little attention in the literature. Previous ankle studies have concentrated primarily on EMG activity of muscles crossing the ankle joint (e.g. Johnson and Johnson, 1993). But, recent studies have indicated that muscles crossing the hip and knee joints may play an important role in stabilizing the ankle joint (e.g. Bullock-Saxton, 1994). No study has combined entire lower extremity kinematic and EMG data to study mechanisms leading to ankle sprains. Considering the dynamic coupling between segments of the lower extremity, an analysis of the entire lower extremity including kinematic and EMG data would provide the data necessary to understand basic muscle function and coordination principles during movements susceptible to ankle sprains. A better understanding of these principles may provide insight into mechanisms which lead to ankle sprain injuries and provide the foundation for future studies.

Therefore, the objectives of this study were twofold. The first objective was to establish a database of kinematic and EMG data to provide a foundation for future experimental and theoretical studies investigating muscle coordination and function during cutting movements. The second objective was to combine the kinematic and EMG data to describe normal muscle function and coordination during these movements.

Methods

To achieve the above objectives, ten recreationally active male subjects volunteered for participation in this study. Informed consent was obtained before the data collection. Data were collected during two different movements, a side-shuffle and a **45** degree forward v-cut. EMG data were collected from twelve lower extremity muscles on the right leg using pairs of surface electrodes. The data were processed to identify the muscle excitation pattern and timing. Five trials were collected for each movement. Ground reaction force (GRF) data were collected simultaneously with the EMG data at 2400 Hz using a force platform (Kistler Biomechanics, Switzerland). Lower extremity joint kinematics were determined using a high speed video analysis system (Motion Analysis Corp., Santa Rosa, CA) from retro-reflective markers placed on each limb segment. Limb segment orientations were reconstructed from the smoothed marker positions using the method of Soderkvist and Wedin (1993). Hip and knee joint angles were then determined using the joint coordinate system of Grood and Suntay (1983) and ankle joint angles were determined using the joint system of Inman (1976).

Muscle functions were determined by comparing the joint kinematics with EMG activity during each movement.

Results

EMG patterns and joint kinematics were consistent across subjects and the muscles functioned similarly during both movements. Examples are shown in Figure 1. The primary function of the hip and knee extensors was to decelerate the center-of-mass during impact and to provide propulsion during toe-off. The hip add/abductors functioned primarily to stabilize the hip rather than provide mechanical power. The ankle plantar flexors functioned similarly during the two movements by providing propulsion during toe-off except for the gastrocnemius which had an additional burst of activity to plantar flex the foot prior to impact during the side-shuffle, possibly to help absorb the impact. The ankle dorsiflexor function was different for each movement: to dorsiflex and supinate the foot after toe-off during the shuffle, and to dorsiflex the foot prior to impact to provide a heel-first landing and ankle stability in the stance phase during the v-cut movement,

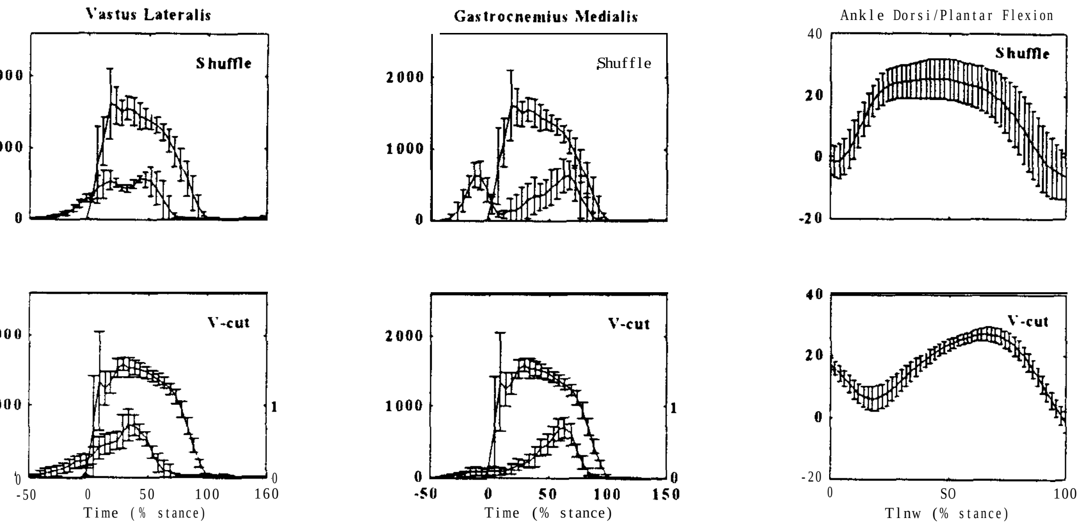

Figure 1: Normalized EMG patterns of two lower extremity muscles for a typical subject during the side-shuffle and v-cut movements. The EMG data is normalized to the maxinum value achieved during both movements. The units for the ground reaction force data is Newtons and the units for the joint angles are degrees. Positive angles indicate dorsiflesion. The average stance phase time was 0.266 and 0.329 seconds for the side-shuffle and v-cut movements. respectively.

Discussion

Ankle sprain mechanisms are not well understood due to a lack of data regarding muscle coordination and muscle function during cutting-type movements. This study has provided the necessary kinematic and EMG data necessary to design experimental and theoretical studies to further understand ankle sprain mechanisms. Theoretical studies using forward dynamic simulations incorporating individual muscle actuators are necessary to fully understand muscle function and coordination, and the segment accelerations induced by active muscles which may prevent or lead to injuries. Several example applications of the experimental data used for development and validation of computer simulation experiments will be presented.

Acknowledgments

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EMG Activities of the Popliteus and Hamstring Muscles During Maximum Isokinetic Knee Flexion

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INTRODUCTION

It is well known that the popliteus muscle acts as an internal rotator at the initial phase of knee flexion and thus unlocks the knee (1,2,3,4). It has been also reported by many investigators that knee flexing action is performed by hamstring muscles. However, there are few studies concerning to the relationship between the knee angle and the activities of hamstring muscles. Therefore, we examined the role of the Popliteus and hamstring muscles in the knee flexion by analyzing EMGs during the maximum isokinetic knee flexion.

METHODS

Ten normal volunteers (10 male, aged from 21 to 36 years) participated in this study. Every trials were performed on prone position. The subject performed maximum isokinetic (30 degree/sec) knee flexion ranged from 0 to 120 degree, using isokinetic dynamometer (KINCOM, USA). Then EMG signals of the Popliteus, semitendinosus, semimembranosus, and long and short heads of the biceps femoris muscles were detected through the bipolar Teflon-coated stainless steel wire electrodes (AM system, USA). The diameter of each electrode was 75 μ m. 2mm of the electrode tip was deinsulated and inter-tip distance was set at 5mm(5). The electrodes were inserted into the muscle with a 25-gauge needle. The electrode placement was confirmed by electrical stimulation. The EMG signals and torque were simultaneously recorded and digitized at a sampling rate of 5kHz with 12bit A/D converter (AD12-16U(98)EH, Contec, JAPAN). The EMG signals were full-wave rectified and integrated (IEMG) at knee flexion ranges from 0 to 15(0-15), 15-30, 30-45, 45-60, 60-75, 75-90, 90-105 and 105-120 degrees. IEMG values of the popliteus were normalized with the value during maximum isometric internal rotation of the tibia with 90 degree in knee flexion and neutral ankle position. IEMG values of the hamstrings were normalized with the value during maximum isometric knee flexion with 90 degree in knee flexion.

RESULTS

During isokinetic knee flexion with maximum force, the activity of the Popliteus was always observed. The normalized IEMG value of the Popliteus at the initial phase of knee flexion from 0 to 15 degrees was significantly larger than that at the next phase from 15 to 30 degrees ($p<O.05$). As shown in Fig 1, the activity of the long head of the biceps femoris at the early phase of knee flexion showed about 150% of the standard IEMG at 90 degrees of knee flexion and then decreased with an increase in the knee flexion. In parallel with these EMG changes, the knee flexion torque decreased with an increase in the knee flexion. By contract, the activities of the other hamstring muscles increased significantly with knee flexion from 0 to 90 or 105 degrees ($p<O.05$) and then decreased.

DISCUSSION

It has been reported that the Popliteus is active at the onset of knee flexion and unlocks the knee. Our result of the Popliteus activity during knee flexion is coincident with this finding.

The flexion torque of the knee with maximum isokinetic flexion decreased in parallel with an increase in flexion angle of the knee. Normalized IEMG vs flexion angle of the knee indicates that the

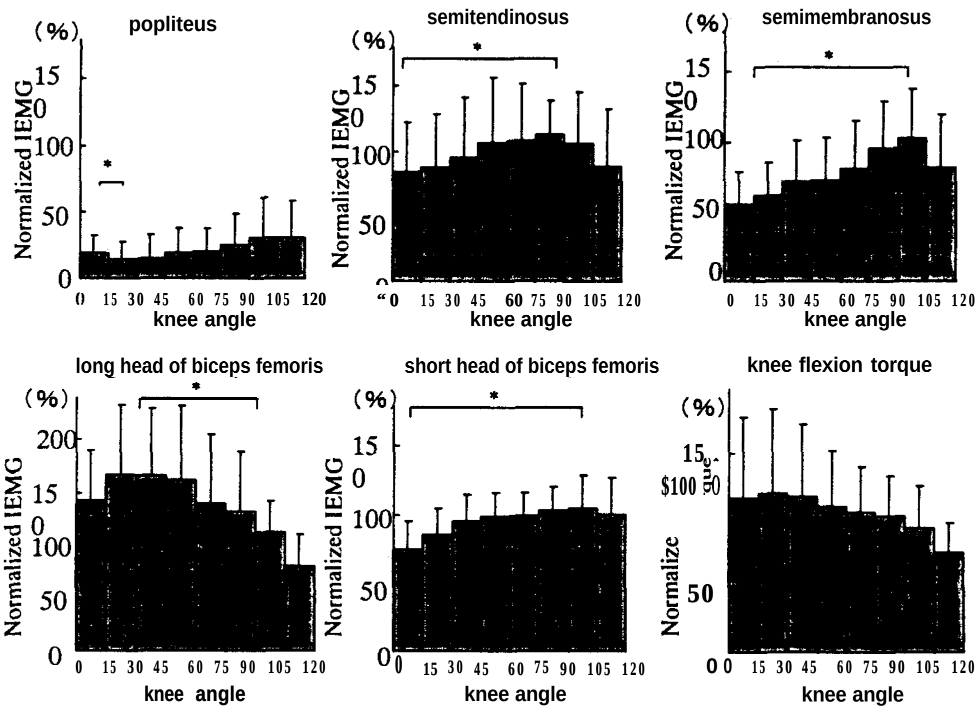

Fig. I IEMG values detected from the popliteus and hamstring muscles and knee flexion torque during maximum isokinetic voluntary knee flexion (n=IO, * : P<O.05)

humstring muscles can be classified into two groups. One is the long head of the biceps femoris which was most active during the early phase of knee flexion and showed a decrease in its activity afterwards. This activity changes seems to be parallel with the torque changes during maximum isokinetic knee flexion. Another is the semitendinosus, semimembranosus and short head of the biceps femoris which became more active as the flexion angle increased. These results may indicate that the long head of the biceps femoris act as a most important prime mover in this movement under the condition of maximum isokinetic knee flexion.

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Synergistic Activation Patterns During Isokinetic Dorsiflexion

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INTRODUCTION

The purposes of this study were to compare the relative level of activation recorded from multiple sites over the Tibialis Anterior (TA) and the Triceps Surae muscles during maximal isokinetic dorsiflexion, and to evaluate the effect of angular velocity on the level of activation measured by surface electromyography (SEMG). How the agonists and antagonists are recruited is important to determine the value of isokinetic dynamometry which has been used to evaluate dorsiflexor muscle performance (1). In particular, Coactivation of the antagonist muscles has not been extensively studied and is important since the net moment of force recorded is affected by the level of antagonist activation. The Triceps Surae muscles are the main antagonists, and are comprised of three muscles with anatomical and physiological differences. Studies of plantarflexion tasks illustrated differences in the activation amplitudes of the individual components of the Triceps Surae muscle group during the performance of various tasks (2), supporting the need to record from the three muscles. Regional differences in activation levels have been reported from different electrode sites on the same muscle (3) illustrating that a single electrode site may not accurately reflect the activation of the entire muscle, in particular for large muscles. The large surface area of the Soleus, and the length of the TA muscle provided the rationale to include two recording sites for each of these muscles.

METHODS

Ten healthy subjects (2 males, 8 females, mean age 30+/- 7.5 years, height 166+/- 4.7 cm. and mass 63 +/- 12.8 kg.) participated in the study. The skin above the electrode sites was prepared by shaving and rubbing the skin with an alcohol and water solution to reduce skin impedance. Meditrace silver/silver chloride (ECE 1801, 1Omm) electrodes were placed in a bipolar configuration at an upper (UTA) and a lower site (LTA) over the TA and at four sites over the antagonist muscles: the Lateral (LGA) and Media1 Gastrocnemius (MGA), plus a lateral (LSO) and a medial (MSO) site over the Soleus muscle. The raw EMG signals were amplified (Bandpass lo-500 Hz, CMRR= 90db and input impedance = 100 Mohms), then analogue to digitally converted at 1000 samples per second and stored on floppy disk for further processing. The knee angle was 170 degrees (°) and the ankle joint moved from 30 degrees of plantarflexion to 5 degrees of dorsiflexion. Raw SEMG data were recorded for two maximal voluntary isometric contractions (MVIC) of the dorsiflexors and plantarflexors for normalization purposes. Then subjects performed three maximal isokinetic dorsiflexion contractions against a Cybex II (Lumex,Inc, NY) dynamometer at three angular velocity settings: 30, 90 and 150 degrees per second. The order of the trials was randomized. The raw data were windowed from the beginning of movement to the end of movement based on the angular displacement curve and the root-mean-square amplitude (RMS) was calculated for each electrode site. The data were then normalized (RMSNji) to the MVIC. A two factor repeated measures analysis of variance (Velocity X Muscle Site) tested main effects and interactions, using *a* of 0.05. Ensemble average curves were calculated for each EMG site to provide a profile of the activation over the duration of the movement.

RESULTS

The EMG amplitudes for each electrode site are presented in Table 1. The ANOVA indicated a statistically significant muscle by velocity interaction, therefore, differences among muscles and differences among velocities were tested separately using a Bonferonni correction (df=90,alpha=O.OOl,tcrit =3.15). The post hoc analyses demonstrated that this interaction was mainly due to the influence of velocity on the LTA amplitudes; the RMSNji of the LTA during the 150 degrees per second movement, was greater than that recorded at the other two velocities. The RMSNji between the two TA electrode sites were not different from each other for any of the three velocity settings, nor was the LGA different from MGA for the 30 degrees per second velocity. All other between muscle comparisons demonstrated statistically significant differences.

Table 1: Normalized EMG amplitude (RMSNij) recorded from each of six muscle sites during maximum isokinetic dorsiflexion. Mean amplitudes (SD) for all subjects (N = 10)

		LGA	MGA	LSO	MS0	UTA	LTA
30 °/s	Mean	26.3	22.4	59.1	51.6	95.2	92.7
	(SD)	(10.9)	(12.5)	(34.4)	(26.9)	(25.9)	(16.5)
90 °/s	Mean	28.8	21.9	60.4	52.8	99.0	97.3
	(SD)	(12.4)	(16.0)	(49.6)	(33.3)	(23.7)	(17.8)
150 °/s	Mean	29.3	20.9	59.8	52.2	100.0	103.2
	(SD)	(13.2)	(14.8)	(45.3)	(33.9)	(28.4)	(19.4)
Grand	Mean	27.8	22.5	59.2	52.4	98.2	98.0
Mean	(SD)	(12.3)	(15.1)	(42.8)	(30.9)	(25.7)	(18.3)

DISCUSSION

The TA muscle was activated to a consistently high level of MVIC for both sites, which is indicative of a consistent level of recruitment within the muscle. The increase in amplitude for the LTA site with an increase in velocity was attributed to an inability to maintain the high level of activation over the longer movement time associated with the slower velocity. The high level of Coactivation, in particular for the two Soleus sites, supports a Coactivation motor pattern for this movement. The difference between the LSO and MS0 may reflect a difference in recruitment within the muscle as was demonstrated for the elbow muscles(3).The differences between the Gastrocnemius and the Soleus muscle, with the Soleus sites much higher than the Gastrocnemius sites is considered an important finding. This may be attributed to the physiological and anatomical differences between the two muscles, as well as the functional differences clearly demonstrated through recording EMG during plantarflexion and knee flexion tasks(2). Consistent with studies of isokinetic exercises at other joints(4), the velocity of movement did not alter the amplitude of Coactivation. The Coactivation amplitudes found in this study would not support interpreting the isokinetic torques as reflecting dorsiflexor muscle function only. In conclusion, the results of this study show a consistent level of activation recorded from within the TA, with differences among the activation levels from the four Triceps Surae sites during maximal isokinetic dorsiflexion, with minimal differences attributed to the change in velocity.

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The Use of Electromyography in Applied Physiology

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Surface electromyogram (EMG) analysis in studies of muscle function has attained increasing attention during recent years and has been applied to assess muscle endurance capacity, anaerobic and lactate thresholds, muscle biomechanics, motor learning, neuromuscular relaxation, optimal walking and pedalling speeds, muscle soreness, neuromuscular diseases, motor unit activities, and skeletal muscle fatigue. Intramuscular EMG recordings has also been employed in many studies to assess motor unit (MU) recruitment and MU firing rate modulation (rate coding) under a variety of experimental conditions.

On the other hand, analyses of reflex or electrically induced potentials have recently gained its popularity. For examples, using the Hoffmann (H-) reflex paradigm it has been possible to determine changes in the gain of the monosynaptic stretch reflex during various motor tasks in man. Evoked potential analysis has also been employed for studying muscle membrane excitability, myotatic reflex and spinal motoneuron excitability.

This presentation deals with the use of electromyographic (EMG) analyses in applied physiology and is divided into three sections: 1) Surface EMG analyses, 2) Intramuscular EMG analyses, and 3) Evoked potential analyses, respectively and will cover the following topics:

1) SURFACE EMG ANALYSES
- Determination of Maximal Power Output at Neuromuscular Fatigue Threshold
- EMG Manifestations of Neuromuscular Adaptations during Acquisition of Motor Tasks

2) INTRAMUSCULAR EMG ANALYSES
- Motor Unit Activity during Fatigue
- Oxygen Availability and Motor Unit activity
- Motor Unit and Muscle Sound Properties

3) EVOKED POTENTIAL ANALYSES
- Myotatic Reflex Development
- Modulation of H-reflex during Various Motor Tasks
- Catch-like Properties of Muscle: Electrophysiologic and Metabolic Responses

A Multicriteria Procedure to Support Surface EMG Acquisition

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INTRODUCTION

The assessment of the electrical manifestations of muscle fatigue has been applied to patients affected by different forms of muscular disorders, to obtain an objective evaluation of muscle impairment. The evaluation protocol we developed and utilized on patients affected by different forms of muscular dystrophies consists of three phases: 1) the determination of the maximal voluntary contraction force (MVC); 2) the execution of one or more sustained voluntary contractions (attempted constant force and isometric); and 3) one or more sustained electrically elicited contractions (constant stimulation parameters, isometric). The surface myoelectric signals is generally detected by means of array electrodes, amplified, and processed to obtain the time course of the power spectrum mean frequency (MNF), muscle fiber conduction velocity (CV), and the root mean square value of the signal amplitude (RMS). The time courses of these parameters, normalized by the corresponding initial values, are jointly plotted in a diagram referred to as “the fatigue plot”. We believe that this diagram summarizes the minimal information that is needed to evaluate the progression of muscle fatigue during the sustained contractions. A critical aspect of the execution of the evaluation protocol is the correct positioning of the detection probe on the investigated muscle. In fact, it was demonstrated that small variations of its position may affect substantially the results of the test. Hence, the electrode position affects greatly the inter-experiment repeatability, particularly if different tests on a specific subject are carried out by different operators. The manual search of a suitable electrode position requires considerable expertise, that may be obtained only through a specific training and extensive practice. It would then be desirable to develop a procedure that allows to locate the detection probe properly, without requiring extensive training of the operator, and assuring a high inter-experiment repeatability. The solution we propose to solve this problem uses a specific outranking method that belongs to the field of multicriteria decision aid. Multicriteria analysis is a research field which provides methods to support the decision aiding process when several, often contradictory, points of view must be taken into account. There is a large variety of multicriteria methods. Among them, the outranking methods model the decision-maker preferences by means of a relation, referred to as “outranking relation”, between each couple of possible solutions. These relations are then used to achieve a ranking of the alternatives or to select the best one. The procedure we developed relies on Electre III, that is a specific outranking method. The aim of the presentation is to describe this procedure and present experimental results relative to its validation.

THE PROCEDURE

The main elements of an outranking method are I) the set of actions, denoted by ***A*****,** that is the set of possible solutions to be explored during the decision aiding process, and II) a coherent family F of criteria which represents all the different aspects of the problem to be solved and that allow for the comparison of the different actions. In the “positioning” procedure, the actions are thirteen different positions of the electrode. Each position is characterized by different values of MNF, CV, RMS, and of the correlation coefficient between the two double differential signals (CC). To speed up the procedure and to ensure the homogeneity of the data, we collected the signals simultaneously from the different positions by means of a sixteen-bar array electrode. The criteria used to compare the actions are derived from MNF, CV, RMS and CC. To evaluate and compare the different couples of actions, our procedure utilizes four criteria and twelve rules. Each criterion may comprise specific preference, indifference, and interdiction thresholds. Although the ranking of the different actions is not straightforward, it follows the general guidelines on the positioning of the detection probe that have been previously published. Oversimplifying this issue, it may be assumed that the optimal position is that in which CV and MNF have the lowest value while RMS and CC reach the highest one. Despite the simplification, it is evident that in general these conditions on the specific parameters are met in different positions along the muscle, and a single point in which all the conditions are satisfied does not exist. The "positioning" procedure ranks all the different actions and suggests the alternative that best satisfies the user requirements described by rules and criteria. Specifically, the sixteen-bar array electrode is placed on the skin over the muscle with its major axis parallel to the fibers. The last three or four bars of the probe are placed over the tendon. The subject is asked to perform an isometric and constant force contraction lasting 5s. The signals are collected, subdivided in epochs lasting 1s, and processed. The values of MNF, CV, RMS, and CC computed over the third epoch provide the input to the multicriteria procedure, while those relative to the other epochs are used to validate the results. The procedure ranks the alternative positions from the best to the worst and gives a measure of reliability of the obtained ranking. When the reliability is too low, the user is directed to move the electrode proximally (20mm) and to repeat the contraction.

RESULTS AND DISCUSSION

First, the variability of the results obtained through the application of the "positioning" procedure was quantified by comparing the values of MNF and CV computed over twenty repetitions of an 80%MVC voluntary contraction lasting 5s. The contractions were performed by a specific subject without changing the position of the probe during the entire experimental session. By analyzing the distribution of the values of MNF and CV computed during the third second of contraction on the best alternative suggested by the "positioning" procedure we observed that MNF ranged from 89 Hz to 106 Hz (97 Hz $\pm$ 5.1 Hz, mean $\pm$ st. dev.) while CV was comprised between 3.4 m/s and 3.9 m/s (3.6 m/s $\pm$ 0.14 m/s, mean $\pm$ st. dev.). In this case, since all the possible causes of variation of the studied parameters were avoided, the variability observed is explained by the following observations: a) in different contractions the force exerted by the subject slightly changes and b) the tibialis anterior is not the only muscle acting during foot plantar flexion and then its recruitment level may differ even if the ankle torque is kept substantially constant. Extremely important in clinical applications is the possibility of obtaining results that do not depend on the specific operator and his/her training. To evaluate the performances of the "positioning" procedure with different users, 20 different operators received a very short training (15 minutes) and then were asked to locate the probe on the same subject previously studied. Twelve operators obtained the correct location with one trial, seven with two trials, and only one operator with three trials. The analysis of the values of MNF and CV showed that MNF ranged from 93 Hz to 105 Hz (100 Hz $\pm$ 4.6 Hz, mean $\pm$ st. dev.) while CV was comprised between 3.1 m/s and 3.9 m/s (3.5 m/s $\pm$ 0.25 m/s, mean $\pm$ st. dev). The statistical analysis of the results obtained by a single and well trained operator and those obtained by different, almost untrained operators, showed that: a) the variances of the two series of MNF values are not statistically different (F-test, $p > 0.67$); b) the variances of the two series of CV values are statistically different (F-test, $p < 0.01$); c) the mean values are not statistically different in both cases (paired t-test, $p > 0.07$ for MNF and $p > 0.15$ for CV).

CONCLUSIONS

These result are extremely encouraging, since they demonstrate that the "positioning" procedure enables almost untrained operators to obtain results that are equivalent to those obtained by a trained expert, thus facilitating the diffusion of the study of muscle fatigue in the clinical environment.

Relation Between Stretch-Evoked Force, Activation Level and Amount of Stretch in Human Muscle

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PURPOSE: The purpose of this study is to investigate the relation between stretch-evoked force, tonic isometric force, and amount of stretch in human FPL muscle.

ESTIMATION OF STRETCH-EVOKED FORCE: Figure 1 shows a force generation system when displacement X is applied to muscle at voluntary isometric contraction. The total force F is represented as addition of forces by passive component (F_p), by viscoelastic component (F_v) and by force generation component (F_c).

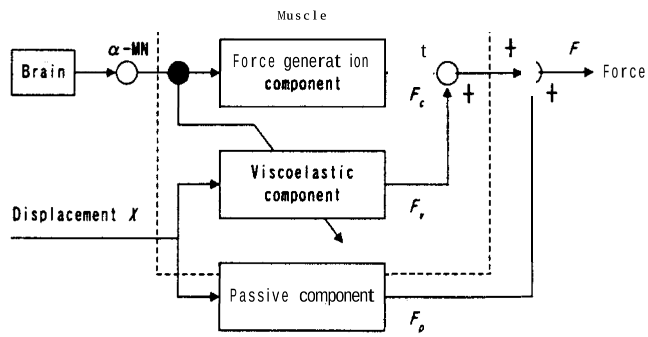

Fig.1 Force generation system of muscle

$$F = F_p + F_v + F_c \qquad (1)$$

In this study, F is a measured force, and F_c is a force of the force generation component in a voluntary contraction. F_p is a force response at resting state of the FPL muscle. If the same displacement X is given at the resting and contracting states, the viscoelastic force (stretch-evoked force) F_v can be estimated by following equation.

$$F_v = F - F_c - F_p \qquad (2)$$

EXPERIMENT SYSTEM: An examined muscle was a Flexor Pollicis Longus (FPL) muscle of left hand. An experiment system is shown in Figure 2. The left thumb was fixed to an aluminum plate (50mm) which is connected to an output shaft of a voice-coil type linear motor. The thumb was stretched or shortened according to back and forth of the output shaft of the motor. The thumb position was controlled with the motor servo system. Strain gauges were put on the aluminum plate; the detected signal from the gauges was amplified and averaged by a strain amplifier. The output signal was defined as tension of thumb, and was monitored on a CRT of oscilloscope. Electromyograms(EMGs) were measured as follows. Two pair of Ag-AgCl surface electrodes were placed longitudinally on the skin above the flexor and extensor muscles, respectively.

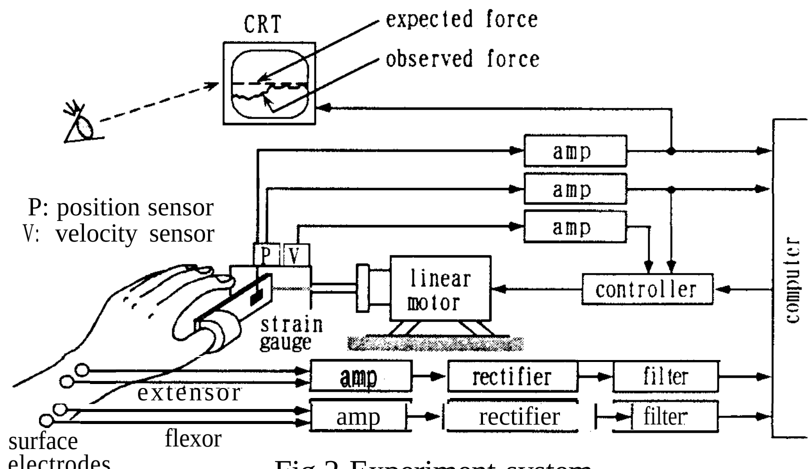

Fig.2 Experiment system

EMGs were passed through a differential amplifier, a full rectifier, and finally a second-order low-pass filter (IEMGs). The displacement, the velocity and the acceleration of length perturbation, the tension of thumb, the IEMGs were A/D converted every 0.5ms, then were input into a computer.

PROCEDURE: The desired isometric force and the strain-gauge output (tension of thumb) were displayed on the CRT. Subjects were asked to generate the tension equal to the desired force (0,5,15,20,30,45% of MVC; 0% means the resting state) with watching the bright line on the CRT. This constant force is denoted by tonic isometric force. Then, ramp perturbations of 7mm stroke and 25ms duration were applied to the thumb by the motor. Measurement was repeated ten times in each condition. Observed data from the onset of perturbation to 35ms were investigated, since the latency of stretch reflex was longer.

RESULTS: The following results were obtained;

1.The stretch-evoked force increased with increasing tonic isometric force; (a) FPL was stretched, (b) FPL was shortened, and t was time after onset in Fig.3.

2.The stretch-evoked force increased linearly with increasing the amount of length perturbation; (a) FPL was stretched, (b) FPL was shortened in Fig.4.

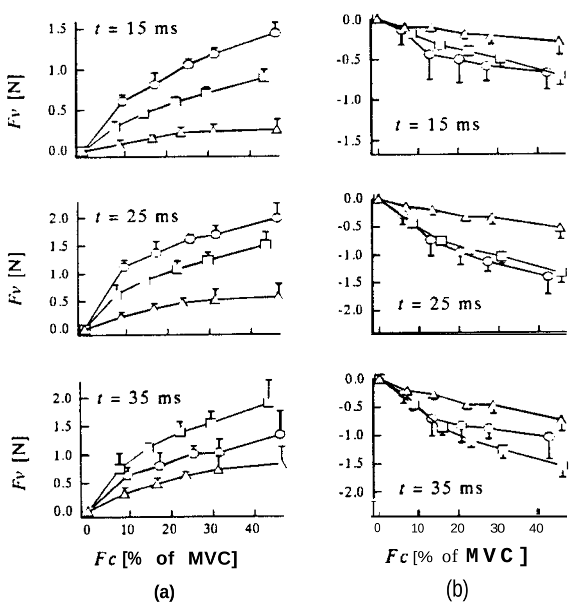

Fig.3 Relation between F_c and F_v

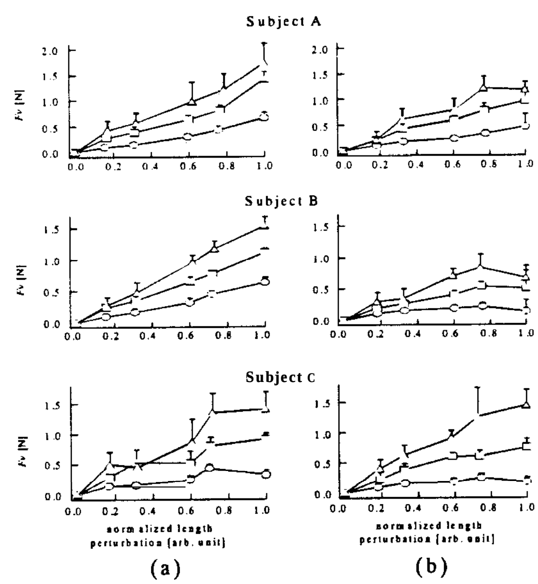

Fig.4 Relation between the normalized length perturbation and F_v

Non-Invasive Identification of Motor Units With Linear Electrode Arrays

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INTRODUCTION

The applications of linear electrode arrays in surface EMG (SEMG) have been described in the last 15 years [1, 2, 3] and have been mostly focused on the estimation of muscle fiber conduction velocity and of the location of the neuromuscular junctions (NMJ). The technique provides multiple observation points along the entire muscle; in this condition the propagation effect is much more evident than in the case of two pairs of detection electrodes and it is possible to recognize and classify the single motor unit action potentials (MUAPs) using information from all channels, such as the position of the innervation zone(s), the length of the fibers, and the conduction velocity. This information is not completely available if only a portion of muscle is investigated.

It is purpose of this work to show the possibilities that linear electrode arrays offer for the study of physiological properties of the muscle, such as the recruitment order, the type and number of motor units (Mus) activeted during the contraction. In particular we will address the issue of the identification of the same MUAP at different contraction levels, in order to investigate the recruitment order.

MATERIALS AND METHODS

Four tests were performed on the biceps brachii of six healthy male adults with no history of neurological diseases. Ages ranged from 24 to 36, with an average of 29.7 years. All subjects were tested supine with the shoulder abducted to 90 degrees and forearm in vertical position placed in an isometric brace. A linear array of sixteen electrodes, with interelectrode distance of 10 mm, was placed over the biceps brachii of the non dominant arm of each subject, aligned with the direction of the fibers. The subject was asked to perform four contractions in isometric conditions at l0%, 30%, 50% and 70% of the maximal voluntary contraction (MVC) sustained for 30 seconds. The signals have been analysed with the purpose of testing the possibility of identification of the same MUAP at different contraction levels.

RESULTS

A 350 ms long record of differentially detected biceps activity at 50% MVC is depicted in Fig.1. Different MUs may be identified from this record and their innervation zones can be detected. It is evident that the entire pattern from tendon to tendon is required to identify each MU and a few channels may not be sufficient for this purpose. Of course the identification of a MU pattern is relatively easy when there is no overlapping between MUAPs and much more complex in case of superposition *or* interferential patterns. Results showed that it is possible to identify single MUAPs even at high levels of contraction (50% and 70% MVC). It was found that some MUs recruited after a few seconds at 30% MVC can also be identified from the beginning of higher level contractions (50% and 70% MVC) when the MU pool is larger, as outlined in Fig.2 (left). At 30% MVC it is possible to observe that MUs that contribute small MUAPs are recruited at the beginning, while larger MUAPs are involved at the end of the contraction; the latter are those found also from the beginning of contraction at higher force levels. Another example is shown in Fig. 2B, where a MUAP is recognized at 30% MVC (at the end of the contraction, with low firing rate) and at 50% MVC (from the beginning, with higher firing rate). In this case, because of the movement of the skin and the slight shortening of the muscle, the MU action potential slightly changes at 50% MVC with respect to 30% MVC, especially near the tendon region.

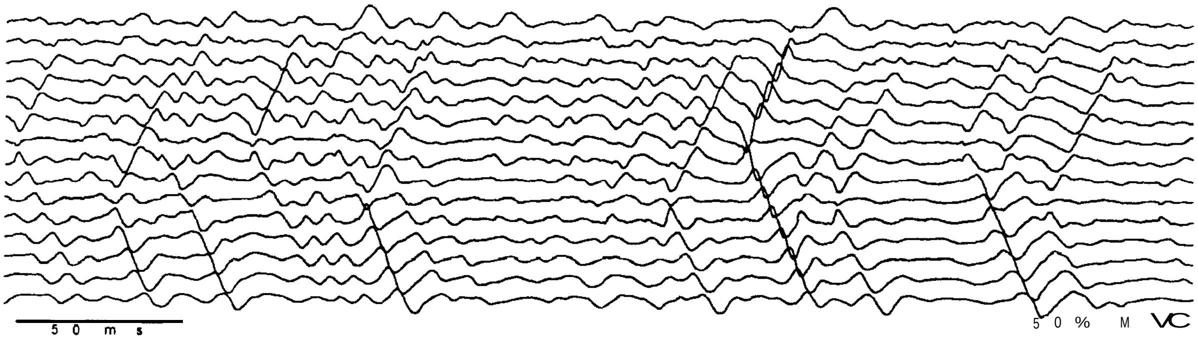
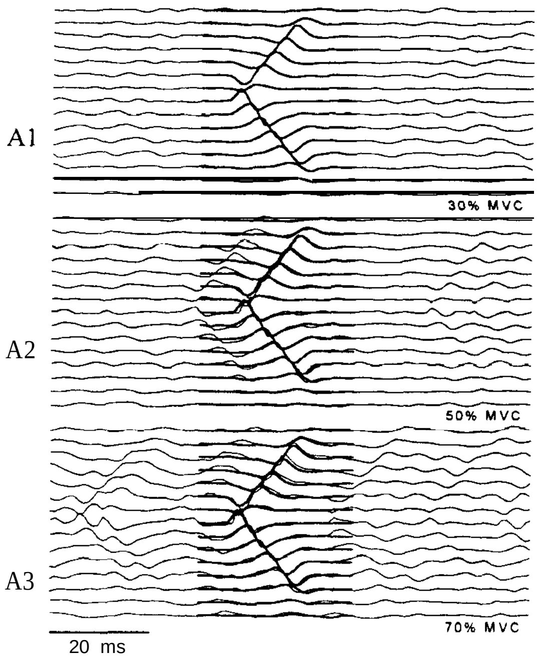


Fig. 1 Record of 350 ms of differentially detected biceps activity at 50% MVC.



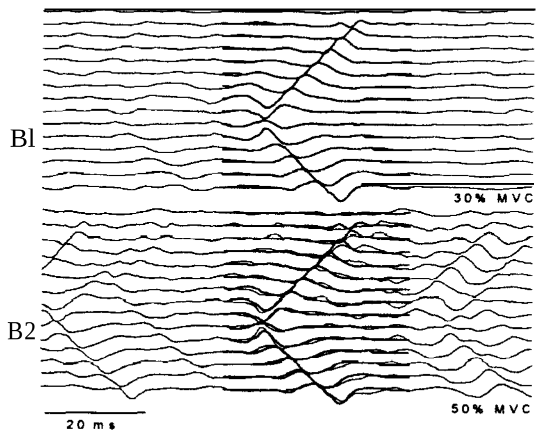


Fig. 2 A: dentification of MU "signature" at 30% MVC and recognition of the same "signature" at contraction levels of 50% and 70% in a healthy biceps brachii; B: identification of MU "signature" at 30% and 50% MVC.

DISCUSSION

The use of linear electrode arrays is very promising for the study of physiological properties of the muscles, such as the recruitment order, the type and number of motor units activated during the contractions. Furthermore, the technique provides a starting point to address the issue of a) SEMG decomposition into the constituent trains of MUAPs and b) estimation of muscle fiber conduction velocity distribution.

ACKNOWLEDGMENT

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Optimal Myoelectric Feature Space for Pattern Discrimination

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INTRODUCTION

Many children have upper extremity amputations as a result of congenital deficiencies or trauma. Powered upper extremity prostheses (PUEPs) have become one of the most commonly used assistive devices for these cases. Currently available PUEPs provide control over hand open/close functions, but the need is to develop a device and control system capable of a greater number of functions.

Since the 1970's, researchers have attempted to find a method that uses the EMG signal to differentiate between distinct muscle contraction patterns. This would provide the ability to control a greater number of prosthetic functions[1]. Results have shown that there may be a deterministic component of the EMG signal which could be used for this purpose. A practical solution has yet to be found.

For this work, fundamental analyses of the EMG signal were conducted in order to determine the optimal representation of the EMG signal for this purpose. The EMG data were sampled under conditions intended to simulate those of intended end-use.

EXPERIMENTAL DESIGN

The failure of most of the research to date is due in part to the neglect of at least one of the following aspects:

- Data collection does not reflect conditions of use
- There is no quantitative comparison of the choices of feature space
- There is no quantitative comparison of the choices of classifier

The first two issues are the most fundamental elements in the problem, thus the focus of the experimental design was based upon them. Practically these issues involve the selection of subjects, and the conditions of product use.

The selection of subjects involved three First, people with upper limb amputations, instead of able-bodied individuals, were used as subjects. This is because it is much more difficult for amputees to reproduce a particular muscle contraction pattern since they perceive little or no visible feedback associated with a given action. Second, there may be large physiological differences between able-bodied individuals and those with congenital deficiencies. Lastly, below-elbow amputees were chosen as the subject population as the most likely end-users.

The conditions of prosthetic use can cause a great deal of variation in the EMG signal. Normal use involves conditions of loading, motion of the residual limb, and fatigue. Therefore, these conditions were simulated in the experimental protocol so that their respective effects on the EMG signal could be observed and accounted for.

EXPERIMENTAL PROTOCOL

The volunteer subject was female, sixteen years of age, with a below-elbow congenital birth deficiency. The subject was a current user of a PUEP. In order to maintain clinical simplicity, the two electrode sites chosen for normal prosthetic use were also used as the experimental sites.

Raw EMG data were sampled from the two sites while the subject generated what she felt were six discrete muscle contraction patterns. The EMG data were sampled ten times for each contraction pattern under a control condition (subject relaxed), as well as ten times under conditions of loading (weight belt on residual limb), and ten times during motion of the residual limb (subject raising arm in front). The trials were conducted in random order.

EMG data were measured using Neuromedical Supplies, 4mm diameter snap electrodes. The signals were then amplified through a Bortec unit and sampled at 1000 Hz. using a DAS- 1600 A/D board. An example of the raw data taken while the subject was contracting the extensor muscle can be seen in Figure 1.

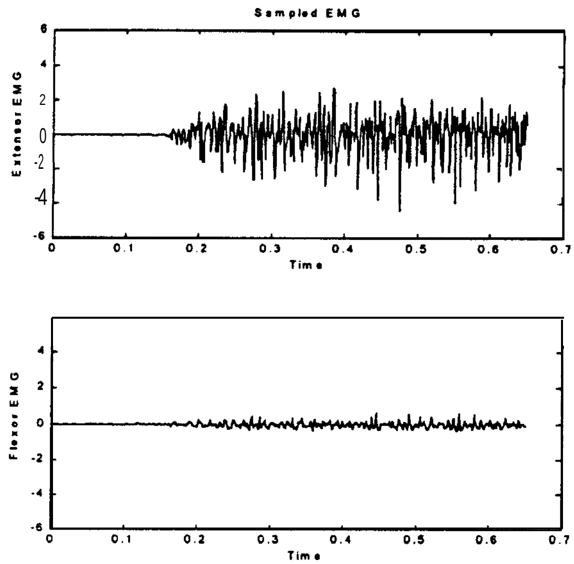

Figure 1: EMG data for subject during extension

DATA ANALYSIS

Representations of the EMG signal are denoted feature spaces. Possible selections of feature space are nearly infinite in number, each selection being composed of a parametric representation of the signal calculated over a specific time window. In order to reduce the magnitude of this problem, a selection of parameters as well as time windows were chosen logically. Since it is likely that all parameters of a particular domain (e.g. amplitude, frequency) would yield similar information, one parameter from each domain was chosen on the basis of minimal use of computational power. In the example provided here, these parameters were the mean and zero-crossings. The time windows required a minimum length such that adequate spectral information could be calculated, and a maximum length such that response time would be acceptable in any final product. On this basis windows of 40, 50 and 60 ms were selected. The data were analyzed over the EMG transient on the basis of the work by Hudgins et al [2].

Using the SPSS statistical software package, quick clustering analysis was then done for each feature space. A comparison of the results was conducted in order to determine the optimal choice.

RESULTS

Initial results have shown that the mean is a superior representation of the EMG signal than the zero-crossings for discrimination between muscle contraction patterns. In addition, for both parameters, the results of the clustering analysis improved with increased length of time window. The maximum clustering accuracy was found for the mean using a 60 ms time window, which yielded a maximum classification accuracy of 80% for some of the muscle contraction patterns.

DISCUSSION

Overall accuracy of the chosen method was much lower than anticipated. It is believed that this is a result of both the use of an amputee as a subject and the noise introduced by measuring under the various conditions. However, this is a more realistic estimation of the problem than has previously been seen.

Since the work described here is to compare feature spaces, a linear classifier was used. This represents the worst-case result. In practice, a non-linear classifier could be used.

Further statistical investigation will be undertaken using a wider variety of parameters. In addition, it is believed that by using a combination of spatial and temporal parameters, and clustering in N-space, a much better result will be achieved.

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Reliability of Biomechanical Measure of Postural Control: CP-CM

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INTRODUCTION

Postural control is a fundamental consideration when evaluating patients with neurological and neuromuscular deficits. The characteristics of the movement of the center of pressure (CP) have been used to infer the neurological and biomechanical mechanisms of postural control. The CP acts as a controller of the center of mass (CM), the controlled variable. In this context, the combined interpretation of the CP and CM displacement measures provide better insight into the assessment of balance. A new variable has been proposed (CP-CM) which represents the distance between the CP and CM. The major argument for the use of the CP-CM variable is that it is highly correlated with the acceleration of the CM during quiet standing [1].

This biomechanical variable (CP-CM) is used to evaluate standing balance in clinical settings. Consequently, more information is needed about the reliability and validity of this measure. The intra-variability of the measure as a major source of variance (unreliability). Acceptable reliability for the test is between .94 and .85 [2]. When one measurement is not reliable enough, the reliability can be improved by averaging many successive measurements. In this context, the purposes of this study were twofold: 1) to evaluate the reliability of one measure of CP-CM and 2) to estimate how many trials are necessary to obtain an accurate measure of postural stability using the biomechanical variable CP-CM.

METHODS

Subjects

Seven subjects (4 women, 3 men) participated in this study. The average age of the subject was 68 with a standard deviation of 4.3 years. Subjects were excluded if they had a history of Parkinson's, stroke or musculoskeletal disease.

Protocol

Subjects stood quietly on two adjacent force platforms for a period of 120 seconds with their feet side by side at pelvis width. They were instructed to look straight ahead with their head erect and maintain balance. Their arms were placed in a comfortable position: hanging at their sides. To ensure that this position remained constant, tracings were taken of foot placement and subjects were required to remain within these tracings in all the trials. Twenty-one infra-red light emitting diodes were attached bilaterally to anatomical landmarks in order to define a 14 segment model. Three OPTOTRAK sensors recorded marker displacement at a sampling frequency of 20 Hz while ground reaction forces and moments were recorded by two AMTI force platforms at a sampling frequency of 20 Hz. Measurements were made in a double leg stance, eyes open condition. Nine successive trials with rests between were done to estimate the reliability of the CP-CM.

Data analysis

The root mean square (RMS) amplitudes were calculated for the CP-CM variable in both antero-posterior (A/P) and medio-lateral (M/L) directions. The intra-variability of one measure of CP-CM was estimated using intra-class correlation coefficients (ICC) that compare within-subject variability (factor repetition) with between-subject variability3. For each ICC, the 95% confidence interval was calculated to take sampling variation into account. According to Dormer & Eliasziw (1987) a sample size of 7 subjects with 9 repetitions allows estimation of ICCs over .95 with a type I error of .05(4). The Spear-man-Brown formula was used to estimate the reliability of the mean of different numbers of trials (k ratings)‘.

Spearman-Brown formula

$$ICC_k = \frac{n\ (MS_B - MS_E)}{n\ MS_B + MS_R - MS_E}$$

where: MS_B = mean square between
MS_R = mean square repetition
MS_E = mean square error

RESULTS

The ICC obtained for one measure of the CP-CM was .75 in the A/P direction and .69 in M/L. Figures 1 and 2 illustrate each ICC obtained by averaging k trials together with a 95% confidence interval. From this graph, it can be concluded that with four trials, the variable CP-CM is reliable at .93 in A/P direction and .90 in M/L.

Figure 1: ICC (A/P)

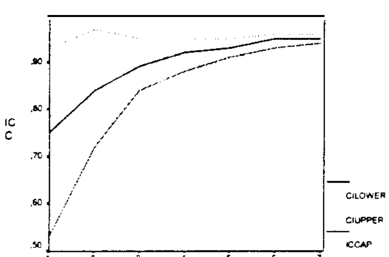
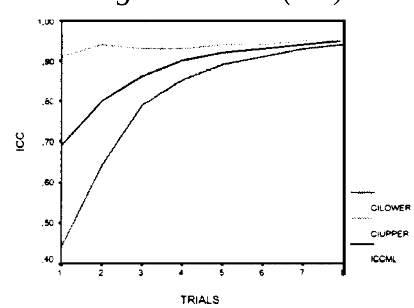


Figure 2: ICC (M/l)



DISCUSSION

The data presented here show that one measure of the CP-CM cannot be considered a reliable measure of postural stability. At least, four trials must be done in order to have a reliability coefficient of .90. The results of this study show that even when the instrument used for the analysis of postural control is very precise, it does not mean that the measurement is reliable. Postural control laboratories are now showing great potential in assisting the understanding of postural disorders in many clinical areas. Nevertheless, in order to fulfill this potential, it is important that the clinician is aware of the errors of the measure that may be present and is able to correct them adequately. Furthermore, it is crucial to ask for any measure of postural control whether or not the results from a single measure are representative of a subject’s balance performance.

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Acknowledgments

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Practice-related Changes in Lumbar Loads During Rapid Voluntary Pulls Made While Standing

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INTRODUCTION

Recent research has suggested that mechanical load and prediction play roles in controlling the lumbar spine during simple manual material handling (MMH) tasks where subjects' body movements are relatively constrained. [1,2] Whether similar effects occur for complex MMH tasks performed by unrestrained, freely-standing subjects is less clear. Neither is much known about how practice alters joint loading during complex MMH tasks performed while standing. The purpose of this study was to describe how external lumbar loads change when freely-standing subjects practice making rapid voluntary pulls on a handle over five days. We hypothesized that lumbar loads would be lower and their correlation with pulling force would be greater after practice. These hypotheses are based on the assumptions that lower lumbar loads reduce mechanical stress to the lumbar spine, and that increased predictability enables subjects to activate trunk muscles optimally to counteract the external load.

METHODS

Ten adults (age = 27.7 $\pm$ 9.3 yr., without a history of low back pain) performed abrupt horizontal pulls on a handle, to targets equaling 20, 40 and 80% of their estimated maximal dynamic pulling force, for five days (36 trials per target, per day). The pulls were brief (mean force rise time = 210 ms). Subjects received feedback about peak pulling force, using a faded feedback schedule (100% feedback on day 1, to no feedback on day 5). They received no feedback about body motion. Subjects had to keep the feet flat on the floor. pull straight back, and make the pulls brief. Otherwise, movements were unrestrained. Ground reaction forces and torques, horizontal pulling force and kinematics were recorded on days 1 and 5. Inverse dynamics were used to compute lumbar, hip, knee and ankle torques (loads) for each trial with a 4-link, sagittal plane model. Joint loads were measured at the time of peak pulling force, at the peak of each torque record, and as torque cost functions computed during the pull. Loads were normalized by body weight and height. The predictability of joint loads was operationally defined as the squared correlation coefficient (coefficient of determination, r^2) between pulling force and joint load at the time of the peak pull. Each subject's data were binned by pulling force before computing descriptive and inferential statistics. Results are reported for joint loads at the time of the peak pulling; similar effects were found for the other measures and their correlation with pulling force.

RESULTS

All joint loads increased significantly with pulling force. A repeated measures ANOVA showed significant practice-related changes in lumbar loads at the time of peak pulling force ($p < .05$; Fig. la). Loads decreased for the 20% pulls, did not change for the 40% pulls, and increased for the 80% pulls between days 1 and 5. Two subjects showed significant practice-related reductions in lumbar load for all three pulling forces. These data suggest that lumbar loads would decrease with practice, but also reveal considerable inter-subject variations in how practice influenced lumbar loads.

Coefficients of determination between pulling force and lumbar and hip loads were significantly higher after practice ($r^2 \geq 0.81$ by day 5; Fig. lb). Hip and lumbar loads always were very highly correlated, $r^2 > .98$. Correlations between ankle load and pulling force also increased with practice, less consistently and to a lesser degree than for lumbar loads. Coefficients of determination between knee loads and pulling force did not change with practice. Thus, subjects showed a relatively selective increase in the predictability of lumbar load from pulling force. The two subjects whose lumbar loads were lower after practice for all three force levels also achieved very high correlations between pulling force and lumbar load by the end of day 1 (open triangles and closed circles, Figure lb).

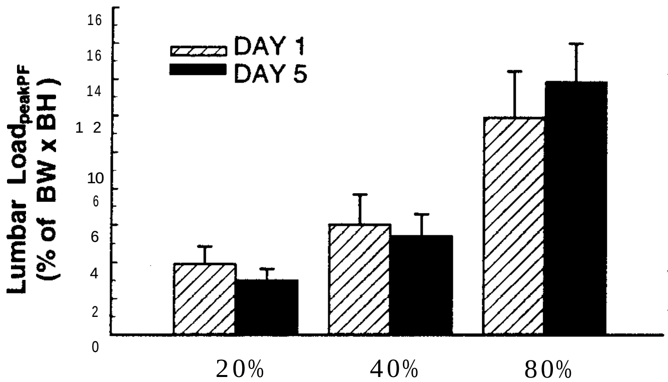

Figure 1 a. Mean $\pm$ 1 SD lumbar loads on days 1 and 5. Decreases in lumbar load for 20% pulls and increases for 80% pulls were significant ($p<.05$).

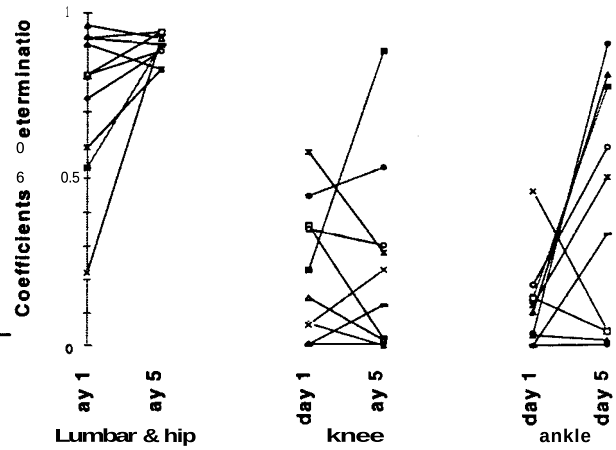

Figure lb. Coefficients of determination between joint load and pulling force, by day and joint. Each subject is represented by a different symbol.

CONCLUSIONS

The results suggest that subjects can learn to use two strategies to avoid low back injury during dynamic MMH tasks like standing pulls that involve complex, relatively unrestrained motion. One strategy is to reduce the lumbar load by changing kinematic and kinetic movement patterns. However, reducing lumbar loads may be difficult or impossible when subjects must generate large contact forces, as during pulls of near-maximal force. The second strategy involves increasing the predictability of the lumbar load. If subjects can predict the lumbar load from the intended pulling force, they could activate the trunk muscles with appropriate timing and amplitudes to balance the anticipated load. The higher correlations of lumbar load and pulling force on day 5 are consistent with this hypothesized predictive strategy. Future studies should determine what changes in movement patterns and EMG activity mediate practice-related changes in the magnitude and predictability of lumbar load.

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Efforts to Condition Vertical Loading Responses to a Dynamic, Angular Perturbation in Older Adults

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INTRODUCTION: The medical and surgical treatment of fractures resulting from falls represent a major health care concern among the geriatric population in the United States (Fried and Guralnik, 1997). Consequently, novel procedures to train older individuals to respond to dynamic perturbations in a timely and safe manner should be developed. Building upon our interest in feedback mechanisms to facilitate control of movement, the present studies were designed to determine whether feedback from transducers embedded in a NeuroCorn computerized balance device could be used to condition vertical loading forces among individuals older than 65 years of age. Data from age-matched control (no feedback) subjects were compared to experimental subjects whose loading responses were either down-trained (decreased) or up-trained (increased). Up-training experiments were prompted by observations from a clinical trial for older adults (Wolf et al, 1996) that indicated increased dispersion of vertical loading during toes-up perturbations with eyes open or eyes closed among Tai Chi subjects, compared to control or balance training groups (Wolf et al, 1997) Tai Chi participants also showed significant delays to first or multiple falls and significantly improved fear of falling.

METHODS: Thirty-four subjects (age range: 63-84 years) participated in these experiments. The first twelve (mean age 71.3 years, 5 men and 7 women) subjects had no fall events in the previous year and were assigned by pairs to the down-training or control groups. The second group of 11 non-faller subjects (mean age = 72.5 years, 3 men and 11 women) were assigned to up-training or control groups. A third group of 11 fallers (mean age = 73.0, **1** man and 10 women) were also randomly divided into two groups and followed the same protocol as the second group of non-fallers. Each subject participated in 12 sessions with 30 dynamic (toes-up) perturbations (8 degree arc at a velocity of approximately 66 degrees/sec). Perturbations were given over a random l-4 sec interval after the subject demonstrated less than 1 degree of sway. Sway information was given as a color on a monitor embedded in the surround at eye level. Absolute voltage data were sampled at 1 KHz and converted to force every 4 msec. Variations in loading forces over the 424 msecs preceding each perturbation were subtracted from each response. The response was divided into response and maintenance windows, The response window (RW) represented the first 225 msec following the perturbation and was subsequently divided into 75 msec bins, called Tl-T3, to represent the segmental, longer latency, and multi-segmental integrative postural responses, respectively (Diener, Horak and Nashner, 1988; Horak, Diener and Nashner, 1989). The maintenance window was defined as that stable loading achieved after the subject had responded to the perturbation. All control subjects had 12 sessions without feedback while all subjects undergoing conditioning received visual feedback in the form of a bar graph after each perturbation. The initial height of this bar was based upon the mean, total response window value for all 180 baseline perturbations. Thereafter a running median (across training sessions) was used to determine the magnitude of change. With successful responses, a line near the bar graph was lowered (down-training) or raised (up-training) by 1% as a new target level. Attempts to deliberately reverse the training direction to subsequently improve were not permitted. Additional functional measures, including the functional reach, the get-up-and-go test, and single limb support were also made before and after all experiments.

RESULTS: All subjects reduced (constrain) vertical loading for all baseline sessions. Little habituation was seen to these dynamic perturbations. In fact, the rate of change over sessions 1-6 for the entire RW and T3 interval was significantly different from zero for all subjects in the down-training experiment. The slope was not significantly different from zero across sessions 7-12. The control group in the down-training experiments showed no significant differences between the first and last six sessions, while a one way ANOVA comparing sway between baseline and training conditions within the down-training group showed significant reductions in the total RW (p=.O33), T3 (p=.O15) and RW (p.=.OO3). There were no significant between group differences. In up-training experiments, both groups showed significant reductions in loading responses for T3 during baseline (p=.O32). The up-training group tended to show a session effect for an increase in T2 (p<.08). The percent change in vertical loading response for both T3 and RW was close to showing a session effect for reduced magnitudes in the up-training group (p=. 109 and .06, respectively). Between group comparisons for loading response relative to baseline mean at each of the last six sessions for T2, T3 and RW average values were significant for the interaction effect of session and group (T2: p=.O47; T3, p=.O29; RW, p=,O21). Differences between groups during the training interval approached significance upon post-hoc testing. Data from the up-training experiments in older faller subjects revealed similar characteristics to those of their non-faller counterparts. There were no significant pre-post experiment clinical measures for any subjects.

DISCUSSION: Collectively, results from these individual studies indicate that older adults do not habituate to dynamic perturbations over a few trials or even a few sessions. All subjects, regardless of the training mode, will constrain their loading behavior, suggesting that perturbations with known dynamic components but delivered randomly, will yield similar responses. Thus, attempts at shaping responses to dynamic toes-up perturbations were unsuccessful; particularly in the up-training mode. Subjects attempting to up-train by developing stepping strategies were frustrated because the strategy and the resulting feedback were incompatible. Older individuals cannot respond to a dynamic toes-up perturbation through increased vertical limb loading in a timely fashion. Other training conditions to facilitate balance using different perturbation characteristics should be studied.

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Postural Strategies During High Acceleration Perturbations

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INTRODUCTION

The purpose of this study was to investigate postural strategies during high acceleration perturbations of quiet stance. Three distinct postural strategies have previously been identified during relatively low acceleration perturbations: ankle, hip and change-of-support, or step strategy. Horak & Nashner (1986) documented the ankle and hip strategies and described the possibility of a blend between the two. McIlroy and Maki (1993) described the step strategy and also demonstrated that a step may be initiated and interrupted during a perturbation. In realistic slip situations the acceleration of the foot/feet may reach and even exceed 1 g with skid velocities in the early phase of the slip approaching 100 cm/s (Strandberg 1983). Postural strategies have not been consistently studied at such large perturbation levels. This study was designed to investigate if high acceleration perturbations trigger similar postural strategies as those previously reported for low level perturbations.

METHODS

Nine healthy, physically active males (age 22-40 yrs, height 1.75-1.95 m, weight 72-83 kg) were perturbed at the feet during quite stance by a balance platform (BALDER, BALance DisturbER) instrumented with an AMTI force plate. BALDER was used to impose forward and backward perturbations with an acceleration and deceleration of 1 g (981 cm/s^2) at three different magnitudes (5, 7 and llcm). This resulted in maximum velocities of 70, 80 and 100 cm/s respectively. Displacements of reflective markers placed bilaterally on anatomical landmarks at the ankle, knee, hip, shoulder, elbow and wrist joints, as well as on the forehead and chin were calculated from recorded video images using an Ariel APAS system. Subjects were standing quietly and relaxed looking straight ahead prior to each perturbation and were instructed to avoid taking a step if possible. There were no objects in the vicinity of the platform that could be used for additional external support.

RESULTS AND DISCUSSION

A new technique was used to identify postural strategies from kinematic data. A line between the subject's ankle and shoulder marker was used to model an inverted pendulum behavior of the subject. Distance changes of the knee and hip marker relative to this line were used to indicate configurational deviations from the inverted pendulum model. Plotting the hip marker displacement vs. the knee marker displacement proved to be a useful tool in identifying different strategies and blends thereof. In addition, bilateral differences in the distance from the each knee marker to the inverted pendulum line were found to be an indicator of step initiation. All three postural strategies mentioned in the introduction (ankle, hip and step) were executed by all subjects. However, figure 1 shows an example of a different response frequently seen during forward perturbations. The response can be divided into 4 phases. In phase 1 the distance to the inverted pendulum line decreases for both the knee and hip indicating knee extension and hip flexion (the body hinges at the hip). In phase 2 the hip continues to flex whereas the knee displays a rapid knee flexion associated with a downward deflection of the center of gravity (COG) and an unloading of the vertical force. When the knees are almost maximally pushed forward and flexed, phase 3 is initiated with a forward hip movement indicating the onset of a hip strategy. The subject finally regains equilibrium and the initial posture is restored by moving knee and hip back towards the original position. Horak and Nashner (1986) predicted that blends of the ankle and hip strategy could result in knee bending. However, the vertical unloading coinciding with a downward movement of the COG occurs earlier (-100 ms) and is larger (>lOO N) than the one seen during a 'clean' hip or ankle strategy. Thus, it may be argued that this represents a different strategy in which a quick collapse at the knee is followed by a sequential, distal-to-proximal recovery of body segments back over the base of support. This strategy would only function during forward perturbations due to the anatomical limitations of knee extension.

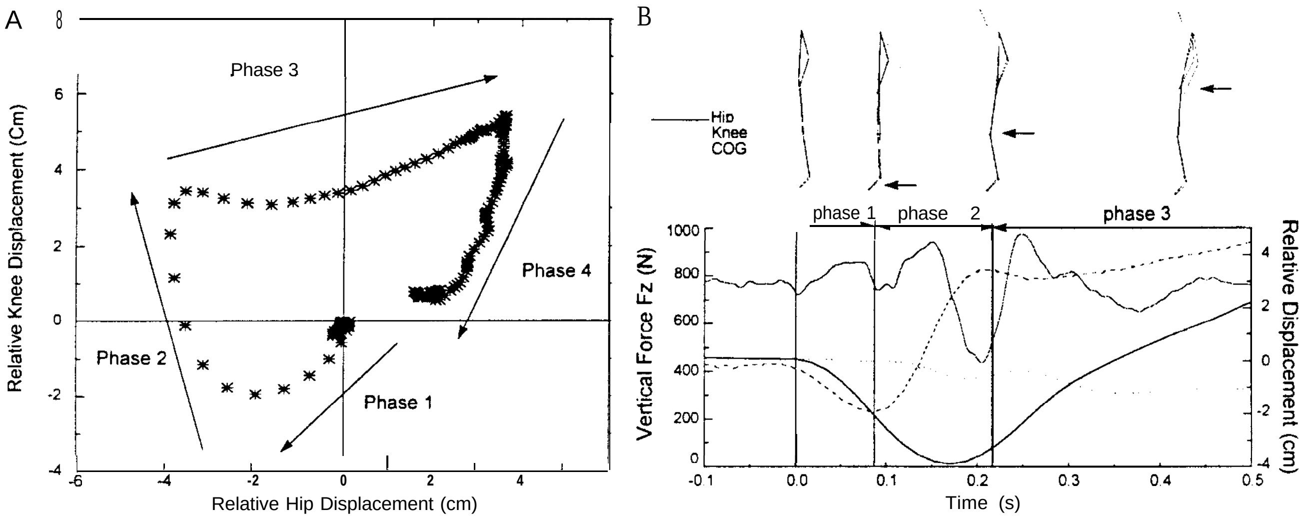

Figure 1. A) Hip and knee displacement relative to the inverted pendulum line (see text). Phase 1 coincides with platform acceleration. In Phase 2 the knee is pushed forward while the hip is still flexing. Phase 3 resembles the hip strategy followed by slow return to initial posture (Phase 4). B) Data from the same perturbation. On top, the stick figures of the subject at phase transitions during the movement. At the bottom, the vertical force measured by the platform, hip and knee displacement and the displacement of the center of gravity (COG).

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ACKNOWLEDGEMENT

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The Influence of Forward and Backward Pelvis Displacement During a Cycling Motion on the Muscular Work in Both Legs and Arms

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INTRODUCTION

Muscular work depends on the length-tension, force - velocity - power relationships of the involved muscles and on the effectiveness of force production, all affected by joint angles, muscles lengths and muscle moment arm lengths. These variables in turn are altered by changes in pedalling rate, in position and orientation of the body and by changes in seat to pedal distance e.g. seat height and crank length (de Groot et al. 1994). Too (1990) reviewed the literature on these issues and described systematically the effects of changes of these variables on bicycle performance e.g. cyclic movements.
In cyclic and partly in rhythmic movements also, we notice repeated periods of active lengthening (eccentric work) followed by active shortening (concentric work), which can affect efficiency as well as power. All muscles crossing the hip and the knee joint exhibit such negative work regions immediately followed by positive work regions. These regions are expressed in one or more octants of a full pedalling cycle. Simultaneous activity of muscles crossing opposite sites of a joint, has been viewed traditionally as paradoxical. Clarys et al. (1988), however denied the existence of this paradox with a simplified anatomical model.
Antonis et al., 1989 indicated a variation in efficiency with a saddle tube more forward as a function of an increasing slope during uphill cycling. However the cyclist's position with the saddle forward offers the best muscular conditions for uphill cycling.
These examples strongly suggest, that somehow body configuration, bicycle geometry, pedalling rate etc. can optimise length-tension and force-velocity-power interaction of multi-joint muscles. Clearly it is a challenge to investigate the relation between the effectiveness and the muscle efficiency.

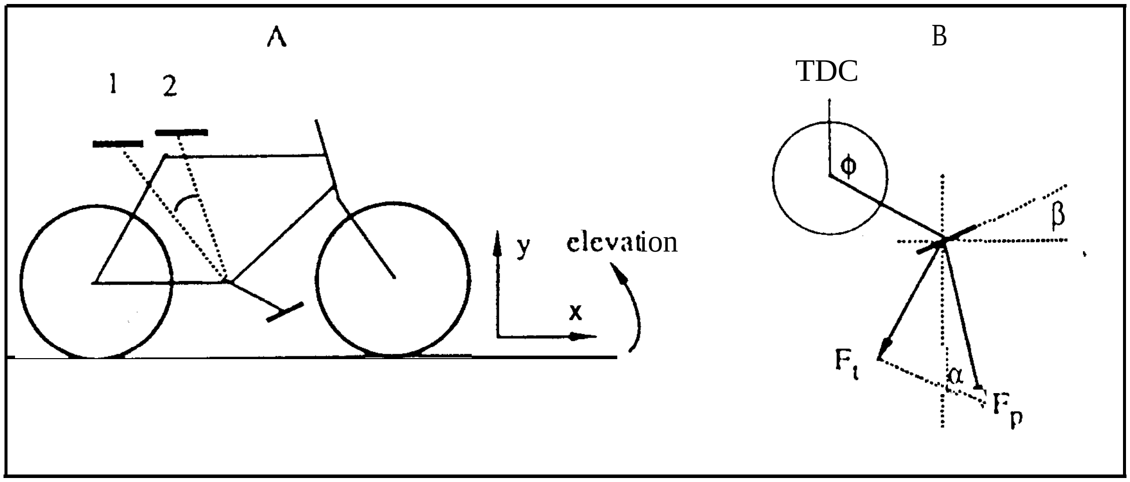
PURPOSE

In this particular study we are looking at the effect of forward and backward transient positions of the pelvis (and trunk) and in different (simulated) road slope percentages on the muscular activity of both the upper- and lower limbs.

METHODS

The EMG's are presented in circle diagrams (and octants) from the Top Dead Centre (TDC) to the Bottom Dead Centre (BDC), both conventional terminology and including both the down stroke and the up stroke dynamic lower limb participation, but also the isometric (static) activity of the arm muscles, synchronised with the cyclic (octant) periods of the leg.
EMG is measured on-line on a seven channel FM data recording system (TEAC) with pre-amplified active surface electrodes and an additional portable amplification unit.
A six-channel portable regulation unit provided the supply for the pre-amplified electrodes and also took care of the supplementary amplification. Signal strength was adjustable from 0 to 40 dB to match the input sensitivity of the portable seven channel data recorder (TEAC HR30). An external calibration device allows adjustment (O.l-1OV peak to peak) of all six channels to the individual EMG signal prior to the actual dynamic measurement. The synchronization of the pedalling cycle is generated electrically during each crank revolution at the (right) bottom death centre.
We have investigated the response of the pedal force direction during the down stroke (seated position) on a change in saddle-tube angle (13" as difference between 67" and 80") and on an imposed elevation angle (20") of the whole bicycle simulating uphill cycling.

An instrumented dynamometer with toe-clips was used. The pedal orientation (ß) and crank orientation (ϕ) were obtained from digital encoders with a resolution of 2 $\pi/150$. Nine elite cyclists pedalled at a cadence of 60 rpm and an imposed external power of 300 watt. In the figure below, the used variables are defined.



RESULTS AND DISCUSSION

Not only during the down stroke, but also during the upstroke positive power can be produced. This seems not to be a discriminatory factor for cycling performance herewith conferming Coyle et a1.,(1991). Higher performance is mainly attributed to generating higher 'down stroke power' by applying larger vertical forces . But this does not impede the importance of the observed orientation of the pedal force during the down stroke. Bicycle geometry can interfere with the pedal force orientation. In competition uphill cycling one can observe two strategies in seated position : climbing with a small and with a large effective saddle tube angle.
The results of this study suggest that no significant decrease or increase in EMG occurs in the lower limbs with significant displacement of the pelvis, while the EMG of the upper limb significantly decreases in all slope conditions with the pelvis (saddle) maximal forward (saddle-tube angle of 67”).

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A Measurement Procedure for the Quantitative Analysis of Free Upper-Extremity Movements

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Introduction:

There are various applications of upper-extremity motion analysis in orthopedics and neurology, but there is a lack of measurement procedures which provide the necessary quality. Usually, the measurement and analysis of upper-extremity movements is simplified by constraining the movement or by neglecting some D.O.F. Starting with the common assumption of a rigid-body model [l] a marker-based measurement of the movements is desired. For the analysis of joint motions usually Cardan angles are calculated from the marker trajectories. This requires the proper definition of joint axes which must be stable during a movement to provide an accurate analysis. Key problems are the relative motions between markers of one segment and relative motions between markers and bones [2].

Earlier approaches of upper-extremity movement analysis which used Cardan angles were restricted to special applications since they were invasive [3] or the setup constrained the movements [4]. The aim of the proposed measurement procedure is to yield a non-obstructive setup for a high quality assessment of all D.O.F. of free wrist and elbow movements.

Methods:

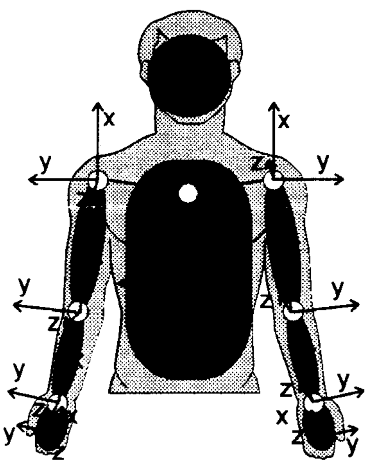

Figure I : Kinematic model and axes definitions

A rigid body model of the upper-body is the basis for measurement and analysis (Fig.1). Especially the shoulder kinematics is to be simplified in order to be able to measure all defined D.O.F.. The movements of each segment are determined from the positions of three non-collinear markers. Motions between each three markers which define one rigid segment are suppressed by connecting the markers. The 3D-trajectories of the markers are measured with a Vicon 370 motion analysis system at 50 Hz.

For the calculation of the Cardan angles, joint coordinate systems at wrist, elbow and shoulder must be established. To align them with anatomical axes additional markers lateral and medial to the flexion axes of wrist and elbow are used during a static trial. The center of the shoulder joint is estimated from a shoulder movement on the basis of the kinematic model. The Cardan angles are calculated by decomposing the total rotations between two joint coordinate systems into a sequence of three rotations around the three joint axes. In accordance with ISB standards proposals the first rotation is flexion/extension followed by abduction and rotation around the longitudinal axis.

The worst effect from skin movements can be expected at upper-arm and forearm, where the markers only inaccurately follow rotations around the longitudinal limb axis. Therefore, the definition of the elbow coordinate system is based on the joint centers of wrist, elbow and shoulder which is insensitive to skin-movements at the upper-arm. For the wrist coordinate system a correction is used which adds a fraction of pronation / supination to the measured value to compensate for a too small longitudinal rotation of the forearm markers. The amount of correction is deduced from a pure pronation/supination movement, where artificial wrist rotations around the longitudinal hand axis are suppressed.

Results:

Figure 2 shows the Cardan angles of shoulder, elbow and wrist during a tracking task. The subject followed with the index finger a marker which moved on an eight-shaped path in 10 seconds.

The curves show that a complex coordination of joint movements is necessary to carry out the movement. The absence of crosstalk at the elbow angles show the stable adjustment of the elbow coordinate system which is not affected by upper-arm skin movements.

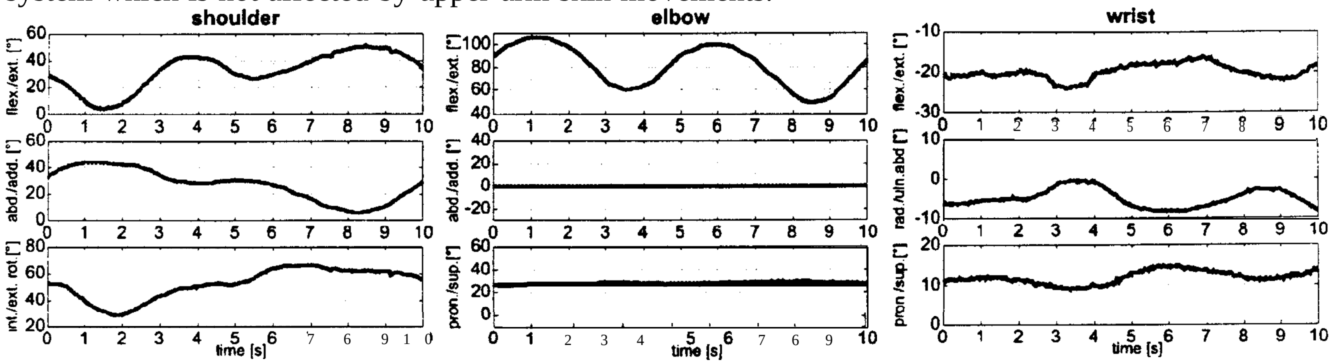

Figure 2: Joint angles of shoulder, elbow and wrist during a tracking task

Figure 3 shows on the left side the wrist angles during a pure pronation / supination of the forearm with straight hand. The rotation around the longitudinal hand axis is a skin-movement artifact. This artifact vanishes (right side) when the original forearm pronation / supination is increased by 35% .

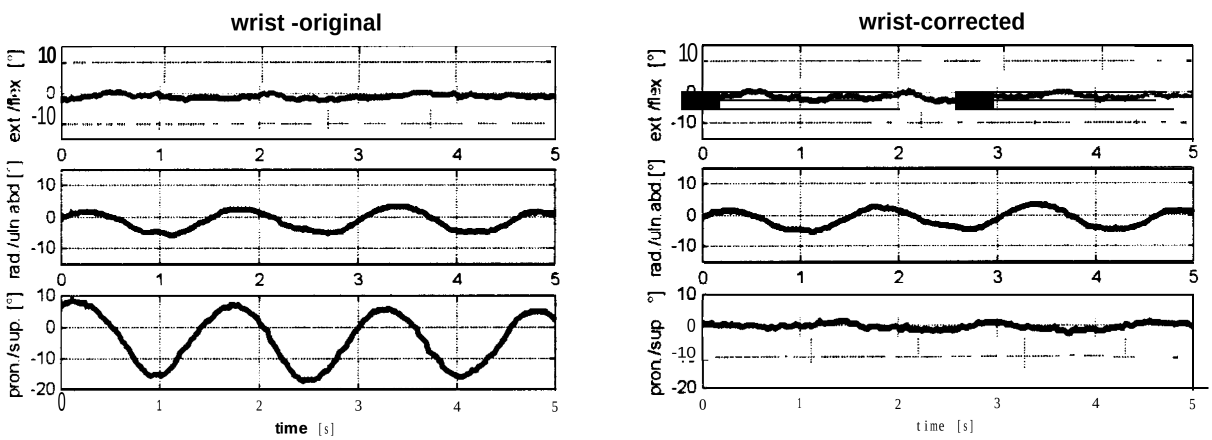

Figure 3: Skin-movement artiyacts at the wrist joint and their correction by increasing *forearm pro-/supination*

Discussion:

The measurement procedure allows to analyze all D.O.F. of wrist, elbow and shoulder during free upper-extremity movements. This yields very detailed information for a wide range of applications.

A reasonable accuracy can only be obtained when skin movements are regarded. At the upper-arm and the forearm they are effectively compensated by the coordinate system definitions and corrections. Without correction the pronation / supination would be largely underestimated. With this substantial improvements upper-extremity movement analysis is in a state to be evaluated in clinical applications.

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Development and Feasibility of a Protocol Using the Optotrak Probing System for Measuring 3D Displacements and Mobility of the Scapula

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INTRODUCTION: Abnormal movement of the scapula is often related to shoulder pathologies such as shoulder impingement syndrome(SIS)[1,2]. However, it is not clearly understood whether or not deficient scapula mobility is a predisposing factor for the development of SIS. Because of overlying skin movement, the use of conventional skin markers to measure scapula mobility during arm raising is impossible[3]. In this context, it is necessary to develop a reliable measure that will allow us to characterize three dimensionnal scapular displacements (3DSD) at joint angles in which subacromial impingement is more likely to happen such as pure shoulder flexion and abduction[4].The Optotrak probing system (OPS), never used for this type of study, offers such potential.The aim of this study was to verify the feasibility and reliability of a new test using the OPS to measure posterior-anterior transverse rotation (P-ATR), external-internal rotation (E-IR) and anterior-posterior tilting (A-PT) of the scapula. A new index to express these angular displacements and the total scapular mobility was also proposed.

METHODS: This methodological study was divided in 3 steps. **a) First, the feasibility of computing accurate 3DSD from data obtained with the OPS was assessed.** An anatomical model of the scapula was fixed on a rigid Plexiglass base that was mobile in all directions through a multiaxial joint system equipped with a locking mechanism. In order to verify the feasibility to estimate both direction and magnitude of 3DSD from OPS data, we first used this anatomical model to which, from a reference position, we imposed angular displacements in P-ATR, E-IR and A-PT of known direction and magnitude. The 3D position of the posterolateral tip of the acromion (Pl), the tip of the inferior angle (P2) and the spinal inferior edge of the spine (P3) of the scapula were digitized using a special probe of the Optotrak system. Using a local coordinate reference system (LCRS) with Pl as origin, a rotation matrix was obtained for the reference position. The same procedure was successively applied for each imposed displacement. Axial rotation angles of the scapula were then calculated using an Euler configuration (sequence-dependent rotations)[3,5,6].Positive rotation was determined using a right hand rule. Figure 1 illustrates LCRS and axis orientation for the scapula. Mean differences in degrees between calculated 3DSD (rotation matrix) and measured displacement (with fluid goniometer fixed on the Plexiglass) were determined along with their corresponding standard error (SE).**b) Second, the concurrent validity of 3DSD estimated with the rotation matrix (OPS data) was determined in comparison with the 3DSD estimated by the Optotrak software (conventional marker data).** Using 3 conventional markers stuck to the anatomical model at Pl, P2 and P3, we successively imposed angular displacements in each scapula direction. For each imposed displacement, we then successively recorded 3D position of these points first using the probe accessory and second the conventional markers. The concurrent validity was estimated by computing the mean of the differences between the two methods (OPS versus conventional markers) and its corresponding 95% confidence interval (95% CI). c) **Third, the reliability of 3DSD measures when using an OPS protocole was determined in a normal subject.** One normal male subject participated to 3 testing sessions over a one week period. During each session, 3DSD were measured for 7 different tasks. These tasks were to successively maintain his arm at rest, then raised (elbow in full extension, thumb towards the ceiling) at 70°, 90” and 110° of flexion (sagittal plane) and at 70°, 90” and

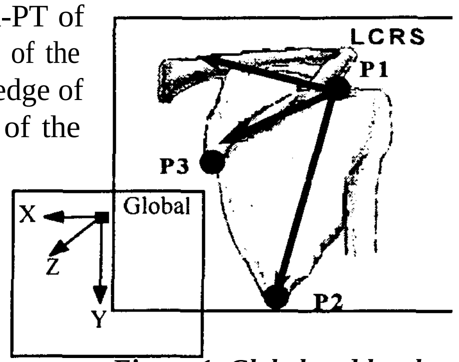

Figure 1. Global and local reference coordinate system (LCRS) and axis orientation for the scapula

110” of abduction (frontal plane). Two trials for each task were performed. The subject’s seated position, was standardized for all trials and sessions, the arm raising angle being insured by a fluid goniometer. For each trial, the following body landmarks were digitized: Pl, P2 and P3 as described above, lateral epicondyle, medial epicondyle, lateral arm marker, C7 spinous process, right and left postero-superior iliac spine. To increase the accuracy and the reliability in scapula landmark detection across tasks, we used a triangular template made of two flexible arms to establish, in the reference position, the exact distance between P1, P2 and P3. The apex of the template was located at P2 and each arm was then aligned with Pl and P3 respectively marking their distance from P2. This template was further used to exactly locate and relate Pl and P3 for all tasks. The reliability of 3DSD measures was estimated by comparing their values (average of 2 trials per task) between sessions. Coefficients of variation (CV) and Spearman rank-correlation coefficients (r,) were calculated. With this preliminary study, we also begun to considered a new mobility index called the scapular mobility index (SMI) that represents the total scapular mobility. The SMI is calculated for each task and is the sum of all scapular angular displacements (P-ATR + E-IR + A-PT) from the reference position (task position minus reference position).

RESULTS: a) Mean differences in degrees ± SE between 3DSD estimated from the rotation matrix and the imposed displacement were 0.78” ± .55 (P-ATR), 0.47” ± .54 (E-IR) and 0.48” ± .28 (A-PT). b) The comparison of angular displacements estimated from OPS and conventional markers shows a mean error between the two methods of 1.74 ° ± .7 within a 95% CI ranging from 0.35 to 3.13 and a r_s of .81. c) Results for repetitive trials for one normal subject showed a mean CV, between sessions, of 8.2 % for P-ATR, 9.1% for E-IR and 9.9% for A-PT. As shown in Figure 2, the total mobility of the scapula expressed by the SMI increases with an increase of shoulder abduction or flexion. The total mobility of the scapula in abduction is almost two times higher than in flexion. The relative contribution of each scapular movement to the SMI is much different according to the plane of shoulder movement. In abduction, P-ATR, E-IR and A-PT are contributing in an equal proportion to the SMI, but in flexion, the major contributor to SMI is the A-PT.

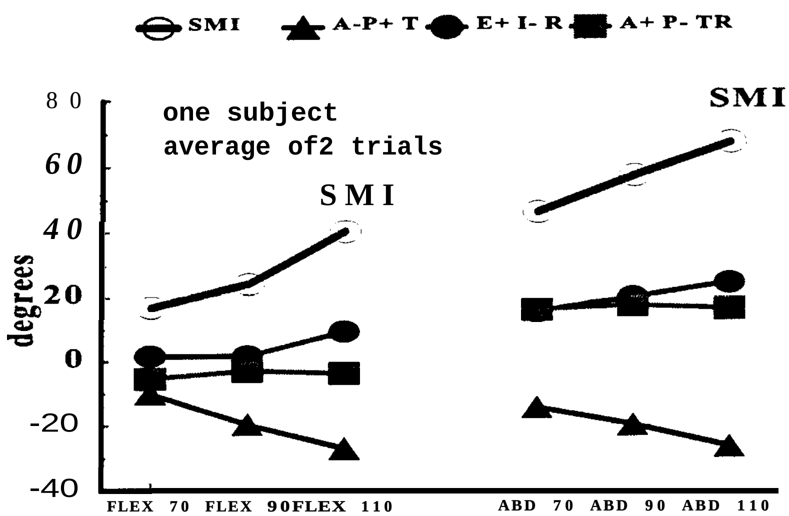

Figure 2, Relative contribution of each scapular movement to the scapular mobility index (SMI).

DISCUSSION: The measure of 3DSD using the OPS as described above is feasible, reliable and has a concurrent validity with the conventional method using markers. These results suggest that mobility of the scapula depends both upon the amplitude and plane of arm movement. The relative contribution of each scapular movement to the total mobility of the scapula may represent individual strategies. This new test using the OPS is promising and will be used on a larger sample of normal subjects and subjects with SIS.

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Evidence for Programmed Movements in Kinematic and EMG Activity Patterns

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Introduction: Muscle synergies resulting from motor programs can be defined as fixed patterns of muscle activation involved in generating a given movement. These patterns, inferred from electromyographic (EMG) signals recorded from muscles executing the movement, have relatively consistent spatial and temporal arrangements. Understanding synergies involved in generating similar movements provides valuable insight into the underlying spinal cord circuitry producing the muscle activation patterns.

The goal of this project is to determine whether the spinal circuitry involved in generating volitional and reflex-evoked synergies are similar. Wolpaw et *al.* and Wolf and Segal presented evidence that the spinal stretch reflex circuitry could be modified through operant conditioning. Therefore, the existence of a common circuitry for generating volitional movements provides support for the use of operant conditioning as a method of improving volitional function in individuals with spinal cord injury or stroke. To test the hypothesis that volitional and reflex-evoked movements employ common circuitry, a comprehensive comparison of muscle synergies observed in both types of movements under a range of movement excursions and velocities is needed.

We report here on the kinematic profiles and the spatial and temporal patterns of muscle activation observed in five subjects performing volitional elbow flexion and extension movements with varying movement excursions and velocities.

Methods: Subjects were seated in front of a computer monitor with the right forearm pronated and secured to a torque motor interface. The shoulder was stabilized around 70" of abduction and the elbow (100" of extension) was positioned coaxial to the motor shaft. Differential EMG activity was recorded through miniature surface electrodes placed over elbow flexors (both heads of Biceps Brachii and Brachioradialis), elbow extensors (Triceps Brachii long and lateral heads), wrist radial deviators (Extensor Carpi Radialis Longus and Flexor Carpi Radialis), and wrist ulnar deviators (Extensor Carpi Ulnaris and Flexor Carpi Ulnaris). Muscles controlling the wrist were monitored to allow for analysis of a potential programmed coupling between the elbow and wrist. Subjects were instructed to perform elbow flexion and extension tracking movements with varying movement excursions and velocities. Three movement excursions (10, 25 and 40") and five velocities (50, 150, 250, 350 and 450°/s) were used and the target width was fixed at 5". Velocity and excursion were independently varied across combinations. Subjects started each trial from a fixed location. The movement was performed 12 to 16 times for each excursion-velocity combination. The output of the elbow potentiometer was displayed on the monitor to provide subjects with feedback regarding the trajectory of movement relative to a target trace.

Data Analysis: The elbow position traces from each excursion-velocity combination were averaged for all trials and the overall accuracy of the performance relative to the target trace was determined. To determine the speed-accuracy tradeoff associated with the traces, movement time, movement amplitude and the amount of overshoot were quantified as a function of velocity and excursion.

The EMG records were filtered and rectified and a 15ms moving window was used to determine the time of onset of muscle activity. Spatial patterns of muscle activation were expressed in a binary fashion and temporal patterns were analyzed based on the time of onset of muscle activity and the sequence of muscle activation. The relationship between the movement kinematics and the spatial and temporal patterns of muscle activation was determined.

Results:

Kinematic Patterns: Fast velocity movements (150 to 450°/s) demonstrated a smooth and continuous transition between the hold and ramp phases of the movement for all excursions. Slow movements, on the other hand, possessed regions of discontinuity. These regions were most evident in the 40” excursion-50°/s velocity combination which required a movement time of 800ms.

While the movement overshoot increased with increasing velocities, the average overshoot, expressed as a percentage of the target movement excursion, decreased with increasing excursions. Primary movement time, the time (in milliseconds) needed to complete the excursion from onset of elbow movement to maximum overshoot, demonstrated a transition from wide to minimal scatter with increasing velocities. This held true for all three excursions. The observed transitional velocity was around 100°/s for the 10” excursions, and 130°/s and 140°/s for the 25” and 40” excursions, respectively.

EMG ***Patterns:*** Muscles operating both elbow and wrist were recruited for performing the elbow movements. The spatial patterns of muscle activation (number of times a muscle was active for a given excursion-velocity combination) did not seem to be affected by the movement excursion. A more observable effect was seen when the movement velocity requirement was varied. At low velocities the movement agonists (elbow *and* wrist extensors for elbow extension movements or elbow *and* wrist flexors for elbow flexion movements) were primarily active and with increasing velocities, antagonists became more frequently recruited. The spatial pattern of muscle activation became very consistent for movement velocities beyond 150°/s.

The effect of kinematic requirements on the time of onset of muscle activity was primarily exhibited in the antagonists, and to a lesser degree in the agonists. Aside from a delayed and more variable onset of agonist activity at the low velocities (50°/s), onset of activity seemed to be very consistent at higher velocities. Similarly, movement excursion did not seem to significantly affect agonist recruitment. On the other hand, the antagonist muscles were recruited sooner with increasing velocities while their onset was delayed with increasing excursions. The effect of movement excursion on the onset of antagonist activity seemed to diminish for velocities of 250°/s and higher.

Discussion: Our kinematic and electromyographic results tend to support the existence of synergies for generating high velocity movements. Synergies resulting from motor programs could be viewed as fixed patterns of muscle activation in which the set of muscles involved in executing the movement and their temporal sequencing are determined prior to the initiation of the motor act and in the absence of peripheral feedback. When such programs are deployed, fast, continuous and smooth movements are generated. Our kinematic patterns demonstrated that such continuous movements are generally obtained for movement velocities beyond 150°/s. At lower velocity movements, discontinuities in the movement trajectories were observed. These discontinuities could be primarily due to the contribution of feedback (visual) to the on-going correction of the movement trajectory. Our results also indicate the presence of a transitional velocity phase between discontinuous and programmed movements (velocities between 100 and 150°/s). The EMG patterns acquired in this study further support the existence of muscle synergies for performing elbow movements, involving both elbow and wrist musculature. At high velocities (250°/s and beyond), the timing of muscle activation was very consistent and did not seem to be affected by movement excursion. For such ballistic movements, velocity appeared to be the primary factor considered by the nervous system and the agonists and antagonists were activated within a few milliseconds of each other. Partial programming of activity may exist at moderate velocities (100 to 150°/s, requiring 250 to 270ms movement times for 40” excursions), where both the movement velocity and excursion requirements had an affect on the generated EMG patterns. At low velocities, the concept of muscle synergies or motor programs did not seem to be upheld. The spatial and temporal patterns of the EMG activity were variable with only a subset of the muscles activated. Under such movement requirements, visual and peripheral feedback could play a primary role in maintaining the proper movement trajectory and control of individual muscles could be more prevalent. Finally, the observed coupling between the elbow and wrist musculature could be indicative of the nature of Ia afferent connections between the motoneuronal pools of the respective muscles, possibly to reduce the effects of inertial coupling.

Acoustic Myography During Voluntary Isometric Contraction Reveals Non-Propagative Lateral Vibration

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INTRODUCTION

The recent technical progress in detection and signal processing has triggered the use of sounds emitted by skeletal muscles during voluntary contraction so called Acoustic myogram (AMG). However, these techniques are of limited interest because, up to now, there is no clear understanding of the underlying mechanism of these vibrations. In vitro experiments (1, 4), have shown that the vibration response of a muscle under contraction is transverse and can be described by the vibration equations of a cylinder or a sphere. The frequency of the pressure waves emitted by the muscle corresponds to that of its transverse mechanical resonance (2, 3). This frequency matching, for each state of contraction, shows that the muscular mechanical filtering is very efficient for frequencies corresponding to the fundamental eigenmode of vibration. The above analysis suggests that the states of mechanical vibration at the surface of the muscle are stationary. To test this hypothesis, we have analyzed the vibrations at two different longitudinal sites and two different sites perpendicular to the muscle's main axis at the surface of the skin overlying biceps brachii (BB) muscle in vivo during voluntary isometric contraction. The two recorded AMGs have been simultaneously recorded and compared through several signal analysis functions.

METHODS

Six healthy male subjects were tested, An ergometer, realized in the laboratory, secures shoulder, elbow and wrist joints in a vertical plane to minimize any participation of muscles other than the elbow flexors. The hand, semi prone, holds a vertical handle. The ergometer imposes a steady torque to the elbow joint. The biceps brachii AMG is recorded with two strain sensors developed in the laboratory (mass: 50 g; 4 Hz to 200 Hz, - 3 dB bandwidth). The sensors are coupled to the skin

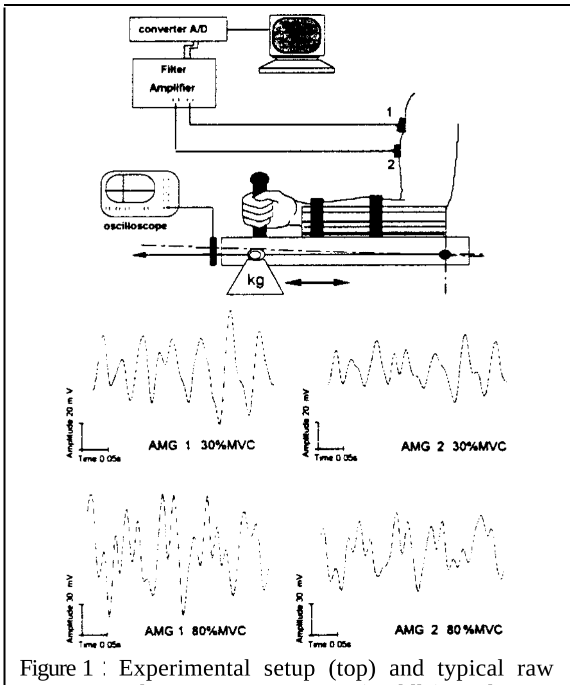

Figure 1 : Experimental setup (top) and typical raw AMG recording at 30% MVC (middle) and 80% bottom) from first (AMGI, left) and second (AMG2, right) sensors.

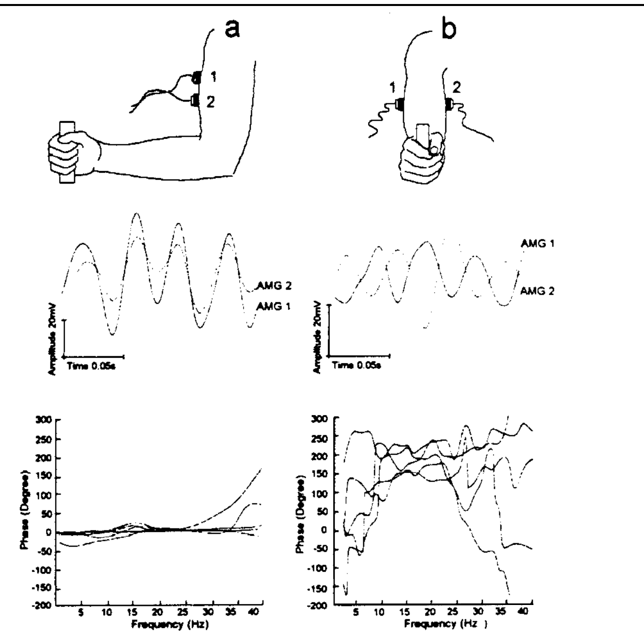

Figure 2 : Phase relationships at 30% MVC between the two AMGs with the sensors placed along the muscle's main axis (2a) or perpendicular to it (2b). Temporal signals for one subject (middle) and cross spectrum phase versus frequency for all subjects (bottom).

through double sided adhesive tape. Two configurations of sensors location are studied. In the first one, sensor 1 is placed on the belly of the muscle at it's thickest section and both signals are acquired for different positions of sensor 2 placed along the longitudinal axis of the muscle at 2 cm intervals above or below sensor 1. In the second configuration, the sensors are placed face to face on opposite sides of the muscle at its thickest section. To ascertain the hypothesis that the muscle follows a bending motion, for one subject the sensors were positioned at several distances between the elbow and the shoulder.

RESULTS

Figure 1 shows the general setting of the experiments and typical AMG recordings obtained respectively from the two sensors when voluntary isometric contractions are performed at 30% (upper trace) and 80% (lower trace) of maximal voluntary contraction (MVC). For all subjects and all of the experiments, AMG amplitude at 80% MVC was always greater than at 30% MVC.

The effect of sensor position over BB is illustrated by figure 2. The two AMG signals (AMG1 and AMG2) of a representative subject during one force trial at 30% MVC are displayed for both sensor setting. When the two sensors are aligned with the muscle's main axis, AMG1 and AMG2 are more or less in phase even though they do not have the same amplitude (figure 2a, intermediate traces). For the second sensor setting (figure 2b), AMGl and AMG2 are in phase opposition and their amplitudes are comparable. Phase relationships versus frequency between the two AMGs are shown for all subjects at the bottom of figures 2a and 2b. These figures show for each sensor setting respectively 0" and 180" mean phases over the resonant vibration frequency band which varies slightly according to the mechanical characteristics of each subject's muscle.

DISCUSSION

The use of simultaneous measurement of AMG signals at two different locations is a first step in spatial sampling and allows a better understanding of what vibration is taking place in the muscle by showing the relationship between the signals at different locations. A strong relationship, i.e. a coherence or cross-correlation value close to unity indicates a global vibration rather than a localized one. Unlike the motor unit action potentials which propagate along the longitudinal axis of BB muscle, the present results show that the mechanical counterpart of the contraction of in vivo muscle translates into a transverse resonant mode of vibration identical to experiments on other muscles in vitro (1, 4, 2). All points at the surface of the muscle located along a parallel to the longitudinal axis vibrate in phase: under voluntary isometric contraction, the muscle is in a stationary state of vibration. Furthermore, opposite points across the longitudinal axis of the muscle vibrate with opposite phases and thus indicate a bending mode of vibration. Therefore, the main component of AMG is the response of the muscle as a resonant structure to the local fluctuations of pressure and not a local response to these fluctuations created by the asynchronous summation of motor unit twitches depending on motor unit recruitment and individual motor unit firing rates.

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The relationship Between EMG Power Spectrum Statistics and the Rate of Rise in Torque during Electrical Stimulation of Elbow Flexor Muscles

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INTRODUCTION

Muscle fiber composition has been shown to influence both the speed of muscle contraction (Haekkinen et al., 1985; Ivy et al., 1981) and the distribution of power in the electromyographic (EMG) power spectrum (Gerdle et al., 1991; Kupa et al., 1995). Because a higher type II muscle fiber proportion is associated with faster muscle contractions and with higher mean power frequency (MPF) or median frequency (MF) values, a positive relationship between spectral statistics and contraction speed should exist.

The purpose of this study was to investigate the relationship between EMG power spectrum statistics (MPF and MF) and the rate of rise in torque during electrical stimulation of elbow flexor muscles.

METHODS

Eleven healthy volunteers (22 to 43 years, 5 women and 6 men) participated in this study. Subjects were seated with their right arm at shoulder height, the elbow flexed at 90 degrees and the forearm placed in neutral position. Isometric elbow flexion torques were measured using a transducer aligned with the elbow joint. Subjects produced the required torques by pulling against a wrist cuff mounted on a metal arm extending from the transducer. EMG signals were recorded with surface pre-amplified electrodes placed longitudinally along the muscle fibers of the biceps brachii (7 subjects) and brachioradialis (9 subjects) muscles.

Two to three maximal isometric efforts in elbow flexion were first produced, from which the highest was chosen as the maximum voluntary contraction (MVC) for a given subject. Supramaximal stimulation of the radial (brachioradialis) or musculo-cutaneous (biceps brachii) nerves was then performed. Single pulses of 0.5 to 1 ms duration were delivered percutaneously with increasing intensity until no further increase in the EMG (M-wave) and mechanical (isometric muscle twitch) responses could be seen. Thereafter, three M-waves and muscle twitches were recorded on computer for further analysis. Sampling frequencies were 10,000 and 5,000 Hz for the EMG and torque signals respectively.

The MPF and MF of the EMG power spectrum were obtained from 60 ms windows (raised cosine window processing, 1024 points FFT) taken from the onset of the M-waves. The rate of rise in the torque signal was calculated between 30 and 70% of the peak force for each muscle twitch by fitting a linear regression line through the data points, and calculating the slope of this line (expressed in %MVC/s). Pearson product moment correlation coefficients were calculated for the MPF and MF data of both muscles.

RESULTS

Significant negative correlations were obtained between the MPF or MF values and the rate of rise in torque. Correlation coefficients ranged from -0.66 to -0.80. Figures 1 and 2 show data for the MPF of both muscles.

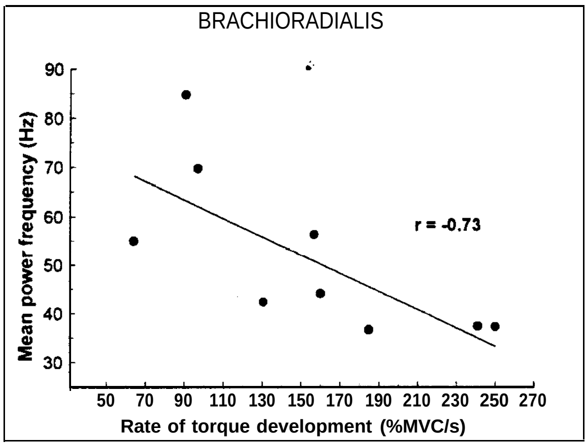

Figure 1

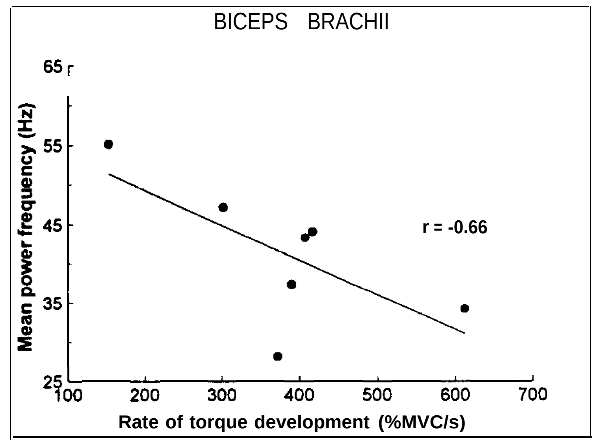

Figure 2

DISCUSSION

Positive correlations between MPF or MF values and the rate of rise in torque were expected due to the previously reported association between these variables and muscle fiber composition. Significant correlations were obtained in this study, however, they were found to be negative. Consequently, this suggests that factors other than muscle fiber composition may play an important role in explaining the variation in contraction speed across different subjects. This is in agreement with recent findings on isometric twitch contractile properties of elbow flexor muscles (O'Hagan et al., 1993).

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EMG to Force Processing of Electrically Stimulated Muscle

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Introduction:

Many studies have addressed the issue of estimating muscle force from EMG during voluntary isometric contractions. Recently, electrical stimulation has been used to identify model parameters, such as force-length relations of individual muscles (Leedham and Dowling, 1995), in order to enable the prediction of muscle forces from EMG during voluntary dynamic contractions. In such cases, the stimulus artifact makes it difficult to examine the relationship between EMG and muscle force. The purpose of this study was to develop a method to remove artifact and determine a reliable transfer function between isometric force and the M-waves elicited by electrical stimulation.

Methods:

The biceps brachii muscle was selected because it has been one of the most widely used muscles in voluntary contractions. Biceps is also notoriously difficult to obtain M-waves when the stimulation protocol elicits force levels approaching MVC values. Four male subjects with no history of neuromuscular problems volunteered for the study. The right arm of each subject was secured to the testing jig at an angle of 90 degrees and the cathode was placed distal to the elbow joint. The anode was positioned over the motor point such that a maximum twitch was evoked. A bipolar EMG electrode with two silver bars, 8mm apart was then positioned such that the clearest M-wave possible was recorded. Often, a clear separation of M-wave from artifact could not be established. A stimulus frequency of 50 Hz was used and the voltage was manually increased and decreased by the experimenter such that a ramp increase and decrease of isometric force to a level of 60% MVC was obtained. The EMG and elbow torque were each sampled at 1250 samples per second and stored on computer.

A stimulus artifact template was established from the initial low voltage stimuli that did not evoke any force. This template was used to remove the artifact and leave an EMG record that only contained the M-waves. This EMG signal was then full wave rectified and systems identification was used to determine the transfer function for EMG to force processing. The EMG to force response was compared to voluntary ramp contractions which were processed in the same manner.

Results and Discussion:

Even though the artifact over-lapped the M-waves in all subjects, the template removal was quite successful in subtracting the artifact and leaving uncontaminated M-waves. The transfer functions were similar in terms of the filter time constants between the stimulated and voluntary contractions and allowed good estimates of the measured torque (Figure 1: Top). The linearity was slightly different which probably reflected the different role that firing rate played in the two contraction types. Figure 1 (Bottom) shows a nonlinear relationship between the EMG and torque that is opposite to that found in voluntary contractions. The firing rate could obviously be increased in voluntary contractions during the ramp increase but this rate was fixed at 50 Hz in the stimulated condition of this study. In either case, the nonlinearity is easily calibrated and good predictions of torque can be achieved.

Reference:

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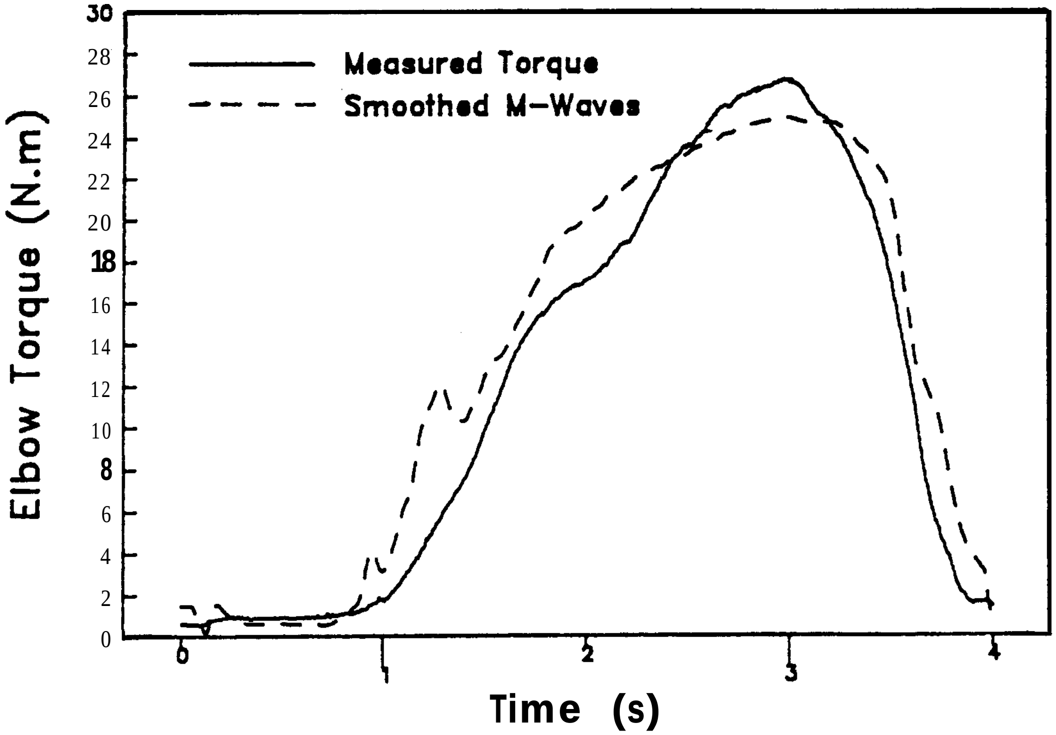

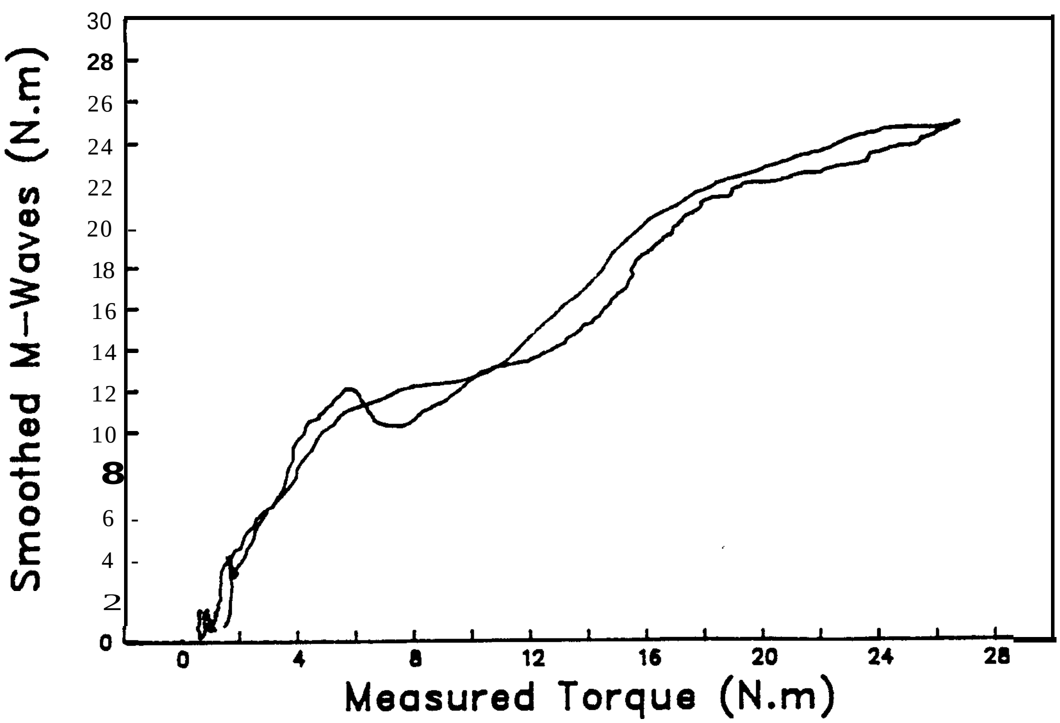

Figure 1. Top: Comparison of measured elbow torque with that predicted by smoothed M-waves following artifact removal and rectification. Bottom: Nonlinear relationship between the processed EMG and the measured elbow torque of a single subject.

Identification of Muscle Activation Patterns: A Developmental Perspective

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INTRODUCTION The acquisition of movement skills unfolds gradually over a time scale during the childhood years. A maturational indicator is the capacity to generate movement patterns referenced to the surface characteristics of environmental interfaces, allowing complex behaviors such as hopping and jumping to be performed. It is postulated that these complex movements are *learned* through self-organizing strategies [1], using trial and error to gradually establish stable and accurate movement solutions for a specific motor skill. Interestingly, self-organizing strategies have usually been defined using inverse dynamic equations with less emphasis being placed on the gradual modifications in muscle activation patterns. However, as environment-based movements are dependent on an accurate matching of limb compliance with the characteristics of the interface acted upon, it is proposed that a more comprehensive definition of the muscle activity patterns associated with the acquisition of a specific movement skill would be essential.

OBJECTIVES

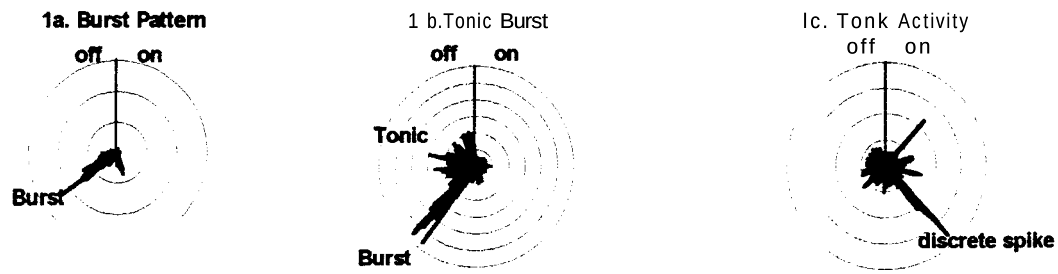
1. To formulate a working definition for muscle activity patterns seen in complex movement skills.
2. To implement these definitions to characterize the modifications in muscle activity patterns associated with the development of skilled landing from a jump.
3. To associate these changes to a biomechanical outcome.

METHODS Nine children (6 to 12 years of age) performed jumps from a platform set at the height of the tibial plateau onto a rigid floor. Surface EMG was recorded for 6 right lower extremity muscles: tibialis anterior, lateral gastrocnemius, soleus, Vastus lateralis, rectus femoris, and biceps femoris. The signal was differentially pre-amplified, band-pass filtered (lo-500 Hz), amplified, and converted to digital signal off-line (1 KHz sampling frequency). Stationarity was ensured by performing linear detrending of the EMG data prior to analysis. The onset of purposeful activity was defined as a rise above the 95% confidence interval of resting baseline activity [2]. For the kinematic analysis, the jumps were video-taped in the right sagittal plane (60 Hz) and analyzed off-line (PEAK Performance Technologies). The body was modeled as a six linked-segment body and the following relative angles obtained: trunk, hip, knee, and ankle. The digital EMG and kinematic data were further elaborated using EXCEL (Microsoft) and MATLAB (MathWorks).

RESULTS Three separate muscle activation patterns were identified in preparation for landing and defined in the time domain. ***a)*** BURST activity consists of a smooth distinct rise in activity from resting baseline, increasing gradually to peak mid-burst, and subsequently decreasing to baseline.

b) TONIC BURST activity consists of a sustained increase in the level of muscle activity above resting baseline, with a characteristic BURST pattern embedded within the increased activity.

c) TONIC activity consists of a sharp rise in muscle activity followed by a sustained level of EMG over an extended period of time (> 50 ms). TONIC activity can be unambiguously contrasted from the TONIC BURST by the absence of a local maximum. However, maximums may occur as a consequence of spiking transients in the EMG signal. These transients are discrete in nature and thus can be effectively differentiated from the EMG peak in the TONIC BURST pattern by calculating the mean trend in a 20 ms window prior to and after the maximum. In TONIC activity, a stationary mean trend line will be observed across both windows, A robust identification of the three activation patterns was obtained by plotting the EMG signal along a radial axis ***(Figure la,b,c)*** ***In*** these radial plots, the duration of the activity is expressed in radians/sec and the amplitude of the EMG in arbitrary units as a function of resting baseline activity. In ***Figure la,*** one distinct BURST is identified, with activity returning to resting baseline outside of the burst boundary. In ***Figure lb,*** a burst is initially identified followed by a sustained increase in the level of activity. In contrast, ***Figure Ic*** shows a non-specific increase in the level of EMG rising initially from baseline resting levels and being maintained throughout the sampled period.



The pattern identification method (PIM) was used to establish a correlation between the emergence of landing skill and a progression from reliance on TONIC activity by the younger children to the use of BURST activity appropriately timed to landing in the older children. By superimposing the radial activity plots onto concomitant angular joint displacement, we were further able to associate the type of muscle activity with a particular pattern of inter-segmental movement of the lower extremity in preparation for landing. The gradual change from TONIC to BURST activity was associated at the kinematic level with a gradual shift from using end-range extension joint positions to a more flexed landing position. Joint flexion at landing affords the older child a smoother control of post-landing stability by minimizing the oscillations experienced at the center of gravity. Interestingly, the TONIC BURST pattern was observed in children who were in the transitory phase of skill acquisition. These children used either the extended or flexed landing positions on different trials.

DISCUSSION The gradual changes observed in the patterns of muscle activity associated with the different joint operating ranges may directly impinge on the compliant properties of the lower extremity and thus directly influence the different post-landing stability profiles seen in younger and older children. It is possible that the different muscle activation patterns afford different mechanisms for stiffness control of the linked-segments depending on the child's ability to accurately anticipate the effect of the ground reaction force on post-landing joint positions. Statistical analyses are presently being applied to determine the significance of the correlation between joint positioning, the type of muscle activity, and the level of skill. However, these preliminary results suggest that the PIM can be easily implemented and is sufficiently robust to further define the self-organizing strategies that are naturally used during development for the emergent control of skilled motor behavior.

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Repeatability of Surface EMG Variables During Voluntary Isometric Contractions of the Biceps Brachii Muscle

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INTRODUCTION. The repeatability of initial value and rate of change of mean spectral frequency (MNF), average rectified values (ARV) and muscle fiber conduction velocity (CV) was investigated in the biceps brachii of ten normal subjects during sustained isometric voluntary contractions. Four levels of contraction were studied: lO%, 30%, 50% and 70% of the maximal voluntary contraction level (MVC). Each contraction was repeated three times in a session and each session was repeated in three different days for a total of nine contractions/level/subject and 90 contractions per level across the ten subjects. Repeatability was investigated using the Intraclass Correlation Coefficient (ICC) and the standard error of the estimates for each subject. Contrary to observations in other muscles, CV estimates appeared to be very repeatable both within and between subjects.

EXPERIMENTAL PROTOCOL. Ten healthy male subjects (age from 23 to 43 years) participated in this evaluation after giving informed consent . Each subject lied on his back with the right dominant arm abducted at 90” and the forearm at 120° with respect to the arm. The arm and forearm were placed in an isometric brace equipped with two torque transducers (one on each side of the arm) and connected to a display which provided the subject with visual feedback about the torque level he produced. The protocol consisted in three experimental sessions repeated in three different days. Three maximal voluntary contractions (MVC) lasting 3-5 sec. were performed at the beginning of each session. Visual feedback was provided. The highest of the three values was taken as the MVC value for that session. Targets were set on the visual feedback display at l0%, 30%, 50% and 70% of the MVC value. Voluntary contractions of the biceps brachii were then performed in random order with a 5 min. rest interval between subsequent contractions. Each lasted for 30 sec. and was repeated three times. One single and two double differential myoelectric signals (SD, DD) were detected with a 4 bar system consisting of 4 silver electrodes 10 mm long, 1 mm diameter, 10 mm apart. The signals were amplified with a 10 Hz to 450 Hz bandwidth (40 dB/decade slope on each side), sampled at 1024 Hz, digitized by a 12 bit ND converter and stored on the disk of a personal computer. In no case any significant power was observed at frequencies above 400 Hz. Skin temperature was monitored and kept between 31.5C° and 32.5C° by means of a fan blowing air with adjustable temperature.

STATISTICAL PROCESSING. The time course of MNF, ARV and CV variables showed a linear behavior that was fitted by a least square regression line whose initial values and slope were computed. The Intraclass Correlation Coefficient is a commonly used index of repeatablity and has been used in this study. This coefficient indicates the percentage of the global variance that can be attributed to the variability between subjects. The remaining percentage of variance is attributed to repeated trials (in the same session) or sessions (days) and is considered “experimental noise”. It is current practice to accept as “excellent repeatability” ICC values in the range 80%-100% and “good repeatability” ICC values in the range 60%-80%, while values below 60% indicate “poor repeatability”.

RESULTS. Table I shows the results of the ICC analysis for the initial values of MNF, ARV and CV. The ICC values reported indicate that MNF initial values are the most “reliable” parameters in the sense that a) are the most repeatable across experimental sessions and trials performed on the same subject and b) are different in different subjects. Initial values of ARV show higher ICC values at higher contraction levels. CV displays a very small intersubject variability at 50% MVC and 70% MVC: when the variance within subjects is larger than the variance between subjects the concept of ICC becomes questionable and its values become artificially low or negative and must be disregarded (TableII).

Fig.1 depicts initial values of MNF, ARV, CV for each subjects, while Fig.2 depicts same parameters versus the contraction level. A small, but statistically significant, increment of the initial value of CV is evident when the contraction level increases from 10% MVC to 30% MVC while a small, but statistically significant, decrement of initial value of MNF is evident as the contraction level increases from 10% MVC to 70% MVC.

Table I. Repeatability indices of myoelectric signals in voluntary contractions. Intercepts of linear regressions (biceps brachii muscle).

	MNF	ARV	CV
%MVC	ICC	ICC	ICC
10	60.2	38.0	40.1
30	67.8	52.1	25.8
50	71.8	77.7	-21.8
70	62.6	74.1	-27.1

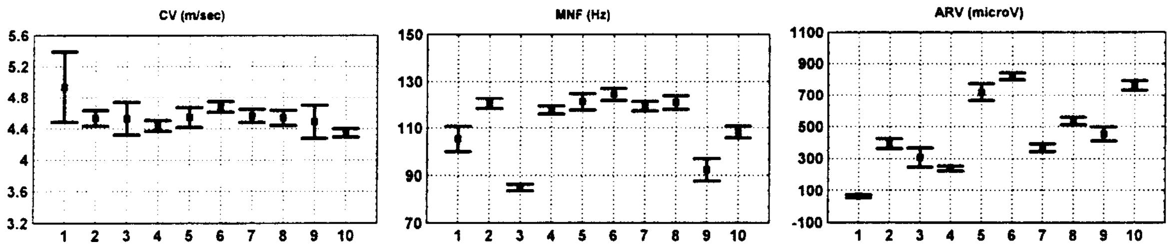

Figure 1. CV, MNF, ARV intercepts. Means and standard error (70%MVC, 10 subjects, N=9)

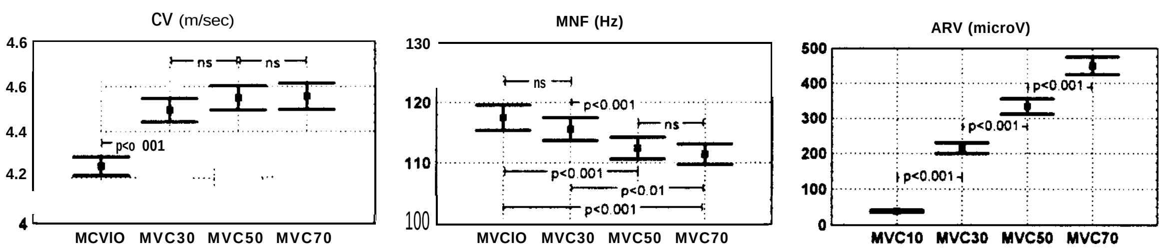

Figure 2. CV, MNF, ARV intercepts at different MVC levels (p=level of significance of the paired Wilcoxon test, N=90)

As expected, ARV increases markedly with the level of contraction. It is well known that a decrement of CV (assuming a single CV value for all the active motor units) would imply spectral compression, without spectral changes of shape, and equal percent changes of CV, MNF and MDF. As a consequence, the normalized slopes of these variables should be identical but this is rarely the case in practice. As depicted in Fig.3 our experiments show that at 70% MVC the percent rate of decrement of MNF is greater than that of CV mostly because of an offset of about 0.5%/s.

Figure 3. Normalized slopes of MNF(70%MVC) (%/sec) versus normalized slope of CV(70%MVC) (%/.sec)

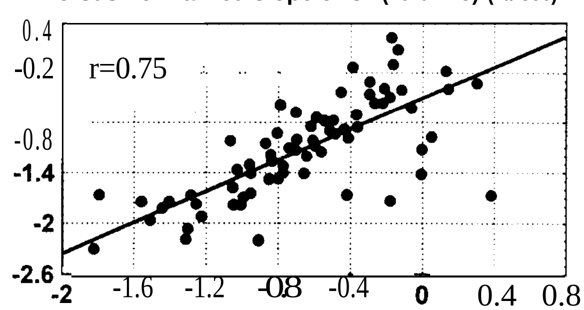

Table II. Values of normalized standard error of means within subjects (N=9) and between subjects (M=lO). Initial values of linear regressions (biceps brachii muscle).

	MNF		ARV		CV	
%MVC	Within(%)	Between(%)	Within(%)	Between(%)	Within(%	Between(%)
10	3.04	4.69	12.15	15.29	1.88	2.21
30	2.59	4.49	12.57	17.62	2.52	2.47
50	2.31	4.37	7.46	18.29	3.09	1.5
70	2.68	3.84	9.66	15.73	3.15	1.1

ACKNOWLEDGEMENT. This work was supported by the European Concerted Action SENIAM, Camera di Commercio di Torino, Compagnia di San Paolo, Fondazione CRT di Torino, and Regione Piemonte.

Preliminary Investigation of Muscle Recruitment Patterns in Dynamic Movements of the Shoulder Using Fine Wire Indwelling Electromyography

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INTRODUCTION

Precise neuromuscular coordination by more than 20 muscles is necessary for normal shoulder function. Pathology of the shoulder is associated with dysfunction of the rotator cuff and scapular-positioning muscles. Previous fine wire electromyography (EMG) investigations of dynamic shoulder movements have focussed on EMG amplitude in the rotator cuff and larger superficial muscles in younger and athletic populations. Since Inman et al's study', recruitment patterns of shoulder muscles, particularly of the scapular-positioners, have not been extensively studied. Critical differences in the timing parameters of muscle activity have been demonstrated between symptomatic and asymptomatic subjects in low back pain studies[2]. Research on symptomatic athletic shoulders has also indicated differences in timing [3]. For clinical range of movement (ROM) tests, minimal EMG research has been done. Few studies reporting the reliability of fine wire EMG in investigations of dynamic shoulder muscle function have been found.

The focus of the study was to record EMG activity from selected shoulder muscles using fine wire electrodes, to correlate EMG recruitment patterns to angular movements at the shoulder during ROM tests and to make preliminary assessment of the reliability of the instrumentation and protocols.

METHODS

Two men (57 ± 1 yrs, hgt 1.8 ± .03m, wgt 87.5 ± 5.5kg) and three women (21-56 yrs, hgt 1.65 ± 0.5m, wgt 56.5 ± 8.5kg) with no history of shoulder problems volunteered for testing of their dominant shoulders. All subjects were right hand dominant.

Movements tested were abduction in the coronal plane followed by flexion in the sagittal plane. Two dimensional recording of movement angles was performed by a motion analysis system (Motion Analysis™ Expert Vision System™) tracking hyperreflective markers attached to bony landmarks on the scapula, humerus, ulna and vertebral column. Subjects self-selected movement velocities. Three practice trials preceded one ten-second data collection trial. For one subject, data was recorded for six successive abduction trials within one session in addition to one abduction trial on two additional days.

Bipolar fine wire electrodes were made based on the technique described by Basmajian and Stecko[4], using Teflon-coated stainless steel wire (bare diameter 0.003" for 2mm) and 23 gauge hypodermic needles. Sterilised electrodes were inserted by a medical practitioner using aseptic techniques. The signals were amplified, bandpass filtered from 10 to lOOOHz, A/D converted at 2000 samples per second and stored in a PC.

EMG recordings were only possible for seven muscles simultaneously - middle deltoid (for abduction) or the clavicular head of pectoralis major (for flexion), supraspinatus, infraspinatus and four scapular positioners (upper and lower trapezius, levator scapulae and serratus anterior).

Standardised sites, depths and angles for insertion of electrodes were predetermined. EMG signals were previewed on an 8-channel monitor and correct positioning confirmed during resisted practice trials. Recordings of muscle activity were synchronised with movement traces by the motion analysis system. Data was analysed using Acqknowledge 3.2.4 software on a personal computer. Movement and raw EMG traces were juxtaposed for visual comparison of EMG onset and offset times and patterns of activity in relation to movement parameters.

RESULTS

Data were recorded for 5 subjects for abduction and 4 subjects for flexion. There was reasonable inter-subject consistency in range, velocity and patterns of movement. Flexion was performed faster than abduction, female subjects had greater ROM and elevation time was relatively shorter. One male subject (the outlier) had much slower movements, a longer relative elevation time and restricted ROM. Middle deltoid, pectoralis major, supraspinatus and serratus anterior showed EMG onset activity prior to onset of movement in all subjects for both abduction and flexion. Infraspinatus generally showed pre-movement onset activity especially for flexion. Upper trapezius and levator scapulae increased activity immediately before or at movement onset except during abduction by the outlier subject. Onset of EMG activity in lower trapezius was post-movement onset in abduction for all subjects and at or post-movement onset for flexion. EMG activity persisted in several muscles after cessation of movement. Some muscles showed consistent patterning between subjects. One subject, tested on three days, including successive abduction testing on one day, showed considerable consistency in EMG onset times for individual muscles.

DISCUSSION

Full-range active abduction and flexion are commonly tested in clinical assessment of shoulder dysfunction, but basic information about spontaneously-paced speed and patterns of these movements is lacking. Results from this study indicate some movement pattern similarities consistent with published information. Flexion was done faster than abduction which may reflect daily practice'. EMG onset times were also consistent with previously published findings and can be supported by knowledge of normal shoulder kinematics and kinetics.

Data from this study indicates that relating EMG onset and offset times to movement parameters is a valid method of investigating shoulder muscle function. The instrumentation and protocols appear to be satisfactory. Reliability requires further investigation with increased sample size for different ages and genders. Problems with subject selection criteria were identified, as data for the outlier subject differed in speed, range and relative timing despite the absence of shoulder problems. There is a need for stricter inclusion and exclusion criteria to include objective movement analysis by expert observers.

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Computer Simulation of Generation of Electromyograms for the Pulse Density Demodulation Processing

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PURPOSE OF COMPUTER SIMULATION

Principle of pulse density demodulation processing of electromyograms (PDD-EMG) and a result of the processing for the electromyogram (EMG) of the lower leg muscles were reported at the previous congress (References). In this paper, a computer simulation of generation of EMG was executed, according to a strategy, which has been proposed by the present authors in this congress, for assessing the accuracy of reconstructing motor commands by means of PDD-EMG.

PROCEDURE OF GENERATION OF EMG

Following the sequence of the computer simulation program (shown in Table1), neural pulse trains in each of the motoneurons before and after synapse were generated from on-off pattern of motor command, and the waveform of bipolar electrode EMG were synthesized (as shown in Figure 1). The characteristics of neural transmission were set by refering to the physiological data, and number of motor units was choosen to be large enough for representing the motor commands of a whole muscle. A tri-phasic waveform of the action potential (as shown in Figure 2) was used for the synthesis. Then, the pulse trains was extracted from the EMG waveform by pulse shaping, and the motor command of a whole muscle was reconstructed by measuring charge in pulse density.

RESULT ON ACCURACY OF PDD-EMG

As the result, it was shown (in Figure 3) that the reconstructed analog pattern of the motor command by means of the PDD-EMG corresponded fairly accurately to the theoretically generated one, except for a few missing pulses in the motor units far from the electrodes, and a few false pulses caused by interference between high density pulse trains. Based on this result, it becomes possible to farther assess the accuracy of PDD-EMG when reflecting neural pulses from the sensors in the muscle are introduced.

Table 1. Sequence of the computer simulation program for generation of EMG, and PDD-EMG.

1. Specification of the on-off pattern of motor command for a whole muscle.
2. Assignment of average characteristics of motoneurons before synapse and their random fluctuation.
   1) Sensitivity of pulse generation in each pulse generator.
   2) Refractory period in each pulse generator.
3. Generation of neural pulse train in each motoneuron before synapse.
4. Assignment of average characteristics of motoneuron after synapse and their random fluctuation.
   1) Delay time of pulse regeneration at each synapse.
   2) Refractory period at each synapse.
5. Generation of neural pulse tram in each motoneuron after synapse.
6. Synthesis of the EMG waveform.
   1) Assignment of the waveform of tri-phasic action potential.
   2) Assignment of location of the motor units around the electrodes.
   3) Detection of the EMG waveform by bipolar electrodes.
7. Extraction of the pulse trains from the EMG waveform by pulse shaping.
8. Reconstruction of the analog pattern of the motor command by measuring pulse density.
9. Generation of the theoretical analog pattern of the motor command.

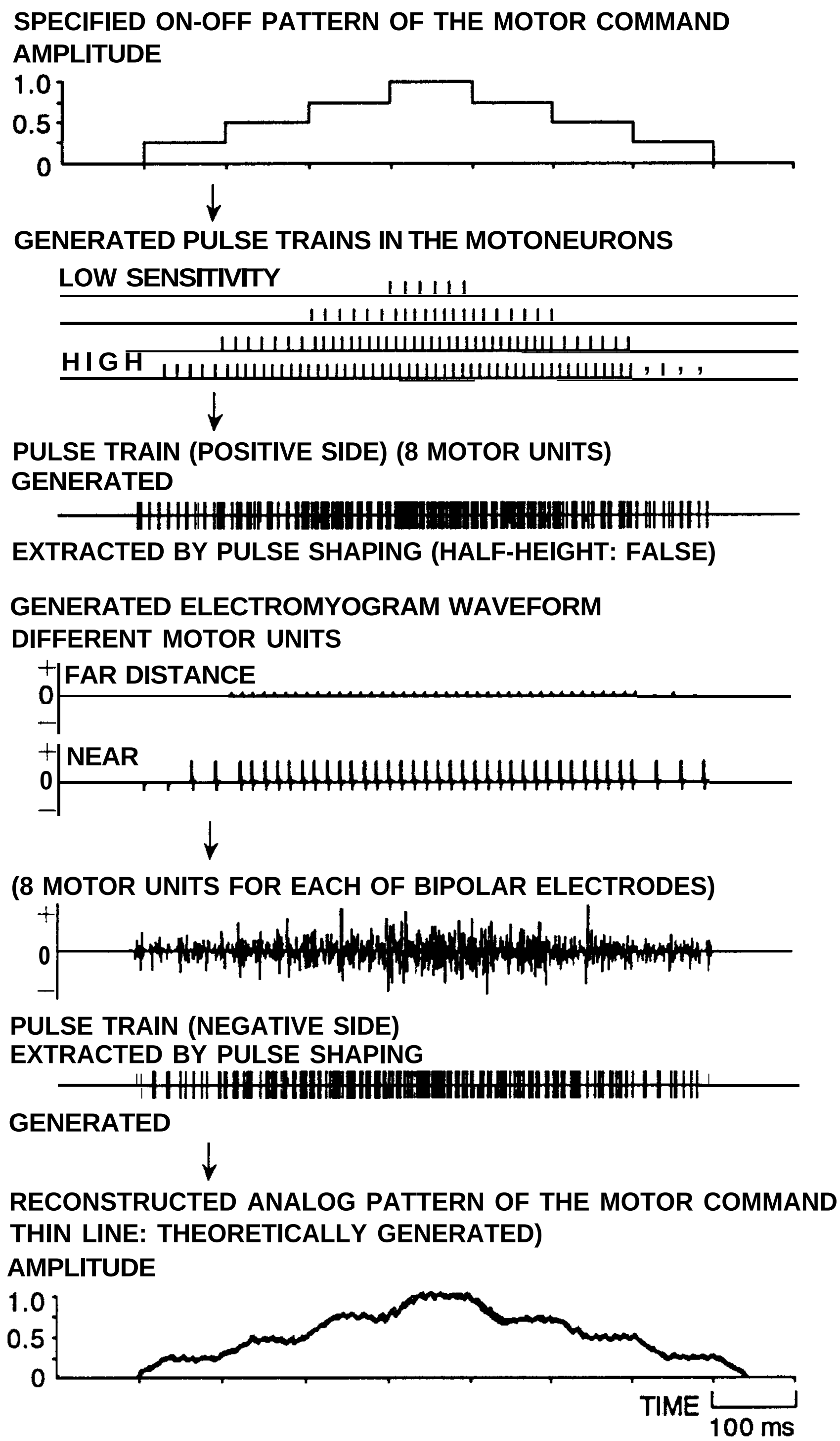

Figure 1 (Left). A series of examples in each steps of the process of computer simulation of generation of EMG.

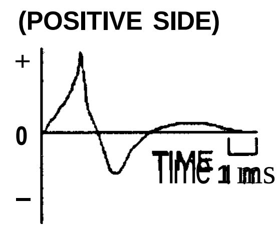

Figure 2. Normalized tri-phasic waveform of the action potential used in synthesis of EMG.

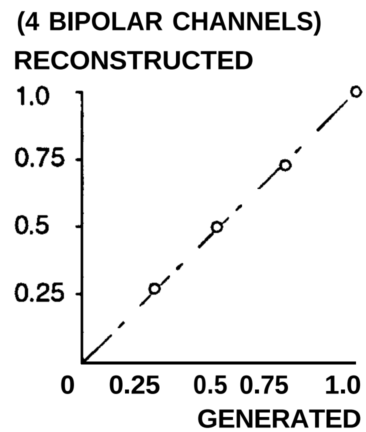

Figure 3. Correspondence of reconstructed analog pattern of the motor command to the theoretically generated one.

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EMG Amplitude Estimation During Dynamic Contractions

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INTRODUCTION

Typical EMG amplitude estimators use a fixed smoothing window length. When the amplitude is dynamic, varying the window length as a function of time has been reported to improve amplitude estimation[2,4,5]. When the amplitude is changing rapidly, the window should be short; when the amplitude is changing slowly, the window should be long. This report describes a study of adaptive window length processing. The technique was studied using a stochastic simulation model of the EMG waveform and then evaluated experimentally. A preliminary report of this work has appeared[1].

OPTIMUM ADAPTIVE WINDOW PROCESSORS

For dynamic contractions, it has been shown that the mean square error (MSE) in the EMG amplitude estimate can be written as the sum of two components: variance (due to random fluctuations in the EMG amplitude estimate about the true amplitude) and bias (due to errors in tracking true changes in the amplitude)[2,5]. Using this criterion, a minimization problem was formulated to determine the optimum window length. The EMG *waveform* was modeled as a stationary, bandlimited, random process multiplied by the EMG *amplitude*. Using this model, Hogan and Mann[3,6] have given an analytic formula for the variance error. For the bias error, the EMG amplitude in a local neighborhood was allowed to vary either in a linear or quadratic fashion. With these assumptions, the bias error could then be determined when using a moving-average mean absolute value (MAV) EMG processor. The resulting MSE was then used to determine the optimum window length.

Note that the optimal window length is a function of the true EMG amplitude and its derivatives. In practice, these values are not known and must be estimated from the EMG waveform. Hence, adaptive processing is implemented in two passes. In the first pass, fixed-length processing estimates the EMG amplitude and its derivatives. The optimal window length is then selected for each sample index. In the second pass, the *waveform* data are re-smoothed, using these window lengths.

SIMULATION STUDY

Investigation showed that the derivatives of EMG amplitude could not be adequately obtained with simple differencing filters. Thus, polynomial filters were investigated. For all conditions, lower errors were found using non-causal filters compared to causal filters. Errors grew as the EMG amplitude bandwidth increased. For non-causal filters, polynomial degree had little or no influence on derivative errors. For causal filters, a degree one polynomial was best for the first derivative and a degree two polynomial was best for the second derivative. The number of samples over which the polynomial filter should be fit varied inversely with the EMG amplitude bandwidth.

The expected performance of the adaptive window length algorithm versus fixed length algorithms was next simulated. Adaptive estimators were evaluated with the derivatives known (to establish “ideal” technique performance) and with the derivatives estimated from the EMG waveform (as would be done with actual data). The preferred length polynomial differentiator was used for each EMG amplitude bandwidth. The results showed that for each EMG amplitude bandwidth, the adaptive algorithm (with estimated derivatives) performs about as well as the *best* fixed-length algorithm and better than all other fixed-length algorithms. Similar results were found with non-causal processing.

EXPERIMENTAL STUDY

Real-time EMG processing of the biceps and triceps muscles was studied. Four electrode sites were monitored for each muscle. Subjects produced constant-angle contractions, modulating their EMG amplitude to track a random target presented on a computer screen. The target moved either as a (1) 0.25 Hz bandwidth uniform random target (ranging from 50% flexion effort to 50% extension effort), or (2) 1.0 Hz bandwidth uniform random target, or (3) binary random target (alternating between 25% flexion effort and 25% extension effort) remaining at each level for 2-5s (the duration was chosen randomly from a uniform density). Four EMG processors were studied: (1) a single-channel fixed-length (250ms) processor, (2,3) four-channel fixed-length (lOOms, 250ms) processors, and (4) a four-channel adaptive length processor.

Comparing the fixed-length 250ms processors, the multiple-channel processor performed better than the single-channel processor for all target types. For the 0.25 and 1.0 Hz random targets, no advantage was found due to adaptive processing. For the random binary target, the adaptive processor displayed both a rapid response to level transitions (similar to the l00ms fixed-length processor) and low variance while the target was fixed (similar to the 250ms fixed-length processor).

SUMMARY AND CONCLUSION

A technique for adaptive window length estimation of the amplitude of the non-stationary EMG waveform was derived. This method includes consideration of the first and second derivative of the EMG amplitude. Simulation and experimental studies found the advantages of the adaptive processor to be situation dependent. In the simulation study, it was found that practical adaptive detectors, *with optimum selection of a polynomial derivativejlter,* should work as well as the *optimum* fixed-length processor. For such situations, a fixed-length detector is likely preferred, because the distribution of error between bias and variance is more stable. Future research should be directed towards improved derivative filters which may alter this conclusion.

Experimentally, it was confirmed that multiple-channel processors performed better than the single-channel processor. When the EMG amplitude varied as a band-limited uniform signal of known, *fixed* bandwidth, then adaptive window length processing did not provide an advantage over fixed-length processing. In other situations (e.g., when EMG amplitude changed in a random binary fashion or when the processor was not allowed to tune its derivative filter length and fixed window smoothing length to the signal bandwidth), adaptive window length processing may be advantageous.

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Effects of the Mode and Velocity of Isokinetic Contractions on the Surface EMG Parameters

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INTRODUCTION

It has been well documented in isometric contractions that surface EMG magnitudes are quasi-linear to the force generated, and the mean power frequency lowers with development of muscle fatigue (Basmajian and De Luca, 1985). Muscular exertions in daily life, however, are conducted by anisometric contractions except in postural activities. While the merit of surface EMG is its applicability to a contraction however dynamic it may be, the behavior of surface EMG parameters in anisometric contractions has been poorly documented except in earlier studies (e.g.Bigland and Lippold, 1954) in which the angular velocity and the line of applied force are not controlled simultaneously in the joint rotation. This study is designed to survey the attitude of surface EMG parameters in anisometric contractions controlled for the above variables by using an isokinetic training apparatus.

METHODS

Using an isokinetic training machine (BIODEX, Biodex Medical Systems), 7 male subjects performed concentric and eccentric elbow flexions with maximum efforts. The subject was seated and placed the elbow on a support, with the upper arm slightly abducted and moderately flexed. Adjustments were made to coincide the rotational axes of the elbow joint and BIODEX.

The subject was asked to grasp a handle attached to the rotating arm of BIODEX, with the forearm supinated, and to make maximum exertion to rotate, or to resist to be rotated by, the arm. The angular velocity of the contraction was set at each 60 degrees interval from 30 to 270 degrees / sec for concentric contractions, while from 30 to 150 degrees / sec for eccentric contractions. A set of 2 consecutive contractions for each condition of the angular velocity was repeated by each subject.

Surface EMG was led bipolarly from the short head of biceps brachii and brachioradialis muscles, with Beckman-type disc electrodes 3 mm in the contact surface diameter, placed 15 mm apart along the muscle fiber. Myoelectric signals were digitized and processed with a personal computer, together with the angular displacement and velocity of, and the torque generated by, the elbow joint as detected by the isokinetic machine. The torque and rectified EMG were averaged for each 20 degrees interval of the elbow angle from 0 to 120 degrees, with the full elbow extension defined as 0 degree. EMG mean power frequency was calculated for each 40 degrees interval of the elbow joint from 0 to 120 degrees.

RESULTS AND DISCUSSION

Mean rectified EMG (MRE) averaged collectively for joint angles and angular velocities was not different between the mode of contraction, except for the brachioradialis in which the EMG magnitude was significantly greater in concentric than in eccentric contractions. While the MRE averaged for angular velocities increased in eccentric contractions with decrease in muscle length, as confirmed significant by ANOVA, the trend was inconsistent between muscles in concentric contractions. Such a consistent change of EMG magnitude with muscle length cannot be attributed to the location of electrodes relative to the innervation zones. Irrespective of the mode of contraction, the angular velocity of flexion did not affect the MRE averaged for joint angles. The independence of EMG magnitude from the angular velocities seems reasonable considering that each contraction in this study was performed with the maximum effort, thus activating motoneurones to the physiological limit in each contraction.

Mean power frequency (MPF) averaged collectively for joint angles and angular velocities was significantly higher in concentric than in eccentric contractions except in the brachioradialis in which MPF was not different between both modes of contraction. While the MPF averaged for angular velocities increased with decrease in muscle length (Fig. l), as confirmed significant by ANOVA, this trend was inconsistent between muscles in eccentric contractions. What found with the concentric contractions are comparable to those reported for isometric contractions (Okada, 1987). Irrespective of the mode of contraction, MPF averaged for joint angles increased with the angular velocity (Fig. l), which was confirmed to be statistically significant by ANOVA. A possible reason, yet to be substantiated experimentally, should be selective recruitment, within the physiological limit, of motoneurones innervating muscle fibers with higher conduction velocity, proportionately with increase in the contraction velocity.

CONCLUSION

In this study, the velocity of joint rotation and direction of the force applied were controlled in dynamic contractions. As a result, it was demonstrated that the muscle length -and velocity of contraction affect the surface EMG parameters differently according to the mode of contraction, or between muscles, and, in particular, that the velocity of contraction has varied influences upon the EMG magnitude and spectral profile, respectively. Further experimental analyses are warranted to clarify individually the mechanism underlying these findings.

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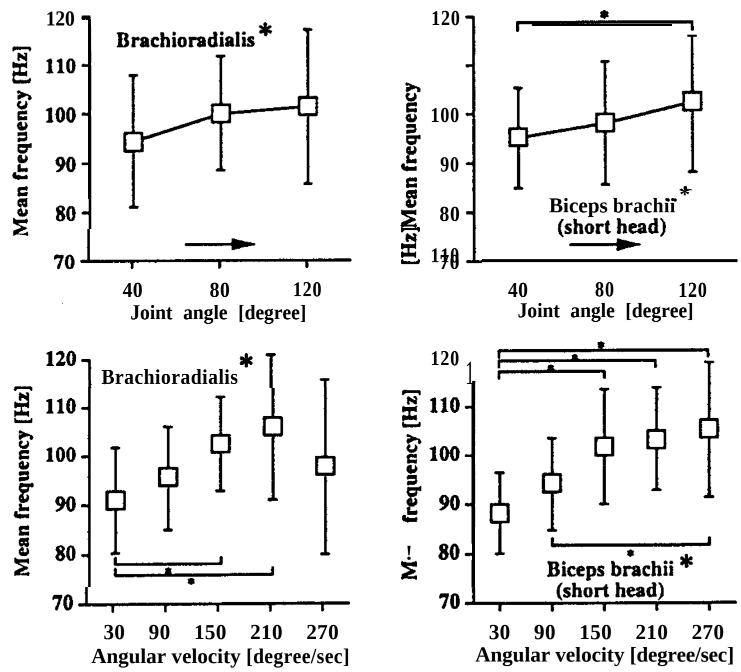

Fig. 1 Relations between the elbow joint angle and mean power frequency (top), and between the angular velocity and mean power frequency (bottom) in concentric elbow flexions. Complete extension of the joint defined as 0 degree. Arrows show the direction of movement. Asterisks show significant results by ANOVA and Tukey’s multiple comparison.

Validity of Inverse Analysis of Surface EMG Verified Through the Rat Experiment

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INTRODUCTION

Inverse analysis is one of the methods for detecting information about the location and intensity of active motor units from surface EMG. In order to execute the inverse analysis, we adopted the image method as forward analysis and examined its applicability by comparing with the numerical results of finite element method[4]. Furthermore, we applied the inverse analysis with image method to human surface EMG[3]. The validity of the inverse analysis, however, is only shown by a numerical experiment. In this study, we verified the validity of the inverse analysis of surface EMG by comparing the numerical and histochemical results.

METHODS

Surface EMG measurement and signal processing

The lumbosacral spinal cords of Wistar rats were exposed by a laminectomy and motor units were activated by stimulating single m. gastrocnemius axons in fine dissected ventral root filaments. The stimulus frequency was 1 Hz and the duration was 1 ms. The criteria for a single unit activity was an all-or-none mechanical twitch.

The induced EMG was recorded monopolarly on the surface of left m. gastrocnemius by using 0.7 mm square electrode. The surface EMG signal was amplified over frequency band of 53-1OOOHz and converted to digital signal with speed of 10 kHz, accuracy of 12 bits and length of 100 ms. In order to estimate the location and intensity of the current source, differences between the sampled signals and calculated potentials were minimized by quasi Newton's method.

Histochemical processing

After the measurement of surface EMG, the glycogen in muscle fibers belonging to single active motor unit was depleted by repeated tetanization. The muscle was quickly excised and frozen in isopentane cooled liquid N,. After storage in a cryostat, the muscle was cut at 10 micro meter thickness and stained using the periodic acid-Schiff (PAS) method. The PAS-negative fibers were determined as active muscle fibers[2].

RESULTS

Figure 1 shows the measured and estimated surface myopotentials plotted on circumferential-time plane. In this case, the inverse analysis was carried out with two current dipoles. The error in estimation was 26.5%.

Figure 2 shows the location of current source estimated by the inverse analysis and the distribution of muscle fibers within active motor unit determined by PAS method. Arrows show recording positions of monopolar electrode. Filled circles show the location of current source estimated by the inverse analysis. The active muscle fibers determined by the PAS method were distributed in medial head of m. gastrocnemius and the number of muscle fibers was 16 1. The depth of two current sources estimated by the inverse analysis was 5.7 mm and 5.3 mm. Their intensity was 7.9 nAm and 11 .O nAm.

DISCUSSION

In this study, the number of muscle fibers within a single motor unit distrubuted in the muscle cross section were 70 to 180. It is difficult to estimate the location of indivudual muscle fibers or the extent of distribution, because the solution of the inverse analysis was unstable and not unique. Therefore, we estimated the location of current source representing the distribution of muscle fibers. In the case of Figure 2, we estimated the distribution of 161 muscle fibers with 2 current dipoles. The location of current dipoles estimated was nearer to the recording surface than the center of distribution, because muscle fibers located near to the skin surface affect surface potentials to a large extent.

The intensity of a current source estimated by the inverse analysis is considered the sum of intensity of 30 to 80 muscle fibers, because we represented the about 70 to 180 muscle fibers with 2 or 3 current sources. In the case of Figure 2, because number of active muscle fibers was 161, the current intensity of single muscle fiber is 0.12 nAm on average. This value is close to that Andreassen and Rosenfalck used in their simulational study[l]. It is difficult to verify acculate value of the intensity of current source without the value of conductivity of muscle tissues or membrane resistance of muscle fibers. In order to verify the intensity of current source estimated by inverse analysis, we need to measure the value with superconducting quantum interference devices (SQUID).

From these results, we verified validity of this method for estimating the location of current source within the muscle.

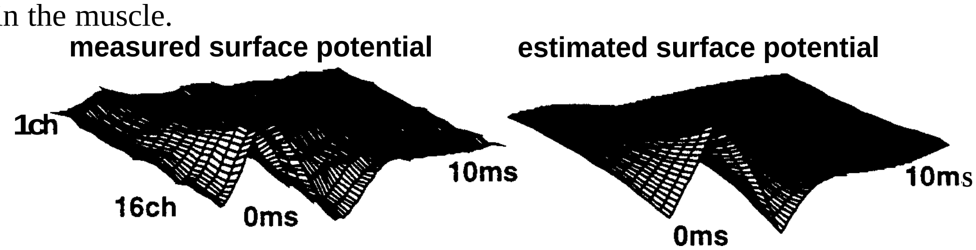

Fig. 1 Measured (left) and estimated (right) surface potentials plotted on circumference-time plane.

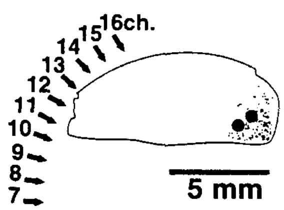

Fig.2 Location of current source estimated by the inverse analysis and distribution of muscle fibers determined by the PAS method.

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The Effect on EMG Patterns During Privoting of Surgical Procedures for Correcting ACL Deficiency

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INTRODUCTION

Knee Stability relies upon muscular action as well as ligamentous stability. The anterior cruciate ligament (ACL) is a major supporting structure in the tibial-femoral joint and surgical procedures are used to improve the mechanical stability of the joint when the ACL is ruptured through injury. The effect of two procedures, an extra-articular (EX) repair and a combined intra-articular and extra-articular repair (IN), on the activity of the muscles around the knee have been investigated.

MEASUREMENT METHODOLOGY AND ANALYSIS

Twenty-six uninjured subjects, ten ACL deficient subjects with IN repair and eight ACL deficient subjects with EX repair were studied during a 90-degree pivoting maneuver. Foot contact signals and electromyographic (EMG) signals of six major muscles have been measured and the linear envelopes (LE) of the EMGs of in each muscle of each subject were calculated. The muscles studied were; rectus femoris (RF), vastus lateralis (VL), vastus medialis (VM), semitendinosis (ST), biceps femoris (BF), and medial gastrocnemius (GS). The magnitude of each LE was normalized by its average over the stride. The stride was divided into 5% intervals. The centers of location of the normalized amplitudes of the LEs at each of these intervals were compared between the normal and the surgical populations using a nonparametric statistical technique called the rank-sum test . This technique was necessary because the distribution of EMG LEs in each population was not Gaussian.

RESULTS

The normal population demonstrated the following EMG profiles:

Rectus Femoris – almost constant activity throughout the stride with peaking during early stance and early swing;
Vastus Medialis – activity throughout stance with the main phase from mid-swing thought early stance;
Vastus Lateralis - similar to vastus medialis;
Biceps Femoris – constant activity from mid-swing through stance;
Semitendinosis – major peak during late swing and activity decreasing through end of stance;
Gastrocnemius – active throughout stance and activity is biphasic .

In Figure 1 are shown two figures; the EMG profiles of RF for the normal and IN populations, and the difference measure for each 5% of the stride. A difference measure greater than 1.96 indicates that the profiles of the two populations are significantly different. The time periods for the difference are; 5% to 15%, 35% to 45%, and 50% to 65%. A summary of the significant difference in EMG patterns between the normal population and the two surgical populations are listed in Table 1.

DISCUSSION

The most remarkable difference is that the EX procedure causes most of the EMG changes during swing phase. Thus the adaptation is concerned with rotating and swinging the leg. Even the enhanced activity during the end of stance phase, 55% to 60%, for the quadriceps group seems to focus on this same task. Only the GS is more active during the weight bearing part of stance, 0% to 10%. The changes in EMG patterns caused by the IN repair induce a lot of adaptations during the stance period. Consider that: RF, VL, BF, and GS are more active during early stance; VM, VL, and BF are less active during midstance. Thus the surgical procedure has changed the mechanics dramatically and has induced a change in synergy. Even during swing there is a complex change in synergy.

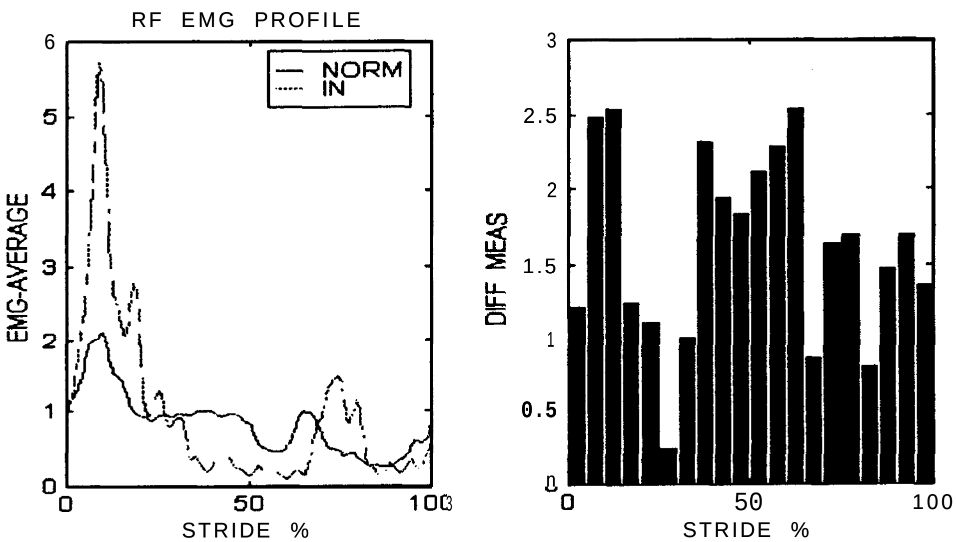

Figure 1.

Table 1 – EMG Pattern Differences		
	IN	EX
RF	more during early stance less during late stance	more during end of stance more during swing
VM	less during midstance more during early swing	more during end of stance more during swing
VL	more during early stance less during midstance less during midswing	more during end of stance more during swing
BF	more during early stance less during midstance more during early swing	more during swing
ST	more during midstance	more during swing
GS	more during early stance more during midswing	more during early stance more during early swing

Influence of Load Uncertainty on Physiological Demand

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INTRODUCTION

In ergonomics, problems related to task uncertainty have rarely been examine in ways that would allow one to quantify their physiologic cost. Belinkii et al (1967) were amongst the first to demonstrate the existence of postural activity preceding and accompanying a voluntary lifting movement of the upper limbs. The role of these anticipatory postural adjustment is to minimise the perturbation of the body's equilibrium associated with intentional movements (Bouisset et al., 1981; 1987). During the handling of unknown loads, uncertainty causes an over-activation of the axial muscles that are responsible for posture (Marchand et al., 1997). However, the duration of this over-activation preceding and accompanying the onset of the unloading phase is very short. Thus, the goal of the present study is to determine if uncertainty increases the muscular demand during a sustained repetitive handling task.

METHODOLOGY

Six subjects participated in the experiment. The subject's task consisted of lifting 3 different loads (2.5, 10.5,18.5 kg) which were randomly varied from one trial to another. The uncertainty was created by using the same boxes for all loads. Also, the load modifications were obscured from the subjects. The subjects performed under 2 conditions involving 192 trials (4 blocks of 48 loads), per condition, done at a rate of 10 per min. In all, each condition lasted 19.2 min. Half of the subjects started with condition 1 in which there was no information about the load to be lifted (i.e., unknown load). The other half of the subjects were informed of the upcoming load 2 seconds prior to the unloading phase (condition 2 - known load). The subjects were given 1 min rest between blocks and 10 min between conditions.

Surface electromyography (EMG) was used to measured the demand for the postural (Erector Spinea - ES, Biceps Femoris-BF, Vastus Lateralis-VL, Gastrocnemius Medialis-GM, Tibialis Anterior-TA) and for load handling (Biceps Brachii-BB) muscles. The skin overlying the muscle of interest was cleaned with alcohol swabs before the placement of the electrode. The electrodes were placed on the middle portion of the muscle and followed the fibres orientation. The raw EMG signals were preamplified (35X) at the skin level followed by further amplification (Therapeutics Unlimited Inc., model 544), fully rectified, filtered (RMS windows of 55 ms) and digitized at 100 Hz. The EMG signals were normalized as a percentage of the mean of 3 maximal isometric voluntary contractions. This normalisation was used to indicate the percentage of muscle utilisation (PMU). Heart rate (HR) was monitored throughout the experiment with a portable system (PE3000).

RESULTS

Figure 1 illustrates the effect of uncertainty on the muscles studied. It shows that uncertainty increases the muscular demand, particularly for the muscles involved in postural control. The PMU was significantly increased (0.13%) in the VL muscle, $F(1,5) = 35.1$, $p < 0.05$. This increased VL activity coincided with an increase (0.18%) in the BF EMG, $F(1,5) = 25.74$, $p < 0.05$. For the trunk muscles, uncertainty also caused an increase (0.28%) in the ES PMU, $F(1,5) = 67.62$, $p < 0.05$. In the case of load handling muscle (BB), the uncertainty resulted in a decreased (0.24%) PMU. $F(1,5) = 49.66$, $p < 0.05$. The statistical analysis revealed no significant differences for GM and TA. The analysis of the HR changes indicated that unpredictable situations increase (0.88) the global energy demand $F(1,5) = 6.9$, $p < 0.05$.

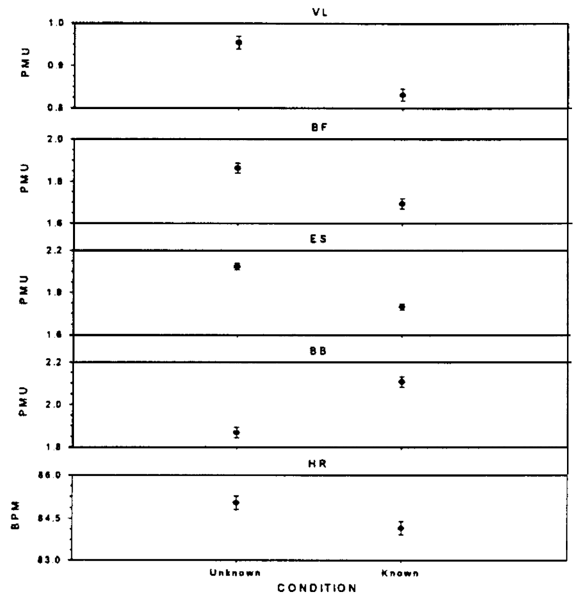

Figure 1: Mean (±95% confidence intervals)for percentage of muscle utilisation (PMU) and heart rate (HR) for known and unknown conditions

DISCUSSION

The present results support the idea that uncertainty increases the muscular demand, particularly for muscles involved in postural control. As a consequence, the over-activation of these muscles also leads to an increase of the mean HR. Moreover. these results show that uncertainty does not only produce a brief increase in the demand of the postural muscles (Marchand et al., 1997) but, it has a more significant physiologic cost. When this extra demand is extrapolated from a short experimental period, such as in the present study, to a full day's work, the physiologic cost would become even more significant. The increased muscular demand suggests that load handlers are incapable of adequate anticipatory planning of muscles responsible for axial stability. This is the direct result of a lack of information about the intrinsic characteristics of the handled load.

ACKNOWLEDGEMENTS

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Open-Loop Tracking Performance with External Control of Antagonist Co-Contraction and Recruitment

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INTRODUCTION:

One approach to restoring fine control of extremities paralyzed by spinal cord injury or stroke has been to emulate the processes of normal motor control. Motor unit recruitment, firing rate and antagonist Coactivation have been extensively studied in the past, and have been emulated using electrical stimulation techniques developed in our laboratory. The purpose of this study is to explore the ability of an externally controlled joint to track various input signals in isometric and load-moving conditions without the benefit of external feedback.

METHODS:

Six adult cats were anaesthetized with a-chloralose. Their sciatic nerves were exposed and all hind limb muscles with the exception of the soleus and tibialis anterior were denervated; tripolar cuff electrodes were wrapped around the common peroneal and tibia1 nerves to deliver stimulation. The limb was fixed in position by a pelvic clamp and a femoral pin. The ankle was aligned and secured to a rotating armature, which was locked in equilibrium for isometric conditions, or loaded with a 250 g weight at 11.5 cm to give a pendulum load. Joint torque and angle were measured through strain gages and a potentiometer in the armature shaft. Stimulation to each limb was generated by a system capable of recruiting motor units by order of size via high frequency blocking. Co-contraction was set by a bi-linear function designed to provide a small degree of co-contraction at low net torques, and antagonistic stabilization at high net torques. Input signals consisted of sinusoidal inputs at 0.5, 1 1.5,and 2 Hz, staircase functions with 3,5 and 7 steps and three band-limited (2 Hz) pseudo-random functions. Tracking quality was evaluated by cross-correlation for sinusoidal and random signals, and by rise time and RMS error for the staircase signals. ANOVA was used for the statistical analysis of the data..

RESULTS:

Sample inputs and outputs are shown in Figure 1. For random signals, two-way ANOVA of peak cross correlation showed a significant effect of signal sequence and ($p < 0.005$) and of load ($p = 0.0105$). The torque in isometric conditions has relatively stable peak cross-correlation near 0.86. The pendulum angle and torque parameters had mean cross-correlations (0.67 and 0.74 respectively) lower than isometric torque. The time to peak cross-correlation had significant effects of load (($p < 0.0001$) and signal sequence ($P < 0.0001$) with significant interaction ($p = 0.0021$). In all three parameters, test signal 2 gave longer time delays (280 to 481 ms) than test signals 1 and 3 (232 to 361 ms). Also, in each case the isometric torque had shorter delays (232 to 280 ms) than the pendulum parameters (309 to 372 ms for torque, 345 to 481 ms for angle). For sine waves, peak cross-correlation deteriorates with increasing frequency, more rapidly in the pendulum parameters than for the isometric torque. Statistical analysis shows significant effects of load ($p < 0.0001$) and of frequency ($p < 0.0001$) with significant interaction ($p = 0.0029$). The mean maximum cross-correlation for isometric torque and pendulum angle decrease gradually with increasing frequency from 0.94 and 0.82 at 0.5 Hz to 0.78 and 0.39 at 2 Hz, respectively. The pendulum torque has its lowest cross-correlation at 1.5 Hz (0.29). The time delay analysis reveals that for the isometric torque and the pendulum angle, the time delay gradually decreases with increasing frequency from 244 ms and 399 ms at 0.5 Hz to 141 ms and 308 ms at 2 Hz, respectively.. The pendulum torque, however had a minimum at

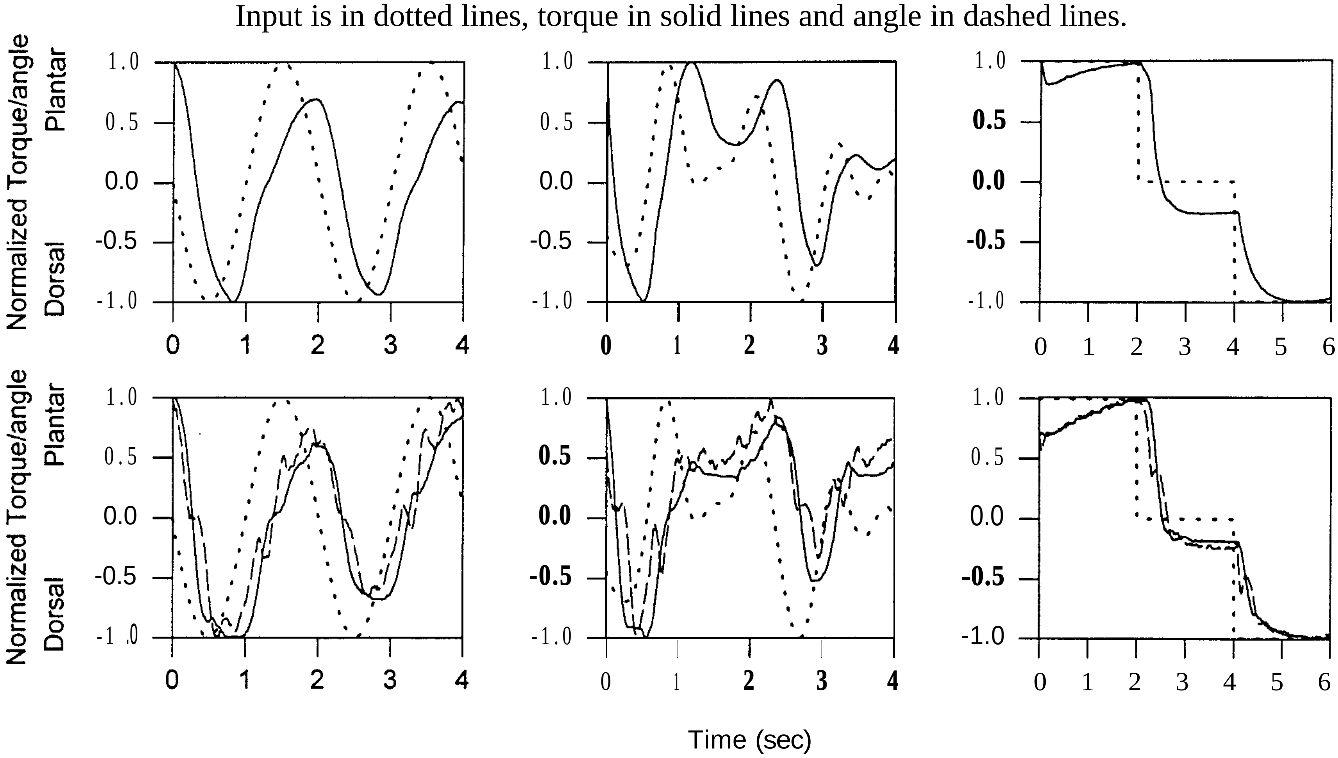
1.5 Hz. Significant effects of load ($p < 0.0001$) and frequency ($p < 0.0001$) were found as well as significant interaction ($p = 0.0268$).

In step signals, rise time had a significant effect of both load ($p < 0.04$) and step ($p < 0.00$) with no interaction. In general, the isometric torque had the faster rise times (between 209 and 535 ms); the pendulum torque had rise times in the range of 395 to 103 1 ms, and the pendulum angle between 4 14 and 923 ms. The staircase signals with the fewer number of steps (larger step size) tended to have smaller rise times. With regard to the steady state error, significant effects of step were revealed ($p < 0.0001$) with no significant effect of load or interaction. All parameters had maximal error in middle of the command range, improving as the command signal approached the extremes of movement. In the seven step signals, error in the middle of the command range was between 35 and 40%; in the five step inputs, the maximum error was in the vicinity of 25%, and in the three step staircases, maximum errors were near 20%.

DISCUSSION AND CONCLUSIONS:

Even when addressing the fundamental issues of motor unit recruitment, firing rate and co-contraction, open loop control of the joint is poor, with relatively large tracking errors and poor response times. Applications needing fine control of joint torque and angle require the use of feedback. Ideally, two levels of feedback would be used; kinesthetic to monitor the resultant status of the force, and proprioceptive to monitor the individual performance of the muscles involved in the movement. In addition, when the joint is required to move, the load dynamics have a profound effect on the overall performance of the joint control system.

Figurel: Sample isometric (top row) and pendulum (bottom row) signals.
Input is in dotted lines, torque in solid lines and angle in dashed lines.



Fine Motor Control of Abductor Pollicis Longus Muscle and Timing of Writing in Healthy Active Subjects of Different Ages. An EMG Evaluative Study

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INTRODUCTION

Electromyography of normal fine sensorimotor ability of a useful action may be an important approach to interpret the level of inability usually seen in motoneuron diseases. This study aims to demonstrate different levels of Fine Motor Control (FMC) ability of the Abductor Pollicis Longus (APL) muscle, (an important muscle form moving and stabilizing the human thumb [1,2]) achieved in healthy active subjects of different ages, giving further additions to results obtained in the only young adult group [2] and to classical EMG evaluations of FMC [4]. The quality of this isolation (FMC with the aid of EMG biofeedback) and its variation were taken into account in order to find out to what extend they could be affected by the level of tests difficulty, types of contraction, speed regularity, or by the age of the subjects.

METHODS

One group of young adults aged from 20 to 40 years old (n=13) and another of active senior citizens aged from 63 to 75 years old (n=14) participated. They were stabilized in a semi-reclined position at a multi-Orthotic system to facilitate the suppression of body muscular activity and to allow only the right thumb-index voluntary motions. With indwelling fine wire electrodes, connected to an EMG apparatus, myoelectrical responses of the APL muscle was analysed. The FMC of APL was tested while the thumb and index, holding a pencil at an upward writing posture (Sl), were free to perform a self writing movement. An electrogoniometric (EGM) device was used to record, simultaneously, the angular displacements during carpometacarpal abduction-extension and adduction-flexion of the thumb, required when slowly writing a line downward and upward (SD, SU) and rapidly writing a line downward and upward (FD, FU). Observations led to establish percentage-level of FMC achievements. So, the interpreted levels of FMC were based on: A perfect isolation of a chosen MU (100%) called "Very Good'; an inhibition of activity, a predominance of the chosen MU, or transfer of MU activity (80%) called "Good'; and an evoked light activity (60%) called "Fair". For the obtained data, the χ^2 statistic test was applied and, a probability of $p < 0.05$ was used as the significance level.

RESULTS

Table 1. Frequency distribution of levels of APL muscle FMC in young and elderly subjects

	Young (n = 13)			Elderly (n = 14)			
Movements	Very Good	Good	Fair	Very Good	Good	Fair	$\chi^2_{(2)}$ P=
SI	76.9	23.1	.0	57.1	35.7	7.1	1.688 p = 0.43, NS
SD	30.8	38.5	30.8	28.6	14.3	57.1	2.586 p = 0.27, NS
SU	38.5	46.2	15.4	28.6'	35.7	35.7	1.453 p = 0.48, NS
FD	7.7	30.8	61.5	14.3	28.6	57.1	0.297 p = 0.86, NS
FU	15.4	38.5	46.2	21.4	21.4	57.1	0.950 p = 0.62, NS

Each test has its own level of success and degree of difficulty and, the perfect FMC in APL could be achieved in any of the tests given. The frequency distribution of levels of FMC of these groups (Table l), being, for example in Sl, Very Good for Young = 76.9 and for Elderly = 57.1, demonstrate the feasible degree of the testing procedure and the similarities of FMC levels of achievements in both groups during other movements. Almost all subjects succeeded to control inhibition at will in APL during a maintained writing posture (Sl). Analyses of the frequency distribution of subjects related to levels of success of each static and dynamic test, showed increase FMC difficulty in fast continuous motions. Subjects's APL FMC ability during voluntary writing motion without assistance had a nonsignificant tendency to be better during eccentric (FU and SU) than in concentric (FD and SD) contraction of the APL muscle. In rapid motion under FMC, was observed a nonsignificant lower level of FMC achievement performed by both young and active senior citizens, pointing out, when comparing these group, a sign of the locomotor system maturity.

DISCUSSION

The level of FMC success of the APL being similar to the one previously obtained among other muscles of the hand [2], the leg [4] and the arm [3], allows the assertion that the summit territory of this muscle can very well act to perform delicate tasks, a matter of re-educative importance. The lowest detectable level of FMC success obtained during the slow and fast writing downward motions can be due to the effect of a concentric type of contraction. The prime mover action of a muscle demands usually the combined action of several motor units. Therefore, to single out a MU during this type of contraction, the subject has to be able to perform the movement with other parts of the examined muscle or with one or more muscle(s) having a similar function [4]. This theory, although based on slow-assisted motions, can be applied to the active concentric voluntary contraction executed without assistance. The discrete highest FMC level of the APL that was demonstrated during the upward writing motion (applied eccentric contraction) confirms earlier work done with arm muscles [3]. During the gravity-eliminated eccentric motion, subjects actively stretched their muscles under less energy and tension requirements, therefore, less MU activity is necessary. The influence of the speed on the FMC achievement levels, clearly seen in this study. In order to suppress the increment of the neuromuscular effect of speed, subjects should have a very high motor ability level to hold the maintenance of a single MU in activity. The lowest performance demonstrated in fast writing motions for both groups corroborates these theories.

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Muscles Within Muscles, Inter- and Intra-Muscle Segment Coordination

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INTRODUCTION

In articulations capable of multiple movement planes, such as the shoulder joint, the CNS demands extensive control over the characteristics of muscle activation in order to produce accurate voluntary movement. Previous research has shown that muscles surrounding a multi-planar joint, are selectively activated dependent upon the movement performed (Scheving & Pauly, 1959, and Shevlin, Lehmann & Lucci, 1969) and that there exists within the radiate musculature of the shoulder individual muscular segments capable of exhibiting specific myoelectric intensity characteristics (Pare, Stern & Schwartz, 198 1, Soderberg & Dostal, 1978, Paton and Brown, 1994) There exists, however, a lack of research quantifying the ability of the CNS to control the temporal activation characteristics of each intra-muscular segment and to coordinate inter-muscular segment activation. This study sought firstly to investigate the patterns of segmental muscle activation, seen both within a single muscle and across groups of muscles. This was achieved through a simultaneous investigation of three superficial shoulder muscles; the pectoralis major, the deltoid and the latissimus dorsi during the production of an isometric shoulder adduction task. Secondly, the study investigated the effect of varying the velocity of the motor task on muscle segment coordination.

METHODS

The myoelectric patterns of fourteen muscle segments from the pectoralis major, the deltoid and the latissimus dorsi were recorded during the production of an isometric shoulder adduction task. The shoulder was positioned at 90 degrees of pure abduction with the elbow encased in a brace which both eliminated unwanted movement and allowed the attachment of a load cell which recorded the adduction impulse force. Twelve male subjects (age 24.25yrs ± 3.67, height 177.68cm ± 6.32, weight 73.78kg ± 6.03) performed 10 isometric impulse shoulder adductions which consisted of a progressive development of isometric force up to the subjects 50% MVC level over a set time frame. The subjects then decreased the force level back to 0% MVC over the same time frame. Five different movement times (300, 500, 700, 1000, and 1500ms to peak force) were utilised. Subjects w-ere required to produce an adduction impulse that matched a criterion movement trace (within ± 10% error). Fourteen microelectrodes, with both active plates housed within the one base and an interelectrode distance of 6mm, w-ere designed for this experiment to minimise electrode pickup area and to reduce crosstalk. Five microelectrodes were placed on both the pectoralis major and latissimus dorsi with 2 microelectrodes each placed over the anterior and posterior segments of the deltoid. The myoelectric patterns were analysed to reveal the temporal characteristics of each muscle segment.

RESULTS

Results have indicated that the CNS has the ability to independently control the activation of muscle segments within each of the investigated muscles, through variation of muscle onset and duration but not time to peak muscle activation (figure 1). Unlike all other muscle segments, the inferior segments of pectoralis major (Pl) and latissimus dorsi (L5) were found to be activated prior to force onset regardless of movement speed (figure 2a & 2b). Both segments were also generally shown to have the longest duration of activation (Figure 3a & 3b). In contrast, more superior segments were activated later and generally had shorter periods of activation.

Interestingly, peak muscle activation occurred simultaneously regardless of the segments anatomical location. This result indicated that all muscle segments were coordinated to peak at the same time in order to produce the 50% MVC force regardless of movement speed.

DISCUSSION

This experiment has shown that the activation of segments within individual muscles are temporally coordinated to produce skilled motor performance. Furthermore, by investigating the activities of three shoulder joint muscles involved in the production of an isometric shoulder adduction force, it was also possible to show that muscle segment activations are highly coordinated across groups of muscles.

By analysing the temporal characteristics of muscle segment activation it was also possible to show that individual muscle segments may assume functional roles which could be termed prime mover, synergist and antagonist. Prime mover segments were identified as those w-hose activation preceded force onset. Generally, these segments were those with the most favourable line of action for the task. Synergist muscle segments were activated after force onset and possessed lines of action which deviated from the desired intended direction of the force impulse. Antagonist muscle segments were characterised by generally later activation and lines of action opposite to the intended motor task. These generalisations were valid regardless of movement speed.

Of interest was the finding that the time of peak muscle activation of all muscle segments, regardless of their function and movement speed, coincided with the achievement of peak force (5O%MVC). This result indicated that for this particular motor task, the time of peak muscle activation was an invariate motor control parameter; changes in movement speed being accommodated through variations in segment onset and duration

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Multi-Electrode Recordings from the Arm: Prelimary Findings

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INTRODUCTION

Since the first paper of Monster et al. (4) reporting that MUs could be located in the volume conductor with an array of surface electrodes, additional information on the muscular physiology has been obtained with this approach. For instance Masuda and al. (3) used an array of 17 small and closely interspersed electrodes applied on the skin and recorded 16 EMG signals from which they could investigate the scattering of the innervation zone of individual MUs. Previously, such information could only be obtained with biopsies and histochemistry. With 40 electrodes and spatial filtering, Reucher et al. (6-7), Rau et al. (5) have demonstrated that, even at high levels of muscle contractions, single MU impulses can be detected in the surface EMG signal; usually, this is achievable only through subcutaneous electrodes. Cram et al. (2) experimented with topographic maps of approximated Laplacean transformed EMG's: with 29 surface electrodes they obtained maps from which location of active muscles and their degree of activity could be assessed. For prosthesis' control, Trio10 and Moskowitz (8) have demonstrated that limb function detection and muscle force estimation is more natural when 4 signals are used instead of one.

METHODS

Simultaneously recorded EMG signals where obtained by using equipment developed in our laboratory (1). We used modified commercially available electrodes by removing the jelly and the self-adhesive collar leaving only a 8 mm Ag-AgCl pellet. The electrodes were fixed on an elastic band measuring 27 x 8 cm at the extremities of which velcro strips served for fixation. The electrodes were placed in a matrix of 8 columns (A to H in Fig.1) and 3 rows. The center-to-center distance was 2.5 cm for the columns and 2.0 cm for the rows. In the absence of extension of the elastic band, the electrodes columns extended for 17.5 cm leaving a length of 9.5 cm which was used to place 2 connectors matching the connection cable from the amplifiers and two velcro strips for securing the electrode array around the arm. The elastic band was placed at mid-section of the biceps with columns of electrodes aligned parallel to the estimated direction of the muscle fibers. A reference electrode was placed on the forearm some 20 cm from the center row of electrodes and a ground electrode was placed near the wrist. Data acquisition was done on the right arm of 6 normal subjects. Torque and EMG signals were sampled at 1 kHz. After two elbow contractions at 100% of maximum voluntary contraction (MVC), five isometric contractions of 5 s were done at 10, 25 and 50% MVC. For each subject, the first data acquisition was followed by a second one the day after at approximately the same hour. Ink markers put on the skin during the first session were used to re-install the electrodes within few mm of their initial position. The force was measured with a strain gauge whose output served as a feedback signal to the subject.

RESULTS

At the mid-portion of the biceps, the circumference of the arms of the 6 subjects measured from 24 to 34 cm at rest and were estimated to range from 30.5 to 37.5 cm during a 100% MVC (5 subjects). As illustrated in Fig.1 (the electrodes columns are lettered A to H), the electrodes covered approximately two-third of the circumference of the arm. The inter-column distance thus varied from subject to subject and changed with the level of the contraction. From the 24 monopolar recordings, single-differential (sd) and double-differential (dd) signals where obtained off-line. From the 5 s signal samples shown in Fig.1 (same scale for all), activity was most intense over the biceps (electrodes B,C,D), was moderate in E, small in A and nil in H. In spite of quality control measurements made for each electrode at the beginning of the experiment, noise was observed in F and G. Similar results were obtained from experiment to experiment and between subjects. For the experimented levels of contraction, coefficients of correlation (cc) between individual signals within a column (one row to the next), from one column to the next (same row) or in diagonal (adjacent'columns but different rows) was above 0.9 as shown in the upper section of Table 1. For sd signals, the correlation was in the vicinity of 0.36 from row to row and above 0.44 from one column to the next (middle section of the table). The dd signals could only be obtained from column to column and they were found to be smaller (bottom section of Table 1) than those obtained for sd signals.

DISCUSSION

For adjacent individual signals, high cc were found indicating that the electrode array offered a spatial sampling which was higher than necessary. Since the columns of electrodes were aligned parallel to the fibers direction, cc obtained from row to row were expected to be higher than those obtained from two adjacent columns either within the same row or in diagonal. Higher values were not found indicating that the columns of electrodes were probably not exactly parallel to the muscles fibers. It is also possible that the tension applied to the elastic band could have induce distortion in the electrodes alignment. Whatever the alignment, it appeared to remain constant from 10 to 100% MVC. For sd signals, lowest cc were obtained from differential signals obtained in diagonal from different columns. This could be attributed also to the misalignment of the electrodes. When column to column results are compared, the slightly lower values for the dd versus the sd signals would indicate the presence of some propagated signals.

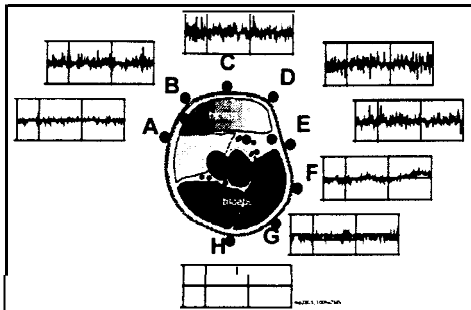

Figure 1: illustration of the arm muscles surrounded by 8 columns of electrodes (A to H) and a sample of double-differential signal obtained from each column during a 100% MVC isometric contraction of the elbow.

% MVC	row to row	column to column	diagonal
individual signals			
10	0.92±0.03	0.92±0.03	0.90±0.03
25	0.92±0.02	0.92±0.02	0.90±0.02
50	0.92±0.03	0.91±0.03	0.88±0.04
100	0.91±0.04	0.90±0.04	0.86±0.05
single-differential signals			
10	0.35±0.12	0.44±0.15	0.23±0.09
25	0.36±0.13	0.472±0.12	0.26±0.08
50	0.34±0.11	0.49±0.13	0.23±0.08
100	0.38±0.12	0.50±0.15	0.27±0.08
double-differential signals			
10		0.35±0.11	
25		0.38±0.15	
50		0.40±0.14	
100		0.42±0 17	

Table 1: mean values (± standard deviation) of the coefficients of correlation obtained from 12 recording sessions.

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Seasonal Variations in Neck and Shoulder Muscle Load Among Female Bus Drivers

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INTRODUCTION

Bus drivers are exposed to whole body vibration and often to cold drafts on neck and shoulders when passenyers ascend or descend the bus. Furthermore bad weather conditions can be a stressful factor. In previous studies drafts and cold have been found to increase the activation of muscles and perceived discomfort (1,2). Bus drivers in countries with yreat seasonal variations, such as the Nordic countries, often perceive that driving in wintertime increases the muscle load and induces pain in the shoulder-neck area. The aim of this study was to compare trapezius muscle activity duriny bus driving in fall and in winter.

MATERIALS AND METHODS

Ten female bus drivers participated in the study after informed consent. Their mean aye was 36 years and their experience from professional bus driving was 11 years. Several of them reported neck and shoulder pain in relation to driving. The measurements were made on the same bus route at the same time of the day once during fall and once during winter. Muscle activity in the descending part of the trapezius muscle vas measured by means of surface electromyography (SEMG) during driving. A portable EMG system, Myoguard (3) was used with two hours uninterrupted ambulatory monitoriny of bipolar SEMG from the right and the left side of the trapezius muscle. The recorded EMG siynal was processed online and an average rectified value (ARV) and a mean power frequency (MPF) estimate were stored on a memory card once every second. The ARV and MPF values are stored in blocks on a memory card. Each block contains 30 values and the content of each block can be controlled by a checksum, which can be used to evaluate the quality of the data. In a MS-Windows integrated software an overview of all trials performed during the normalisation and driving was presented and analyzed. The mean % rest time was also calculated.

RESULTS

Mean ARV for the riyht trapezius for driving duriny fall was 26% and duriny winter 23%. The corresponding percentaye for the left trapezius was 27% and 32%. The rest time was approximately the same duriny fall and winter driviny. In fiyure 1 and 2 the mean ARV and % rest time can be seen duriny driviny.

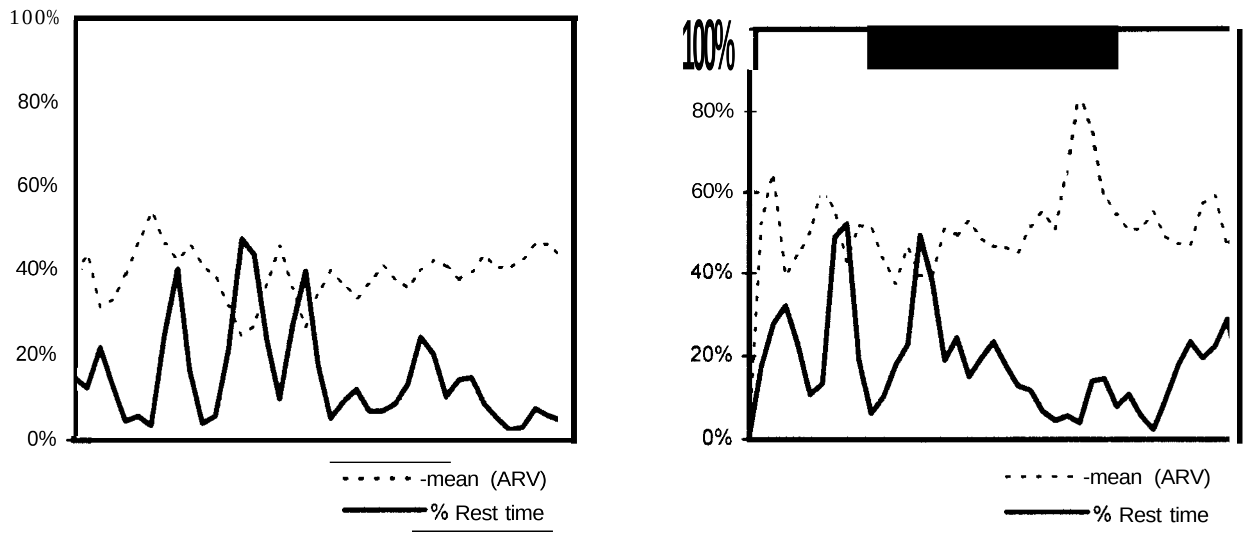

Figure 1 and 2. Mean ARV and % rest time values from the left trapezius duriny fall (left) and winter (right) driviny, divided in six minute intervals. ARV data are expressed in % of RVE and the rest periods expressed as percentage of muscle rest time based on an estimation of the muscle rest level measured before the driviny.

The intraindividual differences of muscle activity were very small when fall and winter driving were compared.

DISCUSSION

Only small differences in muscle activity could be seen between the two seasons. The perceived discomfort during winter driving that drivers often report could possibly be more related to higher stress hormone levels than to increased muscle activity. This study also indicates that the intraindividual differences in muscle activity were small when the exposure was held constant with the same bus route and the same time of the day.

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The Role of Electromyography in the Diagnosis of Post-Polio Syndrome

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Post-Polio Syndrome occurs in many survivors of acute poliomyelitis many years after recovery. The percentage of polio survivors who will experience symptoms of progressive fatigue and weakness attributable to post-polio syndrome has been variably estimated from 25 to 60%. The diagnosis of post-polio syndrome is a clinical diagnosis, and as to date, no objective test is available that is specific for this condition. This diagnosis is made only after other potential pathologies are first reasonably considered and excluded.

After the acute stage of paralytic poliomyelitis resolves, a variable degree of recovery occurs one to two years afterwards, depending on the extensiveness of anterior horn cell destruction by the virus. This recovery process is due in large part to collateralization by the remaining viable neuron. In some cases, the resulting motor units can innervate up to five times the area of muscle fibers than that supplied by normal motor units. These motor units manifest on conventional electromyography as large polyphasic potentials.

The EMG findings in polio survivors is in many respects typical in disorders of chronic motor polyneumopathy. In the conventional clinical electromyogram, large polyphasic potentials are observed, and depending on the degree of chronic denervation, early recruitment and incomplete interference patterns at full effort may be observed. Some reports have also noted occasional widely scattered abnormal potentials of fibrillations and positive sharp waves, in some cases.

Single fiber EMG studies have shown increased jitter and increased blocking in both post-polio survivors who have been diagnosed with post-polio syndrome, as well as in those who have not yet experienced new symptoms of fatigue and weakness. It is felt that the symptoms of new weakness in PPS is the result of motor unit dysfunction at the axonal branch points. These findings suggest that the surviving motor neurons are undergoing a process of ongoing denervation and reinervation, and that the PPS symptoms are related ultimately to the attrition of terminal axons of the oversprouting neurons. After a period of time, perhaps due to chronic excessive metabolic demands, these motor neurons are unable to support all of their axonal sprouts, which ultimately results in a failure of reinervation. This hypothesis is also supported by histological studies of muscle tissue of polio survivors. Atrophy of individual muscle fibers was frequently found, but grouped atrophy, indicative of motor neuron attrition, was not present. This finding suggests that the areas of denervation are at the areas of the motor endplate.

Studies to date, however, have not shown any statistical differences in EMG findings in those polio survivors with a diagnosis of PPS and those who are asymptomatic. The EMG, nonetheless, remains an essential diagnostic tool in selected instances to confirm chronic denervation of poliomyelitis where the history may be in question, and also to confirm the presence of neuropathic or myopathic comorbid conditions.

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Hemiparetic Patients Recruit Additional Degrees of Freedom to Compensate Lost Motor Function

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INTRODUCTION

Previous studies have shown **that** in stroke patients, multi-joint pointing movements are characterized by spatial and temporal incoordination between adjacent arm joints (Levin 1996). Pointing and prehension movements involving postural adjustments are further characterized by a lack of coordination between different body segments such as the shoulder and trunk (Roby-Brami et al. 1997). The goal of this study was to identify the ability of the damaged nervous system to compensate disruptions in interjoint control of the a m with the recruitment of additional degrees of freedom not normally recruited during movement.

METHODS

Two groups of subjects participated in this preliminary study. The experimental group consisted of left hemispheric stroke patients. They were right-hand dominant and had sustained a single stroke between 2-12 months previously leading to arm paresis (stage 2-3 control of the arm according to the Chedoke-McMaster Stroke Assessment Scale). They had no major perceptual deficits which may have interfered with their ability to participate in this study. The control group consisted of right-hand dominant, **age-** and sex-matched healthy subjects. Subjects in both groups made repeated (75) pointing movements with the right arm from an initial target located ipsilateral to the body to a final target located in the contralateral workspace, in front of the subjects. Subjects had visual feedback of the initial position in all trials and of the final hand position on every fifth trial. Kinematic data from 6 infrared markers, placed on the right arm and on the trunk were recorded at 200Hz by a 3-D Optotrak motion analysis system.

RESULTS

Data from the affected arms of hemiparetic subjects were compared with those from the arms of healthy subjects. In accordance with previous studies, our results showed that in hemiparetic subjects, movement times were significantly prolonged (1.519 ± 0.262 s compared to 0.934 ± 0.054 s in healthy subjects), tangential velocity of the endpoint was multi- instead of uni-phasic (fig. 1) and interjoint coordination was modified. Elbow flexion/extension movement was more tightly coupled with shoulder horizontal abduction/adduction (r= -0.76 ± 0.2 in 69 % of trials) and with shoulder elevation (0.74 ± 0.1 in 75% of trials) compared to healthy subjects (0.42 ± 0.1 and -0.38 ± 0.1 respectively). **Also,** the movement amplitudes were lower in affected arms. Ranges of active elbow motion were 28.3 ± 5” in hemiparetic compared to 56.8 ± 7.7” in healthy subjects. For horizontal motion of the shoulder, the range was 63.5 ± 13.2” compared to 127.6 ± 3.5” and for vertical motion the range was 12.2 ± 3.8” compared to 34 ± 5” in hemiparetic and healthy subjects respectively. The decrease in active range of the arm joints occurred together with an increase in the participation of the trunk during pointing movements. The mean values of the trunk movement were 83 ± 23 (mm) in hemiparetic compared to 15 ± 4 (mm) in healthy subjects (fig.2). To reach the target, healthy and hemiparetic subjects used the same sequences of arm and trunk movement (trunk began first and finished last). In only one hemiparetic subject was this sequence changed at the end of movement (trunk finished before the arm). The delay between arm and trunk motion during pointing movements was approximately 100 ms longer in hemiparetic subjects.

This difference may be partly explained by the movement times, which were prolonged in hemiparetic subjects.

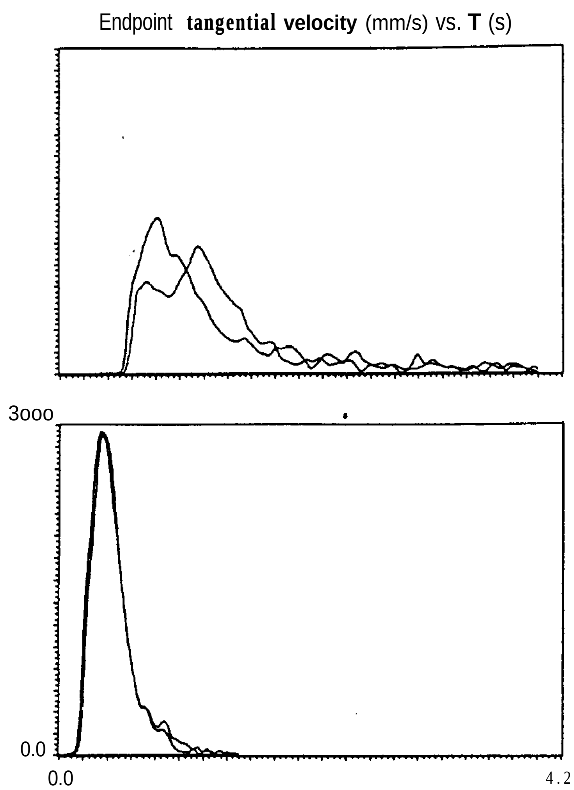

fig. 1

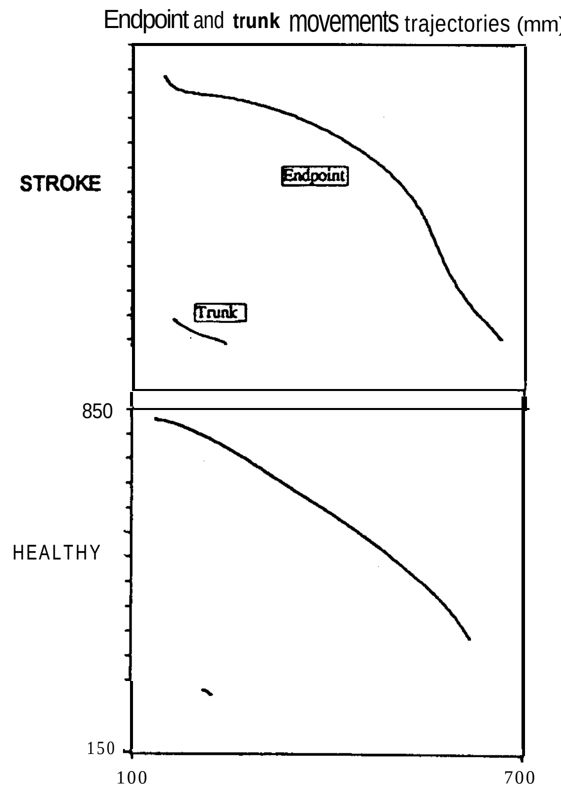

fig.2

DISCUSSION

Clinically, during recovery from stroke, the nervous system may exploit the redundancy of the motor system by substituting a lost element of a motor pattern with a compensatory strategy to achieve the functional goal. Our data shows that hemiparetic patients use less range of motion and an abnormal coupling between joints of the affected arm. To compensate this abnormal pattern of movement the nervous system appears to use a new adaptative coordination represented by more trunk recruitment during multi-joint arm movements. This trunk recruitment does not influence the temporal sequence of segmental recruitment but may represent a compensatory strategy used by hemiparetic patients.

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Fluctuations in Tremor at Rest and Eye Movements During Ocular Fixation in Subjects with Parkinson's Disease

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INTRODUCTION: Transient motor behaviors such as fluctuations in tremor at rest (TR) amplitude over time is a feature of Parkinson's disease (PD). The fact that short-term fluctuations are present in untreated subjects with PD suggests that they are a manifestation of the disease itself It is known that deficits associated with PD affect different neurotransmitters and different neural pathways. We hypothesize that mechanisms causing these transient motor behaviors may cause simultaneous fluctuations in other brain functions such as vision and that these fluctuations may coincide in their timing. To verify if both the stability of the hand (i.e. TR) and the fixating eye were simultaneously affected by PD, recording of TR and horizontal eye movements during ocular fixation was done.

METHODS: Ten subjects participated in this study. Five of those subjects had been diagnosed with idiopathic PD. TR of the hand was recorded using a position laser system. Eye fixations were recorded using an infra-red eye movement recording system (Ober2). Subjects were seated on a chair with the back of their head leaning on supports, they bit on a piece of thin wood (tongue depressor) attached to the structure of the chair, thus fixating their inferior and superior maxilla. subjects' hands hung freely over the edge of an arm support. TR of subjects with PD was recorded on the side of their body most affected by the disease, as for control subjects, recording was done on the side corresponding to the affected side of their respective matched subject in the PD group. While TR was recorded, subjects were asked perform (during 66 s) an ocular fixation on a target (small yellow circle of 3 mm in diameter with a black dot in its center) which was located in the middle of a computer screen.

RESULTS: TR average amplitudes (root mean square) were significantly larger in the group of subjects with PD (2.00 mm ± 2.47 mm) when compared with the control group (0.02 mm ± 0.008 mm, Mann Whitney U=O, $p \leq 0.01$). To compare differences in TR fluctuations between groups, the standard deviation of the linear envelope of TR was divided by the mean of that envelope, creating a ratio of TR fluctuation for each subject. Results showed that ratios of TR fluctuation were higher in the group of subjects with PD (0.06 ± 0.1) when compared to the control group (0.009 ● 0.009, Mann Whitney U=O, $p \leq 0.01$) indicating more fluctuations in TR for the PD group. Fluctuations of eye movement amplitude (EMA) between groups were compared using the same method. Results indicate that overall fluctuations of EMA were similar for both groups (1.63 ± 0.36 for the PD group Vs 1.83 ± 0.21 for the control group, Mann Whitney U=8, $p \geq 0.05$). Possible relationships between fluctuations in TR and fluctuations of EMA during were investigated using hvo simple methods. First, comparison of amplitude of eye movements was done, when TR amplitudes were large and small. The main objective was to detect if major changes in TR amplitude were accompanied by systematic modifications of EMA. An upper and lower threshold was established for each subject using the linear envelope of their respective TR. Once the linear envelope of TR calculated, computation of the upper threshold was done by adding to the mean of that envelope (subtracting for the lower threshold) 0.75 times the standard deviation of that mean. Every time TR amplitude was above the upper threshold, TR was considered as being large, below the threshold it was considered small. Results indicate no significant differences between EMAs during large and small TR amplitude in the PD group (large TR: 1.04 ± 0.12 Vs 1 .O1 ± 0.13 during small TR., Mann Whitney U= 7, $p \geq$ 0,05), meaning that changes in TR amplitude did not coincide with changes in EMA. Secondly, calculation of the rank correlation (Spearman's) behveen each time series (TR amplitude and EMA) was performed: a window of 200 data points (i.e., I s) was moved along the time series, 50 data points at a time. The root mean square of TR and the average amplitude of eye movements were calculated respectively every time the window was moved. No significant correlation was found between TR and EMA in either subjects of either groups. However, at rare times, the correlation was high, particularly in subject 4 from the PD group. Visual inspection of his TR and eye data enabled us to identify one particular instance where his TR amplitude and EMA seemed to be modified simultaneously. Subject 4's right eye seemed to oscillate around the same frequency as his TR. Surprisingly, his left eye did not present such oscillations. A systematic search for similar phenomena in other recordings revealed two other subjects (2 and 5, from the PD group) that presented such monocular oscillatory eye movements. Contrary to subjects 4 and 5, who presented oscillatory eye movements in the right eye, those of subject 2 were more predominant in his left eye. Subjects 4 and 5 were more affected by the disease on their right side, whereas subject 2 was more affected on his left side. To quantify the apparent similarities between the main frequency of TR and the frequency of oscillatory eye movements observed in these particular cases, maximal coherence values were calculated, for each consecutive 5 s available, between TR of the tested hand and each eye. Overall results indicate that the coherence values of data from each eye with TR were markedly different for some subjects in the PD group, mainly subjects 4 and 5 (Fig. 1 A), meaning that both eyes did not present the same level of similarities in frequencies with TR. This was not the case for any subjects in the control group (Fig. 1B). Maximal coherence values were also calculated, for the PD group, during a predetermined 10 s period in which monocular oscillatory eye movements where predominant and square wave jerks were minimally present. For the control group, maximal coherence values were also calculated during a predetermined 10 s period in which square wave jerks were minimally present. The coherence values were higher for all PD subjects (more markedly for subjects 2,4 and 5), between TR of the tested hand and the eye ipsilateral to the side of the body most affected by the disease (Fig. 1C). Coherence values for control subjects were still low even without the presence of square wave jerks (Fig. 1D).

DISCUSSION: Results of the present study indicate that there are no systematic and direct relationships between fluctuations in TR of the hand and fluctuations of EMA during ocular fixation. However, it is believed that the occasional monocular oscillatory eye movements seen in subjects 2, 4 and 5 from the PD group may be the result of the disease itself The possibility that head movement occurred (jaw tremor), resulting in vestibulo-ocular reflex, cannot be excluded. However, if it had been the case, it is suspected that both eyes would have presented similar behaviors since the vestibulo-ocular reflex is believed to be normal in PD [l]. In any event, it is clear that the two eyes behaved differently during visual fixation in contrast to most control subjects. To our knowledge, this is the first time such monocular oscillations have been observed in PD. Similar oscillations, mainly monocular circumduction-related nystagmus, have been observed on rare occasions in patients with multiple sclerosis [2]. Further experimentation will be needed to confirm the present results.

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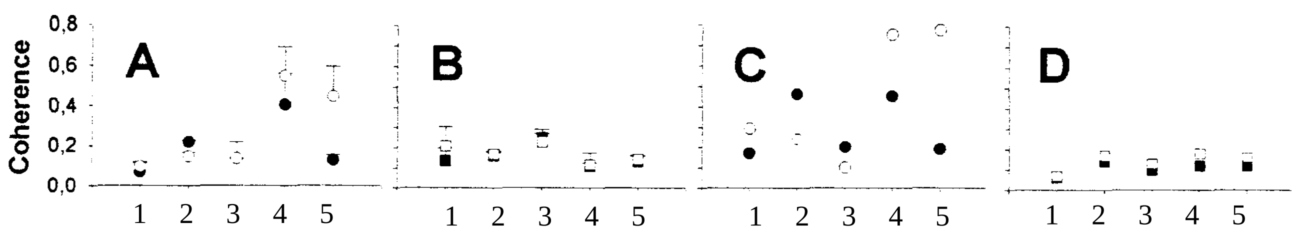

Fig. 1. Coherence values between each eye and TR of the tested hand. Open symbols = right eye, filled symbols = left eye. circles = PD subjects, squares = control subjects. Abscissa = subject's number.

Comparisons Among Simultaneous Recordings of Human Finger Tremor in Displacement, Velocity, Acceleration and Electromyography

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INTRODUCTION: Finger or hand tremor is most commonly evaluated with acceleration measurements and recordings of electromyographic (EMG) signals. Numerical integration of the acceleration time series is then employed if measures in velocity and displacement are required. There exist, however, highly precise means of measuring displacement (reference #l) and velocity directly. The purpose of the study was to compare the time series and power spectra characteristics of simultaneous recordings of displacement, velocity, acceleration and surface EMG data obtained from postural tremor of the index finger in control subjects and in subjects diagnosed with Parkinson's disease (PD).

METHODS: In part I, 8 control subjects (mean age: 38, range: 20-73) and 6 subjects with PD (mean age: 57, range 50-64) participated. The postural tremor of 1 index finger was simultaneously recorded using an accelerometer and 2 laser systems that transduce displacement and velocity, respectively, using different technologies. Surface EMG signals were simultaneously recorded from the extensor digitorum communis (EDC). In part II, 5 control subjects participated (mean age: 29, range: 22-38). The postural tremor of 1 index finger was simultaneously recorded using the same laser systems and a different accelerometer. For recording, each subject was seated with arms supported. The arm from which tremor was recorded was supported throughout the forearm and hand. The wrist was in slight extension, the index's metacarpophalangeal joint in slight flexion and the 2 interphalangeal joints fully extended, resulting in a horizontal finger position. A lightweight card to reflect the laser beams was taped to the accelerometer and the latter was affixed to the dorsum of the index. The subject attained the recording position with visual feedback, and was instructed to maintain it as well as possible for 30 seconds with eyes closed. Two 30-second recordings were obtained from each subject. All signals were sampled at 1000 Hz and recorded directly onto a personal computer. The signals were filtered off-line using 4th-order double-pass Butterworth filters. All signals were filtered with a 2 Hz high-pass filter, then the EMG signals were full-wave rectified. All signals were then filtered with a 50 Hz low-pass filter, and multiplied by calibration constants appropriate to instrument specifications and amplification factors. From the 3 kinematic time series, 6 additional time series were obtained: the first and second derivatives of the displacement signal; the first derivative and the first integral of the velocity signal; and the first and second integrals of the acceleration signal. Power spectra were calculated, and the median and peak frequencies (MF and PF, respectively) were determined as well as coherence function estimates (CFEs) among power spectra from simultaneous time series.

RESULTS: *(A) Comparison of amplitudes in kinematic data:* The standard deviation (SD) of each time series was used as a measure of amplitude. The data from all 14 subjects in Part I were used to calculate correlations among measures of displacement, velocity and acceleration from the laser transducing displacement (LD), the laser transducing velocity (LV) and the accelerometer (Al). Correlations among measures from LD and LV were very high: 0.99,0.99 and 0.98 in displacement, velocity and acceleration series, respectively. (The same order is followed for other correlation coefficients reported below.) Correlations among measures from the lasers with those from the accelerometer were also high: LD with Al, 0.93, 0.96, 0.95; LV with Al, 0.91,0.94, 0.95. One subject with PD had a high-amplitude tremor and these data were omitted for a second analysis in order to compare the instruments using only trials of low-amplitude tremor. The correlations among measures from the 2 lasers remained high: 0.98, 0.98, 0.94. Correlations among measures from the lasers and those from the accelerometer were much lower: LD with Al, 0.19, 0.27, 0.12; LV with Al, 0.20, 0.25, 0.10. Correlations were also calculated for the data from part II in which the other accelerometer (A2) was used and only low-amplitude tremor was observed: LD with LV, 0.99, 0.98, 0.89; LD with A2,0.65,0.95,0.95; LV with A2, 0.72, 0.89, 0.88. (B) ***Comparison of power spectra:*** In part I, the MF and PF values from spectra obtained from LD and LV time series showed high agreement, as defined by being within 1 Hz: in displacement spectra, MF 100% (i.e. 28/28 trials), PF 86%; in velocity spectra, MF 96%, PF 82%; in acceleration spectra, MF 46%, PF 68%. The CFE for spectra from the LD and LV time series was ≥ 0.8 for most frequencies at which there was at least 0.0 1 (mm)’ in displacement spectra and 16 $(mm/s)^2$ in velocity spectra. The CFE from each of LD and LV with EMG spectra was generally < 0.4, was between 0.4 and 0.8 for a few frequencies in most trials and > 0.8 only for the spectral peaks from one subject with PD. MF and PF values from acceleration spectra obtained from the Al and from EMG spectra showed agreement with values from acceleration spectra obtained from the lasers’ time series and with each other only for trials from subjects with PD in which pronounced tremor was evident. The Al acceleration spectra showed CFE values mostly < 0.4 with spectra from LD and LV, values between 0.4 and 0.8 for a few frequencies in most trials, and values > 0.8 only for the spectral peaks from one subject with PD. The Al and EMG spectra showed almost uniformly low coherence. In part II, the finding of good agreement between the LD and LV signals for MF and PF values as well as for CFE was replicated The MF values from the A2 acceleration spectra agreed with those from the LD and LV for 50% of comparisons. The CFE from comparing the A2 spectra to the LD and LV spectra was lower than the CFE between LD and LV, but was higher than was seen for the Al spectra for a comparable amplitude of tremor.

DISCUSSION: The results suggest that the systems transducing displacement and velocity are reliable means of measuring amplitude of tremor in displacement, velocity and acceleration. These systems also appear to give consistent information for measures in the frequency domain through much of the frequency range of interest for human finger tremor, even when its amplitude is very low. However, the lower agreement in MF and PF for acceleration series than for displacement and velocity reflects that high-frequency noise in the twice-differentiated displacement signal affected these summary measures. In light of the high coherence in the range in which most power was found, generally < 20 Hz, a different low-pass filter may improve the agreement without eliminating important information. Neither of the accelerometers was as reliable as the laser systems. A2 was greatly influenced by low-frequency noise, thus compromising the amplitude measures in the twice-integrated displacement signal and reducing the agreement in frequency measures. Al was designed for parkinsonian tremor but uses an older technology and is sufficiently sensitive only for relatively high-amplitude tremor. EMG signals from EDC showed moderate correspondence with kinematic measures at some frequencies in some trials, generally with higher amplitudes of movement, but poor correspondence overall. The poor correspondence may be attributed to the generally low EMG amplitude in this simple task, thus suggesting that for low-amplitude tremor, the laser system measurements permit more precise characterization than EMG signals.

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Effect of Prolonged Bed Rest on Amplitude Modulation of H Reflex in Human Soleus Muscle

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INTRODUCTION and OBJECTIVE

H-reflex is a monosynaptic reflex elicited by electrical stimulation of Ia afferent fiber. The amplitude of the reflex used as the index of the central gain to the monosynaptic stretch reflex is determined by transmission across synapses of Ia afferents and excitable level of a motoneuronal pool.

Recent studies (Neilsen 1993) have shown that a significant change in H reflexes was obtained in response to the level of a habitual physical activities. This change might be reflecting the functional changes in the spinal reflex pathway, produced by an amount of physical activity. The purpose of this study, therefore, was to examine the effect of inactivity of prolonged bed rest on the function of the spinal reflex in human, using the amplitude of soleus H reflex.

METHODS

Experiment I

Five healthy male subjects, aged 21 to 28 years, underwent the continuous bed rest (BR) with tilting of -6 deg head down for 20 days. During the BR period, subjects performed routine exercises with the right leg (trained leg) in the supine position in the morning and the afternoon once in each. Two knee extension exercises were conducted in combination of 10 maximal isokinetic contractions with angler velocity of 60 deg/s and 50 contractions with 180 deg/s. The left leg (control leg) was kept inactive throughout the BR period. During the measurement, the subjects seated in a reclining chair at the position of knee flexed to 120 deg and the ankle to 110 deg. The soleus H reflex was evoked by stimulating the tibia1 nerve in the popliteal fossa with constant voltages and duration of 1 msec. Recruitment curves of the H-reflex and M-wave were obtained from all subjects. The stimulus intensity of 1.05 x MT was used as the threshold intensity for the motor M wave. The H-reflex activity was expressed as a percentage to the maximal amplitude of the M wave (HM ratio).

Experiment II

Presynaptic inhibition of Ia fibers was examined in 7 volunteers (aged 18-28 years) underwent the BR for 20 days. They maintained inactive throughout the BR period. The change in presynaptic inhibition was calculated as the difference in the level of monosynaptic Ia facilitation on the soleus elicited by a stimulation of heteronymous Ia volley on the quadriceps (Hultbom 1987) between before and after the BR. Heteronymous Ia facilitation of the soleus H reflex was obtained by stimulating the femoral nerve in the femoral triangle with a constant intensity (1.2 x MT) in supine position. The stimulating intensity of the soleus muscle was adjusted to give a relative amplitude of 15% to the maximal M response. The interval time of conditioning-test stimulation which produced the peak of facilitation in each subject was used. The level of heteronymous Ia facilitation was expressed as a percentage to the amplitude of soleus H reflex without conditioning stimulation.

RESULTS and DISCUSSION

There was a significant decrease in HM ratio after the BR in the control leg, but no significant change in the trained leg (Fig 1) .This findings seem to suggest that the excitability of the motoneuronal pool and / or the transmission across the Ia synapses might be changed in response to prolonged inactivity. Then, we examined the presynaptic inhibition which reflected a descending influence modifying the excitability of motoneuron before and after the BR. There was no significant difference in the mean value of facilitation of the soleus H reflex by stimulating quadriceps Ia fiber between before (142 ± 39 %) and after (130 ± 36 %) the BR. These results suggested that the amount of presynaptic inhibition was not increased after the BR. Accordingly, we speculated that the decreased excitability of the soleus motoneuron was not related with the extent of presynaptic inhibition on heteronymous Ia terminals, but with a decrease in connectivity from Ia fibers to the motoneurones. In this study the level of HM ratio in the trained leg was maintained during the BR, and this seem to suggest that the level of activity on muscle contraction may influence on the excitability of monosynaptic spinal pathways.

CONCLUSION

The inactivity during the BR for 20 days reduced the level of excitability in spinal reflex, but this change was prevented by the resistance exercise training during the BR.

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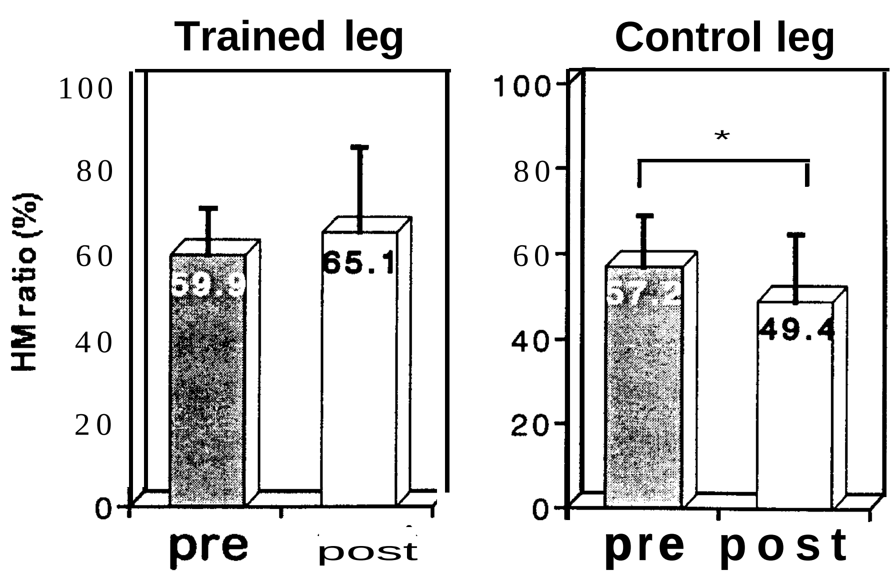

Fig.1 Change in H reflex following prolonged bed rest

Physiological Parameters and Excitability of the Spinal Cord Accompanying Ejaculation in Spinal Cord Injured Men

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INTRODUCTION

Sexual function, more specifically ejaculatory function, is severely compromised in men following a spinal cord injury (SCI). Different techniques that either enable SCI men to obtain or improve ejaculatory function include vibratory or rectal probe stimulation, with or without pharmacological agents, direct sampling in either the vas deferens or the epididymis and testicular biopsies*. However, these techniques often use a *trial and error* approach. The purpose of the present study was to document the physiological parameters and the excitability of the spinal cord (as measured by the H reflex technique) that accompany the ejaculation in SCI individuals. Ultimately, the information could be used to provide bio-feedback in order to improve ejaculatory success in this population.

METHODS

Surface EMG signals were recorded from 3 muscles, Bulbospongiosus (BS), Rectus Abdominis (RA), and Soleus (Sol). Three electrodes were placed on the chest to monitor heart rate (HR). A strain gauge was place on the penis below the glans to measure penile circumference changes (plethysmograph - PG). The signals were filtered (EMG, 1 O-300 Hz; ECG, 0.3-l 0 Hz; PG, O-l 00 Hz), amplified and digitized at 1 KHz. An electronic sphygmomanometer was used to measure blood pressure (BP) changes. The Sol H reflex was obtained by applying a single 1 ms monophasic pulse to the tibia1 nerve in the popliteal fossa through Ag-Ag Cl electrodes fastened to the skin with adhesive tape. The stimulus intensity was set to produce a small motor (M) response that was 10% (± 1%) of the maximal M wave.

With the subject seated in a semi-reclined position, control HR, PB and H reflexes were recorded. The stimulus for the H reflex continued until the end of the experiment at a rate of 8-l 0 per min. A 110 Hz, 2.5 mm vibration (Multicept ApS, model Ferti-Care Personal) was applied to the penis until ejaculation occurred. BP was taken immediately foilowing ejaculation and every 2 min. thereafter for 18 min.

RESULTS

Fig. 1 illustrates the results of one subject (a C_6-C_7 tetraplegic, 19 years post-lesion) in whom the experiment was repeated 5 times with a minimal of 5 days between sessions. The results were practically identical across the 5 sessions. Several small EMG bursts in BS accompanied by rapid changes in penile circumference began to appear shortly after the onset of vibration (Fig. IA). The BS and RA activity became more consistent and increased gradually. At the same time, the circumference of the penis began to increase gradually. As ejaculation approached, there was rapid increase in the BS burst followed by sudden drop of activity which was accompanied by a sharp increase in the RA. Just prior to ejaculation, there was another a sharp increase in the BS EMG accompanied by an inhibition of the abdominal muscles. Ejaculation coincided with this RA inhibition. Also coinciding with ejaculation was a rapidjerky increase in the penile circumference. There was a quick resumption of the RA activity following ejaculation after which there was a gradual decline in the activity of the BS and RA. The penile circumference began to decrease more smoothly and at roughly the same rate as the previous increase.

Before ejaculation and coinciding with the initial increase in the BS and RA EMG, the HR began to slow down while both systolic (SBP) and diastolic (DBP) BP started to rise (Fig 1 A & B). The HR continued to decrease during ejaculation and reached a minimum at approximately 2 min post-ejaculation. Similarly, both SBP and DBP continued to increase to reach a maximum at around the same time. The HR and BP gradually returned to normal values, although neither reached those values by 18 min post-ejaculation (Fig. 1 B).

Fig. 1C illustrates that immediately after the onset of vibration, the H reflex dropped to 60% of its control value. The H reflex began to increase after the offset of vibration and it peaked (169%) at about 6 min post-ejaculation. As with the HR and BP, the H reflex still had not returned to initial values after 18 min.

DISCUSSION

The RA and BS EMG activity results support clinical reports of lower abdominal spasms and strong rhythmic contractions of the testicles that are noted visually during fertility tests in SCI men. Interestingly, this pattern of muscular activity seems to coincide with neurological innervation involved in the ejaculatory process. This process can be viewed in 3 phases, (1) penile tumescence that primarily involves sacral innervation, (2) seminal emission that involves thoraco-lumbar innervation and (3) ejaculation per se involving sacral activity. The data, although from a single subject, suggest that BS bursts coinciding with PG changes correspond to sacral activity that occur during stimulation. These bursts are followed by steady contraction indicating oncoming ejaculation. While these contractions temporarily cease for a few seconds preceding expulsion of semen, RA activity, innervated by thoracic roots, shows strong bursts which may coincide with the phase of seminal emission. These RA bursts are followed by a resumption of BS activity coinciding with the actual expulsion of semen and corresponding to sacral muscular activity of the ejaculatory process. These findings from EMG, PG and HR suggest a physiological pattern leading to ejaculation in SCI men and which could be used in a biofeedback approach to increase the success rate of ejaculation in this population.

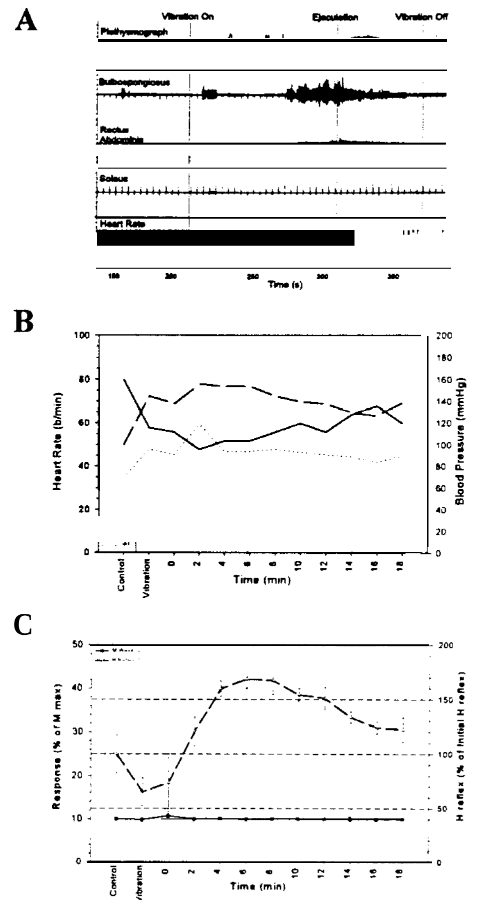

Fig. 1 **A)** Raw Data; B) Heart rate and systolic and diastolic blood pressure changes; C) H reflex and M wave.

The changes in H reflex suggest modifications in the excitability of the spinal cord both before and after the ejaculatory process. The decreased H reflex following the onset of stimulation may be attributable to activation of the cutaneous afferents within the same dermatome (i.e., S,) [1]. The overshoot of the H reflex may be related to the release of neurotransmitters by the autonomic system which increase the excitability of the cord and have long lasting effect on the H reflex response.

ACKNOWLEDGMENTS

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Stability of the Stretch Reflex Threshold as a Measure of Spasticity in Children with Cerebral Palsy

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PURPOSE : The stretch reflex (SR) threshold and its regulation play an important role in motor control. We studied the stretch reflex threshold in children with cerebral palsy (CP) as a potential measure of the effect of therapeutic interventions to decrease spasticity and improve motor control.

SUBJECTS : Ten children with spastic CP aged 6 to 18 years entered the study after informed consent was obtained from parents and/or guardians. Children with dyskinesia or elbow contracture were excluded. A minimal cooperation level was also required for children to participate in the study.

METHODS AND MATERIALS : All children visited the laboratory three times. For each visit, clinical spasticity was rated by a physiotherapist on a 16-point scale based on resistance to stretch of the elbow flexors at rest, biceps brachii tendon jerk and wrist clonus. SRs of the affected elbow flexors were elicited by 100” displacements of the elbow (initial position approximately **40°** between arm and forearm) toward extension produced by a torque motor. EMG recordings were obtained by surface electrodes for elbow flexors (biceps brachii and brachioradialis) and extensors (triceps brachii and anconeus). A total of 56 stretches were performed (8 randomized trials at 7 velocities: 8, 16, 32, 53, 80, 120 and **160°/s**). Only those trials where the subject was relaxed and did not intervene during the stretching were kept for further analysis.

ANALYSIS : For each trial, the angle and velocity corresponding to the onset of EMG activity in the flexor muscles were determined (see fig.1). These points, representing the dynamic SR thresholds for each velocity of stretch, were plotted on mean velocity vs mean angle phase diagrams. Linear regression analysis through the dynamic threshold points was used to determine the slope and the static SR threshold (when velocity = 0). A correlation matrix was obtained for the three subscores of the clinical spasticity scale, SR threshold, and the velocity sensitivity **(μ)** of the stretch reflex.

RESULTS : In contrast to healthy children who only show elbow flexor SR responses to stretch velocities higher than **300°/s**, children with CP showed SRs to velocities as low as **8°/s**. A weak but significant correlation ($r = 0.483$, $p = 0.031$) was found between clinical spasticity and SR thresholds. The variability in both measures was found to be higher in the children with CP than in hemiplegic adults recorded in previous studies. The velocity sensitivity of the SR threshold was also higher in children than in adults. Another interesting finding was that the activation of elbow extensors occured simultaneously with the flexor activation in almost all the children studied.

CONCLUSION : SR thresholds can be used along with clinical measures to document changes in proprioceptive reflex activity after therapeutic interventions. Further studies are necessary to address the question of how changes in the regulation of the SR threshold is related to motor control in this population.

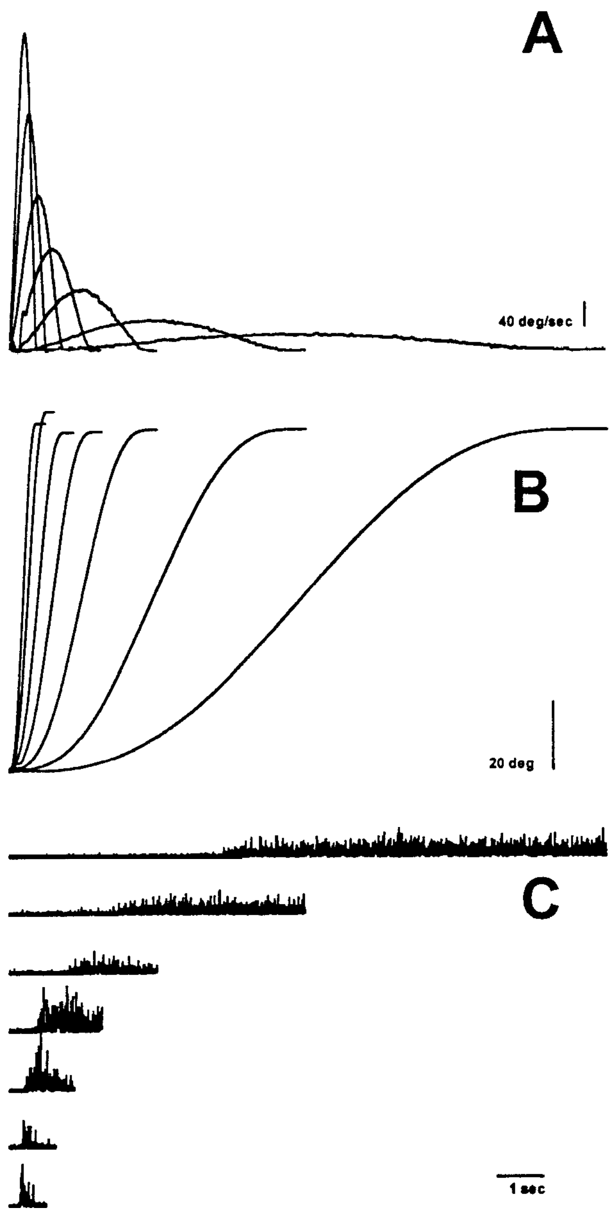

Fig.1 **A.** Velocity vs time; B. Angle vs time; C. EMG traces to determine the angle and velocity corresponding to the EMG onset at each mean velocity of stretch in one subject. A, B, left to right/ and C, bottom to top: 160, 120, 80, 53, 32, 16 and 8 deg/sec..

Neck Flexion and Magnetic Evoked Potential in Hirayama Disease

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INTRODUCION

The symptoms of Hirayama disease are the onset of 15~25-year-old male ,unilateral distal atrophy and weakness of distal muscles of upper extremity, and progressive course in few years after the onset. We studied functional assessment using transcranial magnetic stimulation in flexion and neutral position of neck in the progressive state of Hirayama disease.

METHODS

The subjects were a patient of Hirayama disease and 4 healthy volunteers. The case was 19-year-old male. His symptoms of weakness of rt hand muscles occurred in 1 B-year-old. He was diagnosed Hirayama disease with neurological findings and spinal MRI and CT myelography .4 healthy volunteers were 3 males and 1 female,and the mean of their age, 25.3-year-old. As a method, the recording electrode was put on the abductor digiti minimi muscles in sitting subjects. They bent their neck in maximum flexion(45-50degree) and we measured magnetic evoked potential (MEP) before neck flexion, during neck flexion, and after returning to the neutral position. The output of magnetic stimulation was maximum.

RESULTS

In healthy volunteers, the amplitude of MEP was low by neck flexion but soon it was returned to the same as before(fig1). In Hirayama disease, the MEP was polyphasic and became more polyphasic potentials during neck flexion. After returning to the neutral position, the change of the shape of MEPs remained and returned in a few days after trial(fig2).

fig, 1 MEP in flexion and neutral position of neck

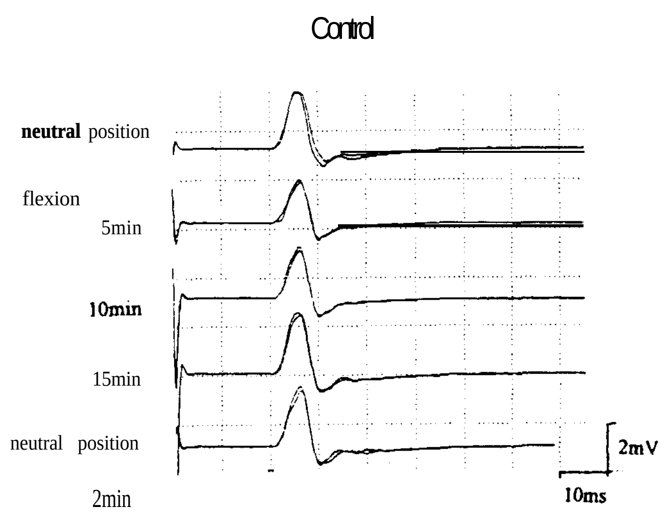
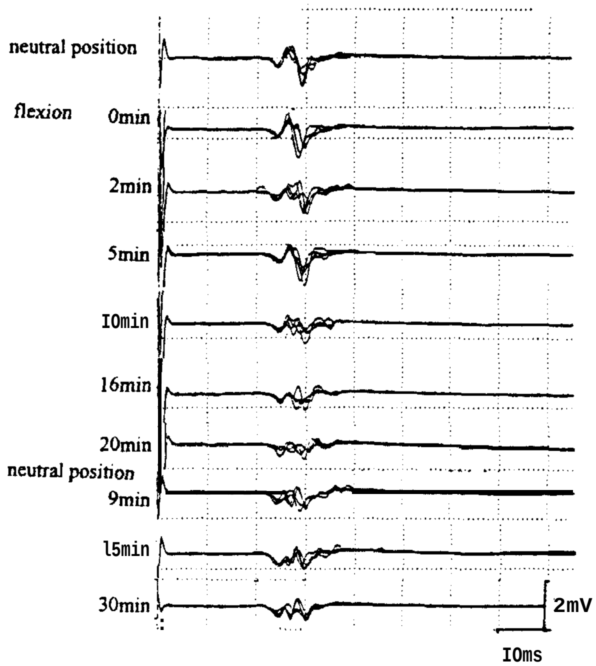


fig.2 MEP in flexion and neutral position of neck

Hirayama disease



DISCUSSION

In Hirayama disease, microcirculation failure in the anterior spinal artery in C7,C8 and Tl by pressing spinal cord with neck flexion could occur, then the therapy for this would be to limit the range of motion to neck flexion and to put on neck collar[1~3]. This date indicates this therapy would be reasonable.

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Differences in Neuromuscular Activity of the Upper Trapezius Muscles Between WAD Grade II Patients and Healthy Controls

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INTRODUCTION

The Whiplash Associated Disorder (WAD) is charactarised by a discrepancy between symptoms and signs. The most frequently mentioned symptoms are pain in the neck, head and the shoulders accompanied by a feeling of stiffness. The only consistent finding at physical examination that is described seems to be the ‘muscle spasm’. In the proposed clinical classification of the Quebec Task Force this ‘musculoskeletal sign’ is one of the criteria that defines WAD II (1). In chronic low back pain patients (CLBP) there is some evidence of abnormal EMG-patterns but in literature on WAD there are no reports concerning the use of SEMG to asses muscle activity.We want to try to develop an instrument to discriminate the WAD II patients according to the alterations in muscle activity. A first step in this proces is to identify abnormal muscle activity in these patients. Therefore the goal of the following exploratory crosssectional study is to demonstrate differences between the EMG-activity of the neck musculature of WAD II patients and healthy controls during different physical tasks.

METHODS

Subjects: A group of 18 chronic WAD II-patients and 19 healthy controls participated in the study. All patients complained of pain in the neck, head or shoulder region for more than 6 months without the presence of either the preexisting or trauma related orthopedic or neurological signs. In all of them pain started within the first 48 hours after a rear-end motor vehicle collision.

Measurements: The participants performed several simple tasks with a varying physical demand after an initial normalisaton procedure of the SEMG amplitude from the upper trapezius muscle using submaximal voluntary contractions (2). Measurement were done during:

A) static activity comprising of 3 assessments: 1) sitting in a comfortable chair with head supported. 2) sitting on a stool and 3) standing.

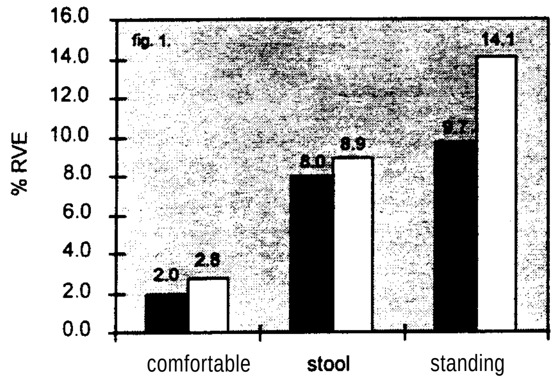
B) dynamic activity: the participant sat at a table and then was asked to put marks with a pencil held in the dominant hand on three spots arranged in a triangular fashion. The non-dominant arm rested on the table. This activity was carried out for around 2½ minutes (3). During this activity EMG-activity of both the dominant ‘active’ side and the non-dominant ‘passive’ side was measured at 10 sec., 60 sec. and 120 seconds.

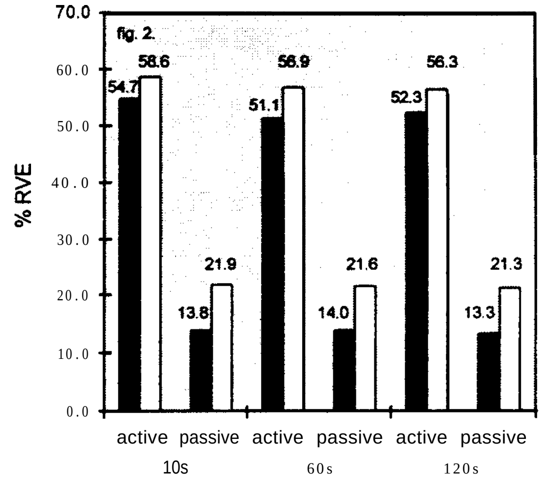
C) a condition requiring relaxation: a second standing task was performed following the dynamic task and was compared to the first standing task.

Each EMG-signal was recorded during 15 seconds and processed to produce a Smooth Rectified Electromyography (SRE). The averaged SRE-value was expressed as a percentage of a reference voluntary electrical activation (%RVE).

RESULTS

No statistically significant differences between groups during the static activity (A) were found (MANOVA, $p< 0.154$) (fig. 1). No statistically significant differences between groups during the dynamic activity (B) were found (MANOVA, $p<0.411$) (fig. 2). During this activity there was some muscle activity in the ‘non-dominant’ passive side. Differences between the two groups were greater in the passive side than in the active side but these were not statistically significant (fig. 2). The difference between the first and second standing task (C) for both groups for the active side (fig. 3a) as well as for the passive side (fig. 3b) was determined. Statistically significant differences were found in the change of EMG-activity between groups for either side (t-test: p<O.OOl, fig. 3a; p< 0.0001, fig.3b).





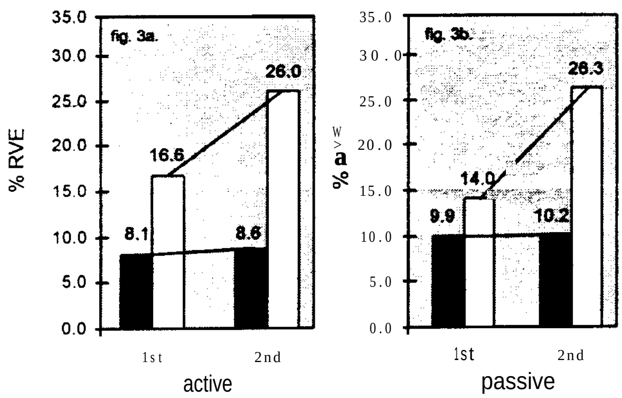


fig. 1. EMG-activity during three static tasks (MANOVA, p<O. 154).

fig. 2. EMG-activity of active and passive side during a unilateral dynamic task (MANOVA, p<O.411)

fig. 3a and 3b. Difference in change of EMG-activity of standing tasks before and after a unilateral dynamic task (left: active side, p<O.001 and right: passive, p<O.OOOl, t-tests).

■ control □ patient

CONCLUSIONS

1. Statistical significant differences in EMG-activity between WAD-2 patients and healthy controls do not appear during a physical activity, but rather after a physical task, reflecting the inability to relax to a baseline EMG-level.
2. During a unilateral dynamic task WAD-2 patients show a tendency of more co-contractions on the relaxed side than healthy controls, which supports the first conclusion that patients are unable to relax.

DISCUSSION

In this first attempt to find an instrument to identify WAD II-patients , especially the inability to relax after physical activity appeared to be a sensitive indicator. However in determining the diagnostic value of EMG-measurements in WAD II additional information of the differences between aspecific chronic neck pain patients and chronic WAD II patients is needed.

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Impact of Median Nerve Compression on Sensation and Hand Dexterity

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INTRODUCTION

Normal hand function requires intact motor and sensory innervation of the hand. Sensory information from the finger tip plays a crucial role not only in perception, but also in the regulation of motor function of the hand (Johansson & Cole, 1992). The aim of this study is to analyse the functional impairments induced by severe carpal tunnel syndrome (CTS) on sensation and hand dexterity.

METHODS

Ten women and four men participated in this study. Their mean age was 55 (S.D. 15.6). The 14 subjects presented the classical clinical features of CTS in 19 hands. They were recruited from electromyography laboratory where nerve conduction studies had shown evidence of severe CTS. Apart from CTS, the subjects were healthy and the EMG was normal.
Hand function was assessed by quantitative tests measuring tactile fingertip sensation and digital dexterity. Index and fifth finger tips sensation was assessed by determining the touch detection threshold and the cold and warm thermal perception thresholds. Digital dexterity was evaluated with the unimanual Purdue Pegboard subtest and a Z score was calculated for every patient by comparing his result with normal values (Tiffin & Asher, 1948; Desrosiers et al., 1995).

RESULTS AND DISCUSSION

Table 1 presents the main results of nerve conduction studies for the 19 hands and shows the severity of the CTS.

Table 1: Results of nerve conduction studies

Variables	Site of detection	Mean ± S.D. (ms)	no response (n)	Norms (ms)
Median nerve motor distal latency	APB (7 cm)	6.7 ± 1.5	4	<4.4
Median nerve sensory distal peak latency	Index (13 cm)	5.2 ± 0.6	13	<3.5
Difference between median and ulnar nerve motor distal latencies	SL&SI (10 cm)	2.6 ± 1.4	2	<0.4

APB: Abductor Pollicis Brevis muscle; SL: Second lumbrical muscle; SI: Second dorsal Interosseous muscle

Table 2 presents the median values with the interquartile range of the sensation variables for the 19 hands along with the results of the statistical analyses. The index touch detection threshold is slightly impaired and statistically different (Wilcoxon Signed Rank Test; $p<0.05$) when compared with the fifth finger indicating a perturbation of tactile sensation in the territory of the median nerve. The thermal thresholds were identical in median and ulnar nerve territories, suggesting that the small-diameter fibres were not affected.

Table 2: Results of the sensation thresholds on the index and fifth fingers

Variables	Median (Interquartile range)		Comparison of fingers		
	Index	Fifth finger	n	W-value	P-value
Touch detection threshold (g)	0.7 (0.4-1.4)	0.2 (0.1-0.4)	19	-120	<0.001
Warm detection threshold (°C)	37.4 (34.9-46.3)	38.4 (35.3-40.1)	19	-41	0.418
Cold detection threshold (°C)	27.5 (21.1-30.0)	27.6 (22.8-28.7)	19	6	0.900

In all but the three oldest patients, the Purdue Pegboard test results are within the two SD of the published normal values. The mean Z score for the 14 patients is -1.6 (range: -5.6 to 0.2) which suggests that digital dexterity was not significantly perturbed in our sample of CTS patients.

CONCLUSION

Despite the severe abnormalities of median nerve conduction, our results suggest that CTS induces relatively few impairment of tactile perception with minor impact on digital dexterity.

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Age and Training-Related Adaptations in Motor Unit Discharge Activity with Force Modulation

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INTRODUCTION. With **aging,** control and regulation of muscular force is characterized by considerable variability which may compromise the execution of motor **performance**[2]. While such impaired control might be explained by age-related physiological effects including increased motor unit (MU) innervation ratios and concomitant increases in MU force, improvements in the control of muscular force have been demonstrated to accompany the early weeks of strength trainings. Such rapid adaptation suggests a conspicuous neural influence. Moreover, alterations in MU discharge behaviour and in co-contraction of antagonist muscle pairs have been demonstrated in elders, suggesting significant modifications in the neural organization of muscular **force**[3,4,7]. The present study was conducted in an effort to better understand age-related differences in the neural mechanisms governing control and modulation of muscular force including those involved in practice -related adaptations.

METHODS Two groups of healthy, physically active individuals, 6 young (mn=19.17 yrs) and 6 older (mn=70.33 yrs) adults, performed a force trajectory task involving isometric ankle dorsiflexion ranging between 0 and 60% of MVC (Fig. 1). Over six weeks, all subjects trained to improve force modulation accuracy as evidenced by reduction of the root-mean-square (RMS) error between the criterion and performed trajectories. MVC force was assessed at the beginning of each week of training, and the trajectory scaled according to the subject's current performance. At baseline (Day 0), following 2 weeks (Day 14), and upon completion (Day 42) of training MU discharge activity was recorded from the tibialis anterior during performance of the trajectory task using a quadrifilar needle electrode. MU discharge occurrences were identified using custom-written spike recognition software. MU discharge rates were calculated as the inverse of the five shortest inter-firing intervals over **a** one second period of sustained discharge at each force level (lo%, 20%, 30%, 40% & 60% MVC).

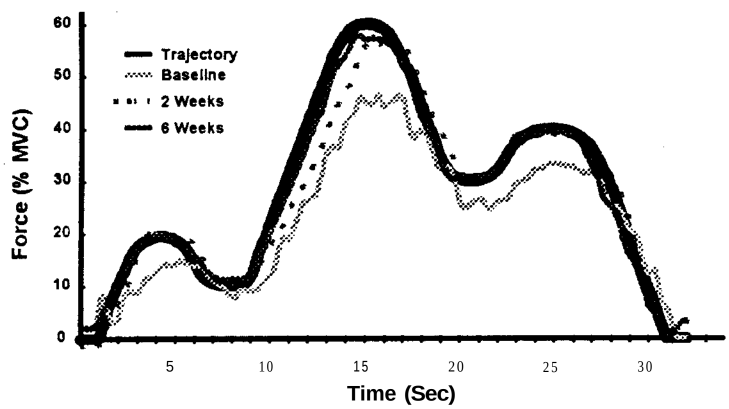

Figure 1. **Illustration of trajectory (solid black line) and performances by a young subject at first exposure to the force modulation task (solid gray line), following 2 weeks (dashed line), and 6 weeks (dark gray line) of training. RMS error calculated between the subject's performance and the criterion task decreased significantly over the first two weeks.**

RESULTS **In** response to training with the trajectory task, both young and older adults demonstrated significant improvements in accuracy of force modulation. Decreases in RMS error were comparable in both groups (44% young, 47% older), although they were evidenced in the young individuals following only 1 week of training while in the older individuals significant improvements were evidenced over 2 weeks (19% wk 1,28% wk 2). MVC values for young and older individuals did not differ significantly

Table 1.		**Day 0**	**Day 14**	**Day 42**
RMS Trajectory Error	Young	**174.5**	**97.4"**	**76.4**
	Older	267.0	139.4"	120.9
MVC Force	Young	169.3 N	203.7 N*	224.2 N**
	Older	163.2 N	167.0 N	167.4 N

RMS **error scores are reported in arbitrary units.** (* p < .05, ** p < .OOI) at the onset of the training program. However, a remarkable finding was a gain in dorsiflexion strength evidenced by only the young individuals (12.4% Day 14,21.4% Day 42). These data are enumerated in Table 1.

In conjunction with the behavioural changes, following the first 2 weeks of training MU discharge rates decreased significantly in both young (5-14%) and older (12-19%) individuals at 30 - 60% of MVC ($p < .03$). In both age groups, the range of MU discharge rates demonstrated at each force level was similar between 10-40%, but was significantly greater at 60% of MVC ($p < .001$). As a result of training, older but not young, adults demonstrated a progressively decreased variability of discharge rates among simultaneously active motor units ($p < .05$).

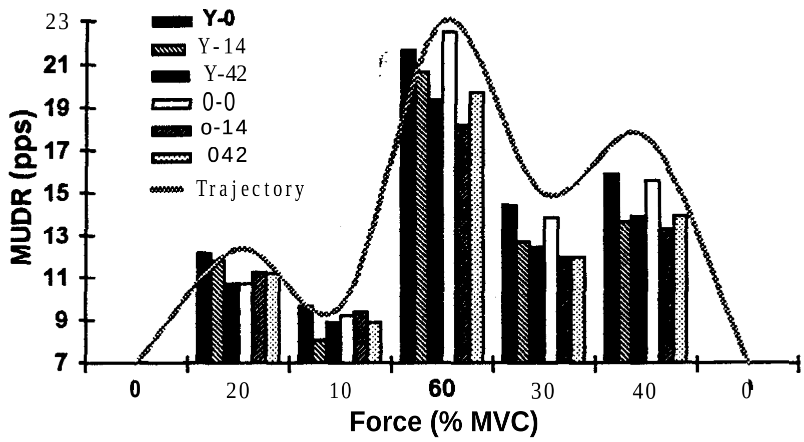

Figure 2 **MU Discharge rates. Both young and older individuals demonstrated significant decreases in MUDR at 30,40 & 60% of MVC force following 2 weeks of training with the trajectory task. Notably this period of training was also associated with significant improvements in performance of the trajectory task.**

DISCUSSION Consistent with earlier investigation, results of the present study demonstrate significant training-related improvement in regulation of muscular force5 and a two-phase pattern of adaptation[6]. In the early phase, behavioural improvements in force modulation are accompanied by physiologic adaptation in MU discharge activity. The rapid nature of these adaptations suggests Supraspinal influences affecting both the level of central drive and the temporal patterning of activation of the motoneuron pool. Despite established age-related changes in MU function, this adaptive process appears to remain functional in older adults. Improvements in force modulation did, however, require twice the effort by older adults, and alterations in MU discharge rate demonstrated following two weeks appeared to regress somewhat towards baseline levels by completion of the six weeks of training. These significant age-related differences suggest subtle degradation in adaptive potential of the nervous system.

Effects demonstrated in the later phase of training indicate that neural influences on the organization of muscular force are more broad ranging than might be described through the discharge activity of individual motor units. Variability in MUDR, described *using the* standard deviation of firing rates among simultaneously active motor units, decreased in the older individuals demonstrating a significant cumulative effect upon completion of the six weeks of training. This aspect of MU discharge behaviour remained stable in the young individuals throughout the study. Disruption of common drive to motor units has previously been observed in elders 1. The results of the present study indicate that a significant effect of force modulation training is enhancement in common drive affecting ensemble activity of the motor unit population.

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Motor Unit Recruitment Strategy Changes Following Exercise-Induced Muscle Injury

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INTRODUCTION

To date, studies of exercised-induced injury in humans have evaluated groups of muscles acting about a joint. This has made it difficult to determine the amount of damage that has occurred in any one muscle. Furthermore, careful analysis of changes in recruitment of motor units within an individual muscle has not been performed. The purpose of this study was to investigate an exercise-induced muscle injury specific to a single muscle acting about a single joint.

In the present study two hypothesis were addressed: 1) injury to large muscle fibers will render them less able to produce force. At force levels where large motor units are normally required, the reduced contribution of injured large muscle fibers will be compensated by increased activation of slow motor units, and; 2) motor function during sub-maximal muscle activation will be impaired following exercise-induced injury of the muscle, but only if the task requires the recruitment of large motor units.

METHODS

General Design

Ten normal male subjects participated in this study (age range 22 - 31). Subjects were tested on two separate occasions, previous to an exercise-induced muscle injury and 24 hours after muscle injury. The exercise program consisted of two sets of 50 maximal eccentric contractions with the FDI muscle in adduction-abduction. To ensure reproducible placement of EMG on both days, an indelible pen was used to mark the skin.. EMG was recorded using active, bipolar, stainless steel, surface electrodes (Liberty Mutual MYO 111) with electrode contacts 4 mm in diameter and 13 mm apart. EMG signals from the FDI were bandpass filtered from 45 - 550 Hz, amplified, digitized at 2 KHz and stored on disk for later analysis. Torque of the index finger was measured by a linear strain gauge, amplified and digitized at 2 KHz.

Testing Procedures

1) Torque tracking the subject was required to contract the FDI muscle isometrically to produce MP abduction torque. The target torque began at zero and increased to 65% of pre-injury MVC at a constant rate over a period of five seconds. The target torque increased at a constant rate and the subject was instructed to match it. Five trials were collected. This experiment was performed to evaluate the ability to control torque output.

2) *Sustained contraction.: the* subject was required to maintain a constant isometric MP abduction torque for 60 seconds, equal to 50% pre-injury MVC. Only one trial was performed. This experiment was performed to evaluate the ability to sustain torque output and the fatiguability of the injured muscle.

Analysis

Torque tracking: median frequency (MF) and mean rectified EMG (MEMG) during each one second interval of the 5 seconds. For MF a 256 point spectrum was computed over an interval of 0.896 s, which allowed 3 spectra to be averaged. The variance about the best-fit straight line from beginning to end of the torque tracking was also determined. The mean parameters for the 5 trials were then calculated.

Sustained contraction: MF was computed at 5 second intervals from the average of nine 5 12 point spectra computed over an interval of 4.864 s. The slope of the MF plotted against time was computed, as well as MEMG for the first and last 5 s intervals.

RESULTS

Variability in tracking a target torque was greater 24 hours after the injury compared to pre-injury ($p<0.05$). The MF of the FDI EMG was similar pre- and 24 hours after injury during the first three seconds of contraction. However, it was significantly lower during the fmal two seconds 24 hours after injury ($p<0.05$ and p<O.OOl). The MEMG, on the other hand, was significantly greater during the first four seconds of contraction 24 hours after injury (p<O.OOl, p<O.OO1, p<O.Ol and $p<0.05$).

The similarity in slope of the MF during a 60 second contraction at 50% of MP abduction MVC suggests that the FDI muscle was not more fatiguable 24 hours after the injury than pre-injury. However, the initial and the final values of MF 24 hours after injury were significantly lower ($p<0.05$) and the initial MEMG was significantly greater 24 hours after the injury ($p<0.05$).

DISCUSSION

Solomonow et al.[1] has shown that an increase in MF was directly associated with the recruitment of large diameter fibers, whereas an increase in the firing rate did not affect MF. They concluded that average conduction velocity during motor unit recruitment was the major contributor to variations in MF. Muscles having a higher percentage of fast fibers would have a correspondingly greater initial MF[2]. The finding that the initial MF at 50% MVC was lower 24 hours after the injury suggests that slower motor units were used to initiate the sustained contraction. Initial MEMG was greater, consistent with higher activation of these units. These observations suggest that the muscle injury caused substantial changes in activation patterns. The lower final MF is consistent with reduced activation of larger fibers throughout the contraction. Berry et al.[3] suggested that if there was impaired function of the high threshold fast fibers following eccentric exercise one would expect the MF to shift to a lower value.

During torque tracking, the MF did not begin to differ significantly between pre- and 24 hours after injury until the torque reached 30% MVC. At this point the MF dropped below the pre-injury value, suggesting reliance on the small fibers. The greater MEMG throughout the first four seconds suggests that the CNS increased the activity of the small fibers to compensate for injury to large fibers. The shift of the MF to lower frequencies was also evident in the task requiring a sustained contraction of 60 seconds. This task was used to investigate the fatiguability of the FDI after muscle injury. Other researchers have reported reduced endurance times and low frequency fatigue following injury. Kroon and Naeije[4] reported an increased rate of decline of the EMG mean power frequency over the first 30 seconds of a sustained contraction and suggested that this was due to increased fatiguability of the muscle. Using this index of fatiguability, the results of this study suggest that the FDI muscle was no more fatiguable after muscle injury, contrary to their findings. However, although the slope of the MF did not change during a 60 s contraction there was an overall shift of the MF to lower values, indicated by the lower initial and final MF. This would suggest greater reliance on slow motor units which would make the muscle less fatiguable.

Following muscle injury, subjects were less able to track torque smoothly. This impairment was most evident at higher torque levels. The increase in torque variance may have resulted from decreased perception of muscle force, irregular jumps in force as the larger injured motor units were recruited, irregular changes in firing rate or tremor[5]. The shift in MF at the beginning of the trial illustrates that changes in neural control occurred even at low force levels.

These findings provide the first evidence that there is a shift in activation of motor units after muscle injury. The finding of a decrease in MF in tasks requiring high torques is consistent with injury and reduced activation of the large fibers. The increase in MEMG also indicates that greater activation of the small fibers is needed to complete the tasks. No other studies have reported such findings. However, this was the first study to investigate exercise-induced muscle injury in a single muscle.

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Stretch Reflex Sensitivity in Human Elbow Flexor Muscles

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INTRODUCTION

It is well recognized that stretch reflex responses are modulated according to motor tasks. However, little has been known about the stretch reflex sensitivity of human muscles, which should be determined from the relationship between the intensity of stimulus and the evoked muscle response, since quantification of mechanical stimuli to muscle spindles has been so far quite difficult in vivo. Recent developments of the ultrasonography, however, have made it possible to measure morphological features of human muscles, such as pennation angle and fascicle length (Fukunaga et al. 1995). In the present study, to quantify the intensity of stimuli the ultrasound images were utilized to estimate muscle length changes induced by mechanical stretches of elbow joint. The aims were 1) to quantify the stretch reflex sensitivities of elbow flexor muscles as the relationship between estimated muscle length changes by mechanical stretches and reflex EMG responses, and 2) to compare the stretch reflex sensitivities among reflex EMG components and elbow flexor muscles.

METHODS

Stretch reflex test.

Mechanical stretches at various velocities (50 to 250 deg/s) were used to induce stretch reflex EMG responses in elbow flexor muscles, the brachioradialis (BRD) and the biceps brachii long head (BBL). A series of ten stretches was applied to the elbow joint in the one trial. Four different background activity conditions, 0%, 3%, 6% and 9%MVC, were employed, and two sets of the stretches were applied to induce stretch reflexes in each condition.

Recording of the ultrasonogram

Ultrasonograms of the BBL were recorded for each subject to estimate the muscle stretch velocities induced by the different angular velocity stretches. The five images were recorded at every 10 deg from 0 deg to 120 deg flexed elbow position in the same posture with that in the stretch reflex test. The same recording was done with the three different background activity levels, 0%, 3% and 6%MVC. Since it was impossible to record clear ultrasonograms from the BRD, the recording was done only for the BBL.

Biomechanical model

A biomechanical model of the elbow joint and muscles was used to estimate the muscle stretch velocities, too. In the present study, the model proposed by van Zuylen et al. (1988) were utilized with the other anatomical data, such as the pennation angle, the tendon vs muscle ratio calculated by MRI images obtained previously.

Data analysis

Reflex EMG responses were divided into four components on the basis of latencies from the onset of the perturbation, i.e., Ml; 20-50ms, M2; 50-80ms, M3; 80-100ms VOL; lOO-150ms. Mean rectified EMG (MR-EMG) activities for those periods were calculated and expressed in units of isometric force (Nm) at the elbow angle of 75 deg. In order to obtain the coefficient, which transforms the EMG unit (mV) to the force unit (Nm), the least square method was used to fit a linear regression line to the relation between background EMG activities in the four contraction levels and corresponding forces, and the slope of the linear regression line was used as the coefficient.

RESULTS

The pattern of muscle length changes with elbow angle measured by the ultrasonogram was basically similar to that estimated from the biomechanical model, although the biomechanical model tended to estimate slightly larger length change than that from the ultrasonogram. There was no significant difference due to the muscle contraction level in the pattern of length change with joint angle. The mean, therefore, of all measurement conditions was used to estimate the muscle stretch velocity induced by the perturbation.

Larger reflex EMG responses were observed in conditions such as the higher mechanical stretches and the larger background activities. In the range of stretch velocity in the present study, there was the linear relationship between the estimated muscle stretching velocity and the reflex EMG responses. Correlations were ranged from 0.47 to 0.75 and from 0.47 to 0.83, for Ml and M2, respectively.

Generally, both Mland M2 to the same mechanical stretch were larger in BRD than in BB especially in lower BGA conditions. However, the regression analysis revealed that for the relation between the estimated muscle stretch velocity by both the ultrasonography and the biomechanical model, and the reflex EMG response, there was no significant difference in the slopes of the linear regression lines for M 1 in all BGA conditions and those for M2 in low BGA conditions, 0% and 3%MVC, while in the higher BGA conditions the M2 gains were higher in BBL than in BRD.

Both MR-EMG and sensitivity of the long latency component M2 were significantly greater than that of the short latency component M 1 except in the no-voluntary contraction condition, O%MVC.

DISCUSSION

Results of the present study indicated that there were highly significant correlations between the estimated muscle velocity and reflex EMG responses. Eng and Hoffer (1997) recently reported that locally detected reflex EMG amplitude was well correlated with the corresponding local fibre stretch velocities but not with whole muscle stretch velocity. They detected EMGs from strictly restricted region with closely spaced steel electrodes, while in our study EMG was measured with surface electrodes, which were placed on muscle belly and conceived to represent whole muscle activity. The result of our study, therefore, strongly suggests that overall, reflex EMG responses, both Ml and M2 are modulated according to total length changes of muscle.

Although both Ml and M2 were increased linearly with the muscle stretch velocity, the slope of the linear regression line of M2 were significantly higher in voluntary contraction conditions (from 3 to 9%MVC), while there was no difference in the no-voluntary condition (O%MVC). This clearly indicates the existence of close relation between the higher nervous center and the long latency reflex response. It might be important to know with different muscle groups to what extent the reflex sensitivities of each component depend on the levels of voluntary activation. Unfortunately, there is no data concerning the reflex sensitivity in other muscle groups evaluated with the comparable method. Therefore, at present it is impossible to compare the present reflex sensitivity with those in the other muscle groups. Future studies are expected to clarify on this issue.

Generally greater reflex EMG responses were observed in the BRD as we have already observed (Nakazawa et al. 1997). However, there’s no considerable difference of the estimated reflex sensitivities in BRD and BB muscles. This means that the observed greater reflex EMG response in BRD might be to a large extent due to greater length changes of this muscle induced by the mechanical perturbations.

Finally, this novel method might give us chances to compare the stretch reflex sensitivity in different muscles, tasks and/or pathophysiological conditions.

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Relationship Between the Degree of Modulated Stretch Reflex Responses and Performance of Adjustment Movements

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[INTRODUCTION] Electromyographical (EMG) reflex responses induced by a sudden muscle stretch are composed of short latency components (Ml), that go by way of spinal pathways, and long latency components (M2), that are closely connected with cortical activity including the motor area. The gain of these reflex responses, particularly long latency component is modified by anticipation of a stimulus and by motor preparation for direction and type of movement. Furthermore, some studies have reported that the change in long latency component was more remarkable for the subjects with short reaction time than with long reaction time, as well as the long latency component was modified by motor preparation of reaction movement (Bonnet 1983; Woollacott et al. 1984, Kizuka et al. 1997). On the other hand, when subjects also have to perform adjustment movements quickly and accurately, it is necessary for them to construct the appropriate motor preparation for those movements at the spinal and supraspinal levels. Therefore, the ability of grading stretch reflex responses with appropriate preparation is thought to be linked with good performance of adjustment movements. In the present study, we used motor tasks that required positioning a handlebar, held in the hand, over two different target positions in response to a rapid muscle stretch in a wrist flexor. We investigated the relation between the performance of quick adjustment movements (adjustment time) and the degree of modulated stretch reflex responses.

[METHODS] Normal healthy subjects (16 males), ranging in age from 20 to 28 years (mean 23.4 years) participated in the study. A non-action and two target-alignment tasks to the same stretch stimulus were made as motor tasks of this experiments. A DC torque motor was used to give a stretch stimulus to the wrist flexor. The target positions were set up to be on a wrist extension side of 30^{o} (for extension-target task) and a wrist flexion side of 30^{o} (for flexion-target task) from the neutral position. In two target-alignment tasks, the subjects were instructed to perform an adjustment so that the position output of the handlebar aligned with a target position as quickly and as accurately as possible when they felt the rapid angle displacement in the direction of extension. In non-action task, the subjects were required to refrain from making any voluntary movements to the stretch stimulus applied to the flexor. The indexes of performance (adjustment time) were the time taken to move the handle to the target position from stretch onset in each target-alignment task. The short and long latency response (Ml and M2 components) were measured by averaging the rectified EMGs which recorded with the surface electrodes over the wrist flexor. In order to compare the degree of change in reflex responses between subjects, the integrated EMG of reflex responses in each target-alignment task was expressed as a percentage of that in the non-action task.

[RESULTS] Taking the integrated area of each reflex component in the non-action task as lOO%, the areas of the Ml and M2 components in the extension-targets task decreased to 80.0±10.0% and 66.3±14.6%, respectively. The area of the M2 component in the flexion-target tasks increased to

140.2±21.4%, but the change of Ml components was not remarkable. For the subjects who had short adjustment times, there was a clear tendency to make the change of the M2 components large in the target-alignment tasks. On the other hand, a tendency to make the change small was observed in the subjects who had long adjustment times. The correlation coefficient between the adjustment time and the change of M2 component was 0.894 for extension-target task (see Figs. 1) and -0.873 for the flexion-target tasks (see Figs. 2).

[DISCUSSION] It has been confirmed that the long latency component of stretch reflexes produces a tension corresponding to its amplitude. In this experiment, the tension produced by the M2 component of wrist flexor should obstruct the adjustment movement when the target position is extension side of the wrist joint. Contrary to this, when the target position is flexion side, the increase of the M2 component should help to perform the adjustment movements. The results in the study show that the adjustment time reduce as the long latency reflex in extension-target tasks is inhibited to decrease the tension to the direction of flexion, and as the long latency reflex in flexion-target task is facilitated to enlarge the tension to the flexion. From these findings, we claim that the degree of M2 modulation is closely connected with the adjustment ability of the neuromuscular system, which demands quick and accurate movements simultaneously.

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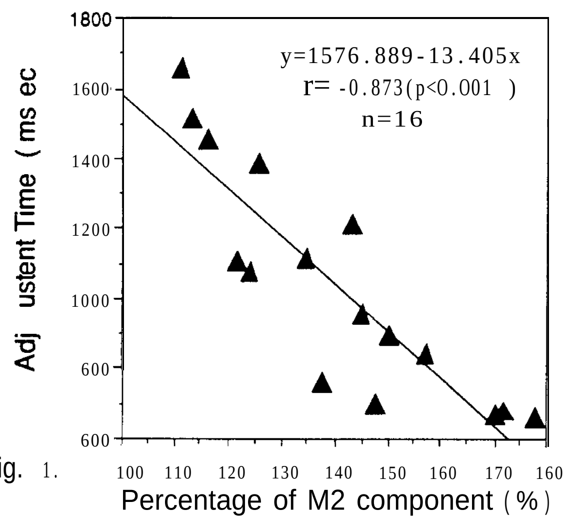

Fig. 1.

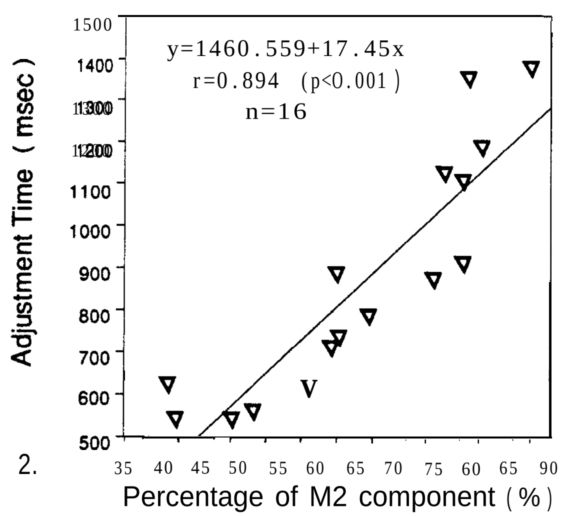

Fig. 2.

Correlation between the percentage of M2 component and Adjustment Time.
The line was determined by linear regression analysis.

Quantification of Time-Course Changes of Stretch Reflexes in Hemiparetic Subjects

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INTRODUCTION

Spasticity has been defined as a motor disorder characterized by a velocity-dependent increase in tonic stretch reflexes with exaggerated tendon jerks[I]. For stroke patients, the spasticity changes gradually since the onset of a stroke and affects limbs functional movements generally. An objective method is highly desired to quantify the changes of spastic hypertonia after the stroke occurs, i.e. time-course analysis, thereby improving the treatment and rehabilitation process for spastic patients[2].

The common approach in previous studies is a constant velocity stretch in a completely relaxed limb. Various indices have been utilized to quantify the reflex torque induced during constant velocity stretch. among which include the threshold and the gain of stretch reflex[3,4]. However, according to other studies, the reflex threshold cannot be easily identified from the measured electromyographic signal [4]. Mechanically measuring the reflex torque is appealing since they likely reflect the contributions of all relevant muscles. However, acceleration inertial of the forearm and the rotation attachment of stretching device usually occurs extremely close to the time of a rising torque, thereby making the determination of reflex torque threshold a difficult task. This study presents a spasticity measurement system, capable of quantifying the time-course changes of spastic hypertonia in which a series of measurements are performed after a stroke occurs. Moreover, the feasibility of utilizing a spasticity index which measures the relative reflex torque rather than the absolute torque value is addressed.

METHODS

The details of on-line spasticity measurement system are described in our previous study [3]. In this study. the subjects were tested at supine position with the upper limb stretched toward the ground, which is different from the seated position with the horizontal stretch reflex used in previous studies. Five different stretch **velocities (i.e. 5, 20, 40, 60, and 80** deg/sec) in a randomly chosen sequence were performed once for a testing session. Among these velocities, the values measured at the velocity of 5 deg/sec were used as the baseline. Four subjects with CVA were recruited to verify the applicability of this spasticity measurement system. For the time-course study, the experiments were performed at 72 hours, one week, one month, three months. and six months after the onset of stroke. In the vertically stretched mode, the measured torque in Fig. l is a combination of gravity, stretch reflex, and acceleration/deceleration inertial torque. The dynamic torque, i.e., the ramp of torque between pl and p2, is mainly attributed to the measured spasticity and gravity but has minimum inertia and vibrational reactive torque. The integration of dynamic torque deviated from the baseline torque between pl and p2 divided by the accumulation angular displacement is then defined as averaged speed-dependent reflex torque (ASRT) which is used for quantifying the spastic hypertonia in this study:

$$ASRT_v = \frac{1}{p2 - p1}\int_{p1}^{p2} T_v(\theta) - T_5(\theta)d\theta$$

Where T_5 and T_v denote the torque values measured at 5 deg/sec and other stretch velocities, respectively. The slope of the linear regression line between the ASRTs and stretch velocities is then defined as either velocity sensitivity of ASRT or in short ASRT sensitivity. These two biomechanical parameters, ASRT and ASRT sensitivity, are used herein to quantify the spasticity.

RESULTS

Generally speaking, the larger area between the baseline torque and measured torque can be observed at the involved elbow joint and a consistently low ASRT can be found in the unaffected side of four stretch velocities. By pooling the ASRTs of five repetitive tests for four stretch velocities together. the ASRT sensitivity of both sides could be compared using comparison between the slopes of two regression **lines.** To observe the change of ASRT sensitivity during the spasticity, multiple comparisons (Turkey test) between ASRT sensitivity of two successive measuring times, e.g. 72 hours versus one week, were performed for each subject. As **figure** 2 indicates that no consistent trend appears in the ASRT sensitivity of four CVA subjects. According to this figure, ASRT sensitivity significantly increases before 1 month and, then, maintains or decreases at latter stages for subjects 1 and 3. In contrast, subject 2 has a delay in occurrence of peak ASRT sensitivity at 3 months; subject 4 has an earlier occurrence of peak ASRT sensitivity at 1 week.

DISCUSSION and CONCLUSION

Our approach alleviates the influence of acceleration/ deceleration inertia1 problems in measuring torque during constant stretching mode. In addition, adequate stretch angle of 75” incorporated with a quick response servomotor have minimized too small of a torque measurement window during stretching. a problem frequently encountered by previous studies [5]. According to the time-course analysis, ASRT sensitivity significantly increased before one month after onset, as shown in Fig. 2. Interestingly, the muscle tone significantly reduced in all subjects at the chronic stage (at 6 months). Similar observations have been shown that the maximum of spasticity is established between 1 and 3 months and a significant reduction occurs at the chronic stage [4]. In conclusion, this work has presented a spasticity measurement capable of performing constant velocity stretch of elbow flexors. The ultimate goal of this research is to utilize the spasticity measurement for assessing either rehabilitation progress or treatment of medication of spastic limbs.

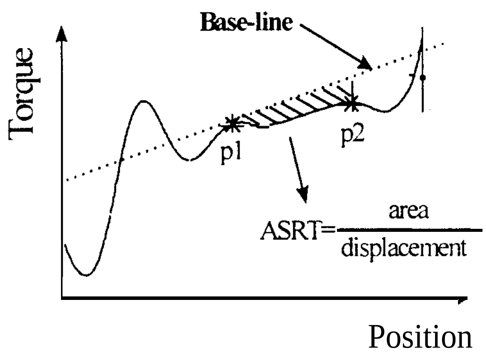

Fig.1. Definition of averaged speed-dependent reflex torque (ASRT).

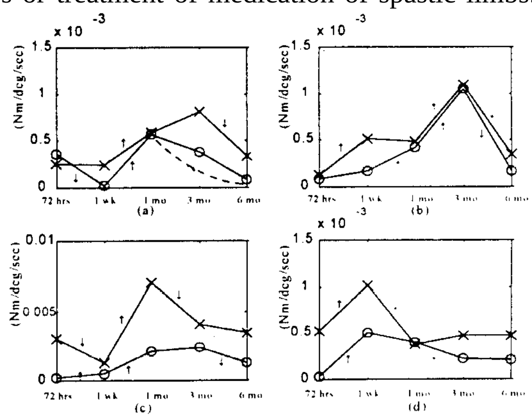

Fig.2. Time-course developments of spasticity in four hemiparetic subjects using ASRT sensitivity.

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Cerebellar Purkinje Cells Responses to Mechanical Stimulation of the Achille's Tendon in the Cat

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INTRODUCTION. It is well known, that a monosynaptic reflex (MR) in the gastrocnemius-soleus (GS) muscle can be evoked either by percussion of the Achilles tendon (T reflex) or by weak electrical stimulation of the popliteal nerve. Fluctuations in amplitude of the T-reflex provide a valuable method of testing the excitability of the motoneuronal pools. It was shown that in decerebrated cats single shocks of hindlimb area of the Cerebellar anterior lobe (IV-V lobulus) cause Potentiation of the ipsilateral monosynaptic reactions and this data suggest that Cerebellar conditioning stimulations is equivalent to an excitatory spino-Cerebellar volley induced by group Ia muscle afferents in motoneuronal pools. Moreover it is also known, that group II muscle afferents can easily induce excitatory or inhibitory responses of Cerebellar Purkinje cells (1,4). On the other hand it is believed that the fast tap of the Achilles tendon induced GS muscle stretch and such phenomenon provide the possibility to correlate the course of the segmental events with changes of discharge frequency of Cerebellar Purkinje cells (PC).The aim of this investigation was to study this correlation using electrophysiological approach to evaluate separate contribution of mossy and climbing fibers driven activity to Cerebellar PC in response to the GS muscle tap stimulation.

METHODS. Ten adult cats were tracheotomized and decerebrated at precollicular level under ether anesthesia. Small hole over the cerebellum then was made exposing vermal and intermedial parts of the anterior lobe (III-V lobulus) . Gastrocnemius-soleus muscle stretch was obtained by electromagnetic device to deliver brief and constant (l/sec,2-4 msec) Achilles tendon blows (5). Tendon reflexes was recorded electromiographycally and continuously monitored.The Cerebellar Purkinje cells in the ipsilateral vermal and paravermal hindlimb area was electrophysilogically identified by presence of simple spikes (SS) and complex spikes (CS) as well as inhibitory pause after PC by climbing fiber (CF) synaptic activation. The response of the Achilles tendon was controlled by bipolar EMG electrodes inserted into gastrocnemius-soleus muscle. In order to get optimal and submaximal monosynaptic response , slight changes of the tap location on the tendon was made in the site of application , the magnitude of the shock maintained as constant as it was possible. Evaluations of the excitatory or inhibitory effects of mossy fiber (MF) and climbing fibres afferents on Purkinje cells were obtained from peristimulus time histograms (PSTH).

RESULTS. It was revealed that out of 47 Purkinje cells 35 (75%) showed changes in their mossy fiber driven activity. Out of this 35 PC ,12 were excited and 12 displayed mixture of excitatory and inhibitory effects, whereas 11 showed solely inhibitory effect on single stimulation of the Achilles tendon (5). The level of spontaneous firing was determined from initial 100 msec of the PSTH, before stimulation onset. Usually, a change in the MF driven activity was considered as "excitation" when the SS firing rate was at least overshoot of 50% the average spontaneous firing level, and as "inhibition" when the SS tiring decreased in the same range. As rule the Achilles tendon tap followed by inhibitory action on the MF driven activity. It is known that the CF activation of PC exert a suppressive action on the SS discharge. In the present study, pauses in SS firing were frequently observed following spontaneous or evoked complex spikes. In half of PC which showed an high CF response probability, the depressions of SS activity could be completely of mainly attributed to the post-CF pauses. These inhibitions on the MF driven activity appeared just after the evoked CS and, therefore, their latencies were usually in range of 20-30 msec. Mean pause ranged from 50 up to 300 msec., roughly the same order pause we observed after full maturation of stato-kinetic reflexes in kittens and guinea pigs (2). In some instances, each spontaneous CS was immediately followed by high frequency burst of SS. For small number of PC, this rebound phenomenon - the post CF burst - was obtained in response to tendon tap. This kind of PC were belonged generally to group "A" and displayed a very low level of spontaneous SS firing , so that effects of tapping on the SS activity was only expressed by a late excitation (20-40 msec after the evoked CS) with a duration of increases in SS discharge ranging from 20 - 200 msec. The main results concerning of the temporal relation between the peripheral events and the PC responses induced by tendon tapping are: first, the earliest excitatory and mainly inhibitory actions of tapping (lo-30 msec) are induced via MF afferent input, these actions start with the beginning of the silent period during muscle stretch; second, for the 70 msec period following stimulation, the inhibitory effects prevail over the excitatory owing to the phenomenon of post - CF pauses. The inhibitory effects on SS activity are concomitant with rising phase of tension in the muscle and the first part of the silent period of muscle; third, the most important excitatory effects (50-170 msec) are mainly correlated with the peak of the contraction, the phase of muscle relaxation . Out of 61 PC 39 (65%) were climbing fiber (CF) responsive to the tap stimulation. This cells can be divided onto two groups. To the group "A" belonged PC which displayed a high probability (P> 0.5), a short CF response latency (20-35 ms) and low variability in latency time of each CF response (lo-30 ms). To the second group "B" belonged PC with low probability (P< 0.5), long and variable latency time (35-60 ms).The evoked CS activity of group "A" cells occured quite at the beginning of the twitch, whereas the CS activity of the group "B" cells only during the late phase of contraction.

DISC USSION Thus the received data suggest that various effects ef the Achilles tendon mechanical stimulation by taps on the Cerebellar PC responses may be regarded as consequence of the diversity of the spino-(olive)-Cerebellar pathways involved in conductivity of the peripheral information. We think that good sensitivity of the CF input is attributable to the conjunction of excitatory effects at once of muscle spindle afferents (Ia + II) induced by the brief tendon stretch, and of Golgi tendon organ affrents induced by reflex contraction (3). It may be assume that Ia and II muscle afferents were probably contributing to short latency of MF and CF responses of the Purkinje cells of the group "A", whereas Ib group muscle afferents were apparently involved in the late MF and CF responses belonged to the second "B" group of the Purkinje cells. Moreover it is possible also to assume that peripheral information which reached of the Cerebellar Purkinje cells used by long-loop of the "myoelectrick feedback" for the control of "muscle relaxation stream" (Biofeedback: principles and practice to clinicians, Ed. by John V.Basmajian, 3ed., Baltimore, 1989).

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Neuromuscular Control During Gymnastic Landings

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Neuromuscular control during gymnastic landings

Introduction: Lower extremity injuries are a common occurrence for athletes participating in gymnastics. Brueggeman (1996) reported that 55% of international level gymnasts complained of various injuries to the lower extremities. Many of these injuries occur during landing after dismounts from gymnastic apparatus. For this reason it is important to further describe the muscular control of the lower extremities during landing. In addition the effect of various muscular loads on specific muscles and their effect on foot stability during landing will also be examined.

Methods: In this experiment 6 male subjects performed a series of landings for a height of 105 cm onto a 5 cm strong Sanaige-Mat. The landings were performed before and after 3 different treatments. These were an isometric contraction of the upper leg and the calf muscles (A and B respectively) and a dynamic loading of the calf muscles (C). During the landings electromyographic activity of the muscles v. medialis, v. lateralis, biceps femoris, tibialis anterior, peronaeus longus, gastrocnemius medialis and soleus of the left leg was recorded using the Ernst-System. Pressure distribution under the left foot was recorded using PEDAR-system insoles. Kinematic data was collected using a side view video camera (50 Hz) and a high-speed video camera (200 Hz) positioned from a rear view.

Summary of the results: Examining all of the performed jumps it was found that the ankle angle change (25±5°) during the landing was less than that of the knee joint (79±10°). The greater stiffness observed in the ankle joint in relationship to the knee joint is probably caused in part by the activity of the muscles found at those joints. In particular the tibialis anterior and the soleus showed amplitude values of 115±37 and 79±20 %-MVC (maximum voluntary contraction). The upper leg muscles showed a mean amplitude of 46±18%-MVC. All of the muscles showed an increased innervation during the pre-activation phase before ground contakt. The highest pre-activation amplitudes were found in the soleus, peronaeus and gastrocnemius with values of 22±88, 205±32, and 195±61 %-MVC respectively. The high innervation of the ankle joint muscles allowed the reduction of the (passive) impact forces during the landing. Compared to the forefoot the touch down of the heel was delayed 27±25 ms. An earlier time of ground contact was measured for the lateral part of the foot (25±7 ms) with respect to the medial par-t of the foot. The lower stiffness of the knee and it's moderate joint movement allowed a controlled deceleration of the body mass.
After completion of the treatment trials the activation patterns during landing and the foot stability were different from the normal values without treatments. The reason for the increased amplitudes of the knee extensors (v. medialis 13±32, v. lateralis 7±18 %-MVC) after treatment A seemed to be due to the adaptation of the muscle spindles during the treatment and therefor resulted in increased sensitivity during the landings. Because of the increased activation less time was needed for the upper leg muscles to optimally stabilize the knee joint. The contraction time of the v. medialis and lateralis until maximum knee flexion was shortened by 291±150 and 267±537 ms respectively. The pre-activation

levels of the peronaeus and gastrocnemius were shortened by 52±24 and 49±38 ms respectively. AI1 of the examined muscles of the lower leg showed a decrease in activation levels particularly the soleus and tibialis anterior whose amplitudes decreased by 14±30 and 1 1±29 %-MVC. The lessened stability of the foot resulted in 22±25 ms sooner heel ground contact and an increased force maximum of 75±256 N. The active muscular absorption during the landing was clearly less than during the non-treatment landings.
After treatment B the lower leg muscles showed higher amplitudes (tibialis anterior 10±58, peronaeus 4±16, soleus 14±28, gastrocnemius 2±15 %-MVC) possibly also due to adaptation of the muscle spindles. This seemed to be the cause of increased foot stabilization of dorsal extension (-5±5°) as well as pronation (-3±5°). The time of maximal pronation was delayed by 61±76 ms. The tension of the muscles was great enough that just after the heels touch down during the impact phase they were again raised from the ground. The force-time-integral was reduced by 70±108 Ns. Because of the smaller contact surface when the heels were lifted the forefoot force-time-integral increased by 22±126 Ns. The stiffer ankle in comparison to the non-treatment trials led to a reduction of the ground reaction force under the heel of 174±304 N.
After treatment C an increase in the mean power frequency (v. medialis 19±37, v. lateralis 10±24, biceps femoris 16±30, tibialis anterior 10±3 1, peronaeus 8±22, soleus1 0±26 and gastrocnemius 8±29 %MVC) as well as an increase in amplitude (v. medialis 20±48, v. lateralis 10±26, biceps femoris 4±14, tibialis anterior 44±1 16, peronaeus 1±25, soleus24±70 and gastrocnemius 9±27 %MVC) were seen in all muscles measured. Both findings are probably a result of the recruitment of faster fibers (cp. VOLLESTAND et al. 1984, MORITANI et al. 1986) because of the early fatigue of the slower fibers (cp. GOLNICK et al., 1972, BRUGGEMANN 1996). Fatigue resulted in greatly decreased foot stability. In comparison with the non-treatment trials the treatment C trials resulted in 13±25 ms earlier heel contact and a maximum force increase of 288±53 1 N. Like after treatment A the active muscular absorption during the landing was clearly less than during the non-treatment landings.

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Foot Force During Constrained Motion Exercise

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INTRODUCTION

The force that the environment exerts on the terminal segment of the limb during multi-joint movement (e.g. the force of the floor on the foot in walking) can be quantified by its magnitude (F) and direction (0). These quantities affect limb loads in distinct ways: F directly affects the load induced on each limb structure, and θ affects the *relative* distribution of joint torques within the limb. Small θ changes may produce significant joint torque re-distributions and thus modify the *relative* loads in anatomical structures. This can in turn greatly affect relative tissue loading. An application of this mechanical concept can be used in rehabilitation strategies in order to prescribe optimal loads on both healthy and recovering structures.

Constrained motion exercise, where the terminal segment is constrained to move along a prescribed path such as in cycling, has been shown to elicit stereotyped patterns of force direction. Since some constrained motion exercise devices are frequently used in the rehabilitation setting (l-2), it is important to measure their interaction F and θ, as these influence loading of anatomical structures. The goal of this study was to measure the interaction force magnitude and direction at the foot-pedal interface during exercise at varying workloads on a novel constrained motion exercise machine.

METHODS

An exercise machine was instrumented to allow measurements of interaction force magnitude and direction at the foot-pedal interface. The Cross-Trainer @e-market prototype, Life Fitness Inc., Franklin Park, IL) is a weight-bearing exercise machine designed with foot pedals and arm levers (Figure 1). Pedals restrict foot trajectory to an approximately oval path with similar dimensions to the walking foot path with respect to the hip. The pedals are driven through the application of forces by the feet, with exercise speed determined by the balance of forces applied by the exerciser and the motion resistance provided by the machine. Feet are not attached but remain in contact with the pedal at all times and are prevented from sliding forward by vertical guards. The results presented here had the subjects resting their hands on the stationary hand rests.

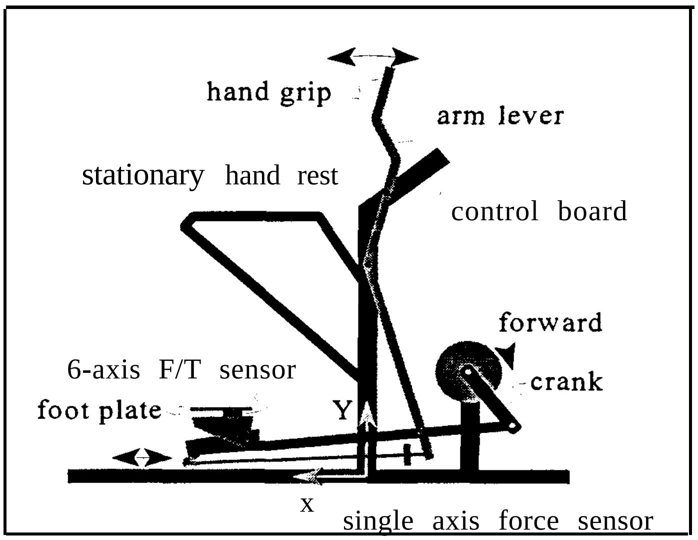

Figure 1: Cross-Trainer

Data was collected from nine healthy subjects during three-minute bouts of constant speed (55 rpm) exercise against three workloads (76, 124, and 196 W), presented in random order. A potentiometer measured crank angular position throughout motion. Sagittal plane right foot forces were measured using a force/torque sensor (Assurance Technologies Inc., Garner, N.C.). Ten seconds of crank position and foot force data were collected every 0.03 crank angle radians, sampled at 200 Hz, and recorded on a computer hard drive during the last minute of each exercise bout. Data for each pedal position were averaged over five consecutive crank cycles for each subject at each workload. Force data at four discrete locations of the right pedal were analyzed: Bottom-Dead-Center (BDC), the Most Anterior Point along the pedal trajectory (MAP), Top-Dead-Center (TDC), and the Most Posterior Point along the pedal trajectory (MPP). Instantaneous mechanical power (IMP) delivered to the pedal by the right foot was calculated for the four pedal locations.

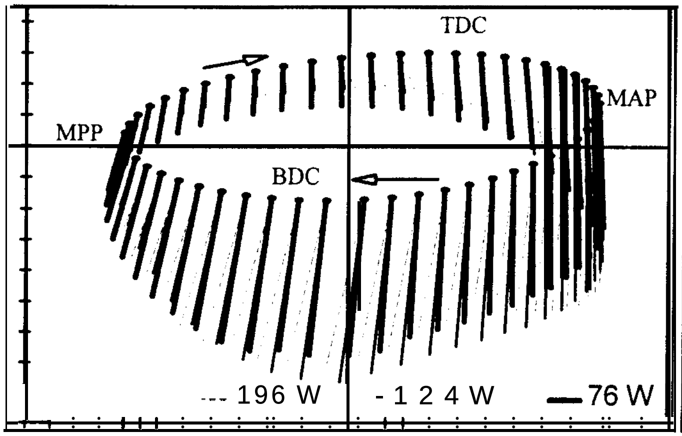

Figure 2: Right foot force (axis marks = 100N)

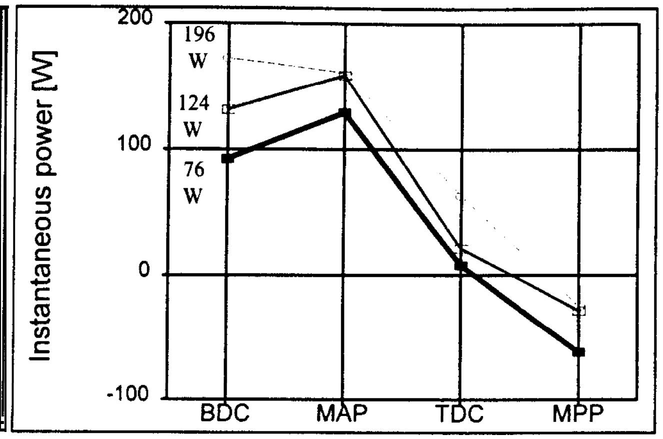

Figure 3: Instantaneous power (right foot)

RESULTS

Figure 2 displays sagittal plane force vectors emanating from various pedal positions along the right pedal path (dots indicate ball of foot location), with motion sense as shown by arrows. The figure shows a relationship between θ and workload at TDC and BDC only (p < O.Ol), and a relationship between F and workload at MAP and MPP only (p < 0.01). Both of these force vector adjustments contribute to increasing the amount force applied in the direction of the pedal travel, and thus contribute to increasing work done on the system. Figure 3 displays changes in instantaneous mechanical power of the right foot, relative to workload increases, at the four pedal locations, and shows that when workload is increased from 76 to 124 W, IMP increases are mostly seen at BDC, MPP and MAP, whereas from 124 to 196 W, IMP increases are mostly seen at BDC and TDC. Further analysis of IMP shows that the predominant strategies used to increase IMP are F increases from 76 to 124 W, and θ modifications from 124 to 196 W.

DISCUSSION

When comparing these results to data from the seated cycling literature (3), we see that cycling forces show a wider range of force direction. This can be explained by the fact that support of the body mass has to be actively maintained during Cross-Trainer exercise, and therefore a considerable part of the pedal reaction force has to counteract gravity. In contrast, since the seat offers an important stability component during cycling, the body is less constrained to actively maintain balance during cycling, and it has greater freedom to redirect the interaction force along a direction that will enhance performance. However, cycling force magnitude only (and not direction) seems to be influenced by changes in workload. This implies that although loads induced on the limb will be higher with increased workload, *relative* loads will remain constant.

This study shows that significant force direction changes occur during exercise against varied workload on the Cross-Trainer. In turn, these may create a range of relative loading patterns within the limb, which could be exploited in rehabilitation strategies. An in depth analysis of muscle activation levels during exercise on the Cross-Trainer, which was collected during this protocol but not reported here, will help in understanding the consequence of these relative joint torque re-distributions on muscle activity. Further steps would be to evaluate the influence of other motion cycles on relative joint torques.

This research was partially supported by a grant from Life Fitness, Franklin Park, IL.

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Influence of Complexity of Final Equilibrium on the APA Associated with Vertical Jump

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INTRODUCTION

The adequate performance of most voluntary movements must be accompanied by postural adjustments. The postural adjustments which occur at the same time as, or prior to the onset of a movement are called anticipatory postural adjustments (APA). APAs are used to counterbalance the perturbation created by several limb and trunk movements (3,5,9). In contrast, in gait initiation, APAs create the conditions necessary for progression by initiating a forward fall (4,7). In a precedent study (6) we demonstrated that a single voluntary vertical jump is associated with APAs similar to those described for many forward-oriented movements. Vertical jumps obviously imply an initial postural disturbance and are upward oriented. This kind of task requires the subjects to return to their initial position to stabilize it in a single jump, or to rebound in repetitive jumps. This study was therefore carried out to determine how the APAs associated with a vertical jump depend on the complexity of the task and the final equilibrium.

MATERIALS and METHODS

The experiments were performed on 5 male subjects, who were not experts in the task studied. Two kinds of task were tested: a series of single jumps and series of 3 repetitive jumps (hopping). The series of single jumps or hopping tasks were always separated by a few seconds of rest. Subjects had to jump to different heights, keeping their hands clasped behind their back and their head and trunk as vertical as possible, their knees flexed at 30" with respect to the full extension. Vertical jump movements were initiated from a semi-squat position to avoid countermovements (2) and subjects were told to land at about the same piace on the piariurm. The subjects stood barefoot on two force platforms which the reaction forces under each foot to be recorded in three orthogonal directions and the corresponding moments calculated as well as the displacement of the center of pressure (CP) under each foot, The jump vertical impulse was obtain by integrating the vertical reaction force (Fz) over the duration of the vertical acceleration phase of the body Surface EMG of the Soleus (SO), Tibialis anterior (TA), Biceps femoris, and Vastus lateralis muscles were recorded from both sides of the body. Ail signals were sampled at 1000 Hz.

RESULTS

Heel-off was preceded by a Soleus EMG burst, the onset of which was taken as the onset of voluntary movement (t0). This time was taken as time zero (t0) for all measurements. A backward shift of the CP occurred 100-500 ms before t0 in both single jump and hopping. The amplitude of the CP shift was greater in the hopping task (4.1 ± 0.9 cm) than in the single jump (3.6 ± 11 cm). The CP shift occurred earlier in hopping (287 ± 102 ms) than in the single jump (241 ± 82 ms). It was always preceded and accompanied by a TA excitation which followed a SO deactivation Bursts of Biceps femoris and Vastus lateralis EMG occurred shortly before or after t0. The latency of TA excitation did not differ under the two experimental conditions, but the SO deactivation appeared earlier in the single jump (467 ± 145 ms) than in hopping (368 ± 99 ms). Hence, the interval TA-SO was shorter in hopping. The latency of SO deactivation and TA activation were linearly correlated for each kind of vertical jump. These latencies were also correlated with the onset of the CP shift. The APA depended partly on the jump amplitude in the two jump conditions: the onsets of the CP shift and EMG changes occurred earlier for higher jumps. In contrast, the amplitude of the CP shift did not depend on this parameter.

DISCUSSION

The above data show that the two kinds of voluntary vertical jumps are associated with APAs, which have two features in common: i) a backward CP shift whose onset depends on the jump amplitude, and, ii) a pattern of SO deactivation - TA activation whose timing also depends on the jump amplitude. Like gait initiation, APAs associated with vertical jump create the disequilibrium needed to perform the movement.

It has been suggested that the duration of the APA may depend on the complexity of dynamic postural equilibrium in other tasks. For example, Dietrich et al. (7) studied single step and multistep initiation and found that the relationship between the peak velocity of the center of gravity and the duration of the APA in single and multistep initiation were similar. The APA associated with multistep initiation were longer because the peak velocity was greater and the APA are correlated with the velocity of the forthcoming movement. Nevertheless, they interpreted their results in terms of the complexity of the control of dynamic equilibrium. The difference in amplitudes and CP shift times between single jump and hopping reinforces this idea, since we compared single and repetitive jumps of similar amplitudes. This is also in agreement with Do et al. (8), who found that the APA in a leg flexion-extension task were longer with a unipodal man possible which equilibrium to be ;ess stable

In a previous study (6) we stressed that SO-TA synergy was likely to be a functional synergy used by the central nervous system to simplify the postural motor command. As discussed by Bernstein (1) this synergy may depend on reflex loops or on upper nervous system programs Crenna and Frigo (5) have shown that the two components of SO-T.4 synergy might act in uncoupling manner. They showed that SO deactivation consistently appeared with slow movements in gait initiation and with the initial forward leaning posture, while TA activity was reduced or suppressed under the same experimental conditions. These authors found that the modulation on the CP shift by the **Soleus** was quite slow. TA produced a fast modulation of the controlled variable: the location of CP. We find that the SO-TA time interval was shorter in hopping than in single jump. Thus, the data suggest that this functional synergy is the result of a central program rather than a reflex linkage.

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Midsole-Surface Influence on Muscle Activation and Impact Shock

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INTRODUCTION

Midsole and surface properties independently influence the impact shock in magnitude and frequency in running and walking (Light *et al.*, 1980; Nigg *et al.*, 1983; Shorten *et al.*, 1986). In addition, muscle activation patterns (EMG) may change as the control system adapts to provide the necessary shock attenuation. The purpose of this study was to investigate the influence of midsole hardness and surface stiffness on impact shock and EMG measures during running.

METHODS

Six male fitness runners, free from lower extremity injury, volunteered to participate in this study. The subjects ran at 3.4 $m \cdot s^{-1}$ for six minutes in each of six conditions (C 1 -C6, *Table I).* Condition order was balanced across subjects.

Table I. Six shoe and surface conditions.

	Cl	C2	C3	C4	C5	C6
Shoe'	S	S	S	H	H	H
Surface'	S	M	H	S	M	H

[1] S - soft (40 Shore A), H - hard (70 Shore A)
[2] S=lOO kN/m; M =200 kN/m; H=350kN/m.

Leg and head shock were measured with two 1.7 gram accelerometers (PCB) mounted on the anteromedial distal tibia and forehead.' The accelerometers were interfaced to a microcomputer via an A/D converter and sampled at 1 kHz. A 10 trial mean value was calculated for peak leg (PL) and head (PH) acceleration and transmission (PH/PL). The power spectral density (PSD) was obtained through an FFT of 10 trials per condition (Press *et al., 1989).* A transfer function (TF) was calculated using the formula:

$$TF = 10 \log_{10}(PSD_{head}/PSD_{tibia}).$$

EMG data were collected at 1 kHz from six muscles in the lower extremity, gluteus maximus (GM), biceps femoris (BF), rectus femoris (RF), Vastus lateralis (VL), tibialis anterior (TA) and gastrocnemius (GA). Linear envelopes of EMG data were created through full-wave rectifying and low-pass filtering (22 Hz, zero-lag, 4th order Butterworth). Ensemble averages were calculated from ten strides and EMG onsets and offsets for each muscle were determined at 10% of the peak EMG for each condition.

A two factor repeated measures ANOVA was conducted on accelerometry and EMG variables ($p < 0.05$).

RESULTS AND DISCUSSION

Shock

The hard surface generated the greatest peak leg and head shock. It was twice as great in the leg on the hard versus the soft surface while head shock tripled. Others have shown that the surface can be effective in attenuating external impact force (McMahon and Greene, 1979).

Shock frequencies were separated by the frequency analysis into active (4-9 Hz), impact (lo-20 Hz), and resonant frequencies (>40 Hz) similar to Shorten and Winslow (1992). Leg and head spectra power were greatest on the hard surface at impact frequencies (lo-20 Hz) regardless of midsole. Peak impact frequencies in the leg were nearly fivefold higher in power on the hard versus the soft surface, but the range remained similar.

Head shock frequencies increased in power within the same range (lo-20 Hz). However, increases were minimal compared to the leg. The head was thus appropriately guarded from impact shock vibration.

Shock attenuation

Shock attenuation of impact frequencies (IO-20 Hz) entering the body was greater when the surface was stifler *(Fig. 2).* This provides evidence for a musculoskeletal attenuation mechanism between the tibia and head, protecting the visual and Vestibular systems. Others have found that surface did not influence shock attenuation (Lafortune et al., 1996). This may have been due to use of a pendulum to impact supine subjects and point compliant surfaces.

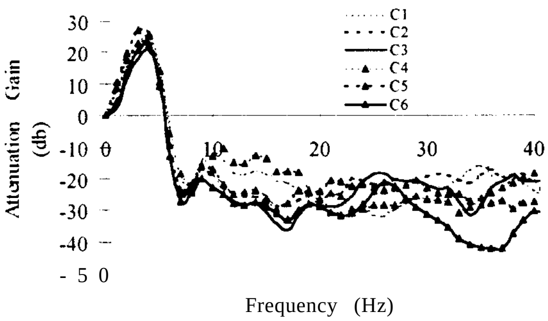

Fig. 1. Transfer *function in six shoe and surface combinations averaged over six subjects.*

Surface-dependent attenuation may signify that the system cushions passively, in a nonlinear manner, resulting in amplitude dependent shock attenuation. An alternative explanation is that muscle coordination is changing to provide more attenuation with a harder surface.

Muscle *activation patterns*

Muscle activation patterns appeared to be influenced by impact shock. Specifically, BF and TA onset times occurred earlier in the swing phase as surface stiffness increased *(Figs.* 2 *and 3).* These muscles were activated in preparation for footstrike, at 100%. Other timing variables did not show significant differences.

The modification in the BF onset is significant because this muscle has been proposed to have a large role in muscular cushioning. Hamstring forces at foot strike can be as large as body weight which provides potential for cushioning to the knee and hip joints due to rapid muscle shortening during the impact phase (Cole et al., 1996).

Early onset of the TA before footstrike may be related to control of plantarflexion_ Running on the hard surface produced greater shock suggesting a higher ankle moment may have been created. In this situation, greater control would be necessary to prevent high impact forces in the forefoot.

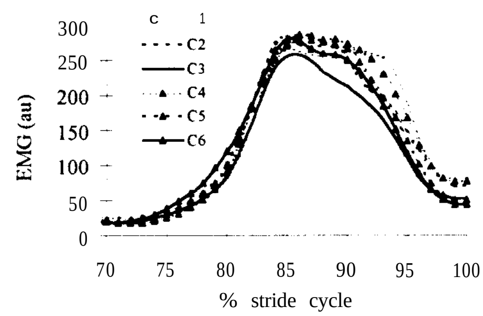

Fig. 2.Biceps femoris (BF) muscle activation in six shoe and surface combinations averaged over six subjects.

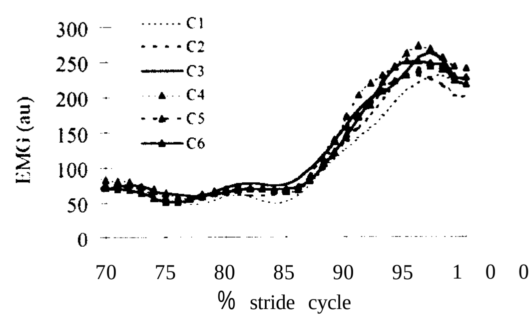

Fig. 3. Tibialis anterior (TA) muscle activation in six shoe and surface combinations averaged over six subjects.

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Gastrocnemius and Vastus Lateralis Actions During Jumping in 1G and Simulated Microgravity

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INTRODUCTION

Exercise protocols have been developed to counteract the adverce reactions to space travel; however, to date no modality has been adequate in preventing the problems of space-flight induced osteoporosis and muscle atrophy. In the case of bone loss, exercises which deliver high dynamic loads to the lower extremities (such as running or jumping) are thought to stimulate osteogenesis by providing strain and strain rates that significantly contribute to skeletal homeostasis. Muscles that are particularly prone to the adverse effects of microgravity are the leg extensors, and decreases in strength of 25% have been reported. It has been proposed (Convertino, 1990) that these losses may be due to a lack of eccentric muscle actions during orbital missions. In light of this, our working hypothesis is that jumping exercises may not only benefit bone density, but may also help alleviate the loss of muscle strength in microgravity. This in turn is based on the fact that leg extensors typically experience considerable eccentric activity during the landing phase of jumping. Thus the purpose of the present study was to investigate whether eccentric muscle activity in the Gastrocnemius and Vastus Lateralis muscles during jumping exercises in simulated microgravity was comparable to the 1G situation.

METHODS

A zero gravity simulator (ZGS) was constructed using ten foot long latex cord and rope to suspend subjects from the ceiling in a supine position (D'Andrea, 1997). The cords were attached to the center of gravity of the lower extremity segments. Body segment masses were calculated using regression equations and anthropometric measurements to assure that the tension in the cord matched the weight of each segment (Vaughan et al., 1992). Rope also supported the subject at chest and waist harnesses. A gravity replacement system consisting of two steel springs, attached at the waist in front and back. was used to tether the subject to the wall (thereby recreating the sense of "weight"). The springs were tensioned in order to provide forces equal to 45,60, 75 and 100% of the subject's body weight. Twelve subjects (six females and six males, ages 30-48) performed a countermovement jump in the zero gravity simulator with their feet landing on a wall mounted force plate Markers were placed on the greater trochanter, lateral femoral epicondyle, lateral malleolus, posterior aspect of heel, and above the second metatarsal head and 3-D kinematic data were recorded at 60 Hz using four cameras (Motion Analysis Corp). EMG activity was recorded at 2400 Hz using bipolar electrodes spaced 1.5 cm apart and placed on the belly of each muscle of interest.

Post processing involved (i) using the forceplate data to determine lift-off and landing times, (ii) passing the kinematic data through a low pass Butterworth recursive filter (cut-off 9 Hz) and calculating sagittal plane knee and ankle angles, (iii) using the angle data to determine muscle lengths, and times when the muscles were lengthening, shortening or isometric, and (iv) obtaining a linear envelope for the EMG data and determining the times when the muscle was active and quiescent. The logic used to combine the resulting EMG "on-off' times with the muscle length data to determine concentric, eccentric, isometric and quiescent durations has been previously reported (Jiang, Davis and Cavanagh, 1992).

RESULTS

During the push off phase, the amount of concentric activity exhibited by Vastus Lateralis was significantly higher ($p < 0.001$) for jumping exercises performed in the ZGS (mean 0.15 s, s.d. 0.004 s) compared to the 1G situation (mean 0.12 s, s.d. 0.004 s). The differences caused by the various spring tensions in the ZGS were minimal (duration of concentric activity at a tension of 100% BW was only 0.01 s higher than at the 45% BW setting). Gastrocnemius concentric activity was similar across both spring tensions and gravity situation (mean 0.11 s. s.d., 0.004 s).

Eccentric activity during the landing phase was significantly higher for jumping in 1G for both Gastrocnemius and Vastus Lateralis (Gastrocnemius in IG: mean 0.1 s, s.d., 0.004; in ZGS, mean 0.06 s, sd. 0.004 s; Vastus Lateralis in 1G: 0.14 s, s.d. 0.003 s; in ZGS, mean 0.11, s.d., 0.003 s). During the landing phase the ground reaction forces were 10% higher for exercises performed in 1G (2,464 N per foot) than for exercise in the ZGS (across spring tensions, forces ranged from 2,245 to 2,291 N).

DISCUSSION

This investigation reveals the first attempts to quantity concentric and eccentric muscle actions during jumping exercises in 1G and in simulated microgravity. Gastrocnemius and Vastus Lateralis were chosen due to the fact that they are known to lose significant strength following long duration orbital missions (Thornton and Rummel, 1977). Furthermore, emphasis was placed on the amount of concentric and eccentric activity during an exercise which is thought to be a candidate for counteracting spaceflight-induced bone loss.

The finding that both ground reaction forces and eccentric activity were significantly reduced during ZGS exercises, even at the highest spring tension levels, indicates that it is difficult to design exercises that mimic the 1G situation. Possible reasons for the decreased eccentric activity in Gastrocnemius and Vastus Lateralis include (i) the fact that the impact landing force was higher in lG, and (ii) differences between a uniform gravitational force (in IG) and a force caused by springs which is dependent on their length (and thus highly affected by knee angle at landing).

In conclusion, this study has shown that jumping exercises performed in a zero-gravity simulator result in less limb loading and eccentric activity in key “anti-gravity” muscles of the lower extremity. Future work includes the possibility of studying constant-force springs and their affect on the mechanics of jumping exercises in the ZGS; validation that the suspension-type simulator recreates microgravity conditions; and investigations into physiological thresholds that need to be exceeded in order to prevent bone loss and muscle deconditioning in microgravity

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Electromyographic Analysis of Selected Muscles During Rope Skipping

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Introduction

Rope skipping is one of exercise that can be performed indoors, which required less skill and easy to learn. So people may choose rope skipping as exercise mode for fitness training. A daily 10 minutes workout of rope skipping was as good as 30 minutes of jogging a day (Tibor et al. 1986). A few study about the energy expenditure of rope skipping by using the measuring of VO_{2max} (Glenn et al. 1980), the intensity of rope skipping (William et al. 1981). Not much biomechanical study on rope skipping, John et al. (1986) described the biomechanics of rope skipping. Lack of study about motion analysis or EMG signal analysis in rope skipping. The purpose of this study was to determine if differences in EMG signal of the selected muscle occurred between trained and untrained rope skipper.

Method

20 secondary school students (male) were participating in this study. The subject's age was 15.65 ± 0.59 years, body height was 172.51 ± 5.36 cm and body weight was 57.80 ± 6.51 kg. All subjects were healthy and without any symptoms from the musculoskeletal system. Ten subjects were the trained rope skipper with training period for six weeks and others were considered as untrained group.

Surface electrodes (silver / silver chloride, ARBO Medizin-Technologie, Germany) attach to the selected muscles in a standardized manner; in the direction of the muscle fibres, with an inter-electrode distance of 3 cm. Before attaching the electrodes, the skin will shave, and rub with alcohol in order to lower the skin resistance. The selected muscles were vastus medialis, rectus femoris, vatus lateralis, tibialis anterior, semitendinosus, biceps femoris and gastrocnemius.

After attaching the electrodes, the electrode cable will attach to the electrodes in described sequence before. The electrode cable connected with the pre-amplified which close to the pads, permits to eliminate completely the artefacts caused by patient movements. The pre-amplifier will be fixed with the paper adhesive tearable tape (Albupore, Smith+Nephew) on the skin to prevent the vibration of amplifier.

The signals form the electrodes will be pre-amplified, and transmit using telemetric radio transmitters (915 Transmitter Unit, TELEMG, Italy). Those signals will be received by the receiving unit (920 Diversity Data Receiver, TELEMG), and then pass through the optical fibre to the main unit. The main unit will then amplify the signal by 1000.

Full-wave rectified, low-pass filtered (600 Hz) and high-pass filtered (10 Hz) were recorded. All signals were simultaneously displayed on an oscilloscope(KENWOOD CS-5260) connected in parallel. This was used particularly for testing electrode function.

Subject was then performing rope skipping for ten minutes with standardized style and intensity. Two-leg form was the common style discovered during pilot study and the most comfortable intensity was 96 turns per minute. EMG signal was collected at first minute, fifth minute and ninth minute.

The collected EMG signal was full-wave rectified. The time duration for each muscles contraction was calculated, the sequence of the muscle contraction and the timing of contraction. All data were tested with t-test to see any differences between trained and untrained group in muscle co-ordination. EMG signal analysis is starting at landing position and then jumping.

Result

T-test showed the significant differences in the duration of vastus medialis contraction between the trained and untrained rope skipper with $p < .001$. The contraction timing between vastus medialis and tibialis anterior showed significant differences between different groups with $p < .001$. There is a significant differences in contraction timing between muscles rectus femoris and tibialis anterior in different groups with $p < .01$. The significant differences in contraction timing between muscles vatus lateralis and tibialis anterior between two groups with $p < .01$. Muscle gastrocnemius and tibialis anterior shows significant differences in contraction timing between two groups with $p< .01$.

Discussion

Under controlling the frequency and style of rope skipping, EMG signal were collected. The time duration of muscle contractions were compared between trained and untrained rope skipper. Moreover, the timing of muscle contraction were calculated and used to compare if differences between two groups of rope skipper.

The muscle Vastus medialis contraction have shorter duration in the group of trained rope skipper. The shorter in duration of EMG signal can reflect the working time of muscle is shorter. When compare with same intensity of exercise, shorter working time of muscle results in less in energy expenditure.

Co-ordination of muscle for exercise is necessary, the better in co-ordination of muscle group results in better exercise performance.

EMG signal was calculated starting at the moment of landing and then jumping. During landing, tibialis anterior starting eccentric contraction. When starting of jumping, gastrocnemius and quadriceps(vastus medialis, rectus femoris and vatus lateralis) undergo concentric contraction (John et al. 1986).

Under the same frequency of skipping, the significant differences in muscle contraction timing between trained and untrained rope skipper. The trained rope skipper have a bigger time gap of muscle contraction between quadriceps(vastus medialis, rectus femoris and vatus lateralis) and tibialis anterior when jumping up. The contraction timing between tibialis anterior (landing) and gastrocnemius (jumping) have a bigger time gap in trained group.

Combining the result of duration of muscle contraction and the timing of contraction under same skipping frequency. The shorter in muscle contraction duration and large time gap between different muscle contraction reflect more resting time to the muscle, as a result in less energy expenditure. This phenomenon can find in the group of trained rope skipper. This can reflect the different in co-ordination ability in lower limb muscle and working efficiency in trained rope skipper.

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Time-frequency Methods Applied to Muscle Fatigue Assessment During Dynamic Contractions

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INTRODUCTION

In the last two decades, surface myoelectric signal analysis has been proved effective for assessing the electrical manifestations of localized muscle fatigue, that in turns provide an estimate of muscle impairment in different neurological, muscular, and orthopedic disorders.

During a sustained contraction, the progression of localized muscle fatigue causes the power spectral density function of the signal to be scaled towards the lower frequencies. It is generally accepted that the mean or median frequency of the power spectrum density function effectively track the spectral scaling due to fatigue. In the past, the analysis of the electrical manifestations of muscle fatigue has been restricted to isometric, constant force contractions. In fact, traditional spectral estimation techniques require the signal to be wide sense stationary within the observation window. When muscle contractions depart from this constraint, as during dynamic contractions, the stationarity of the myoelectric signal can no longer be assumed.

Although there are numerous situations in which isometric and constant force contractions are fully satisfactory for providing a precise quantitative evaluation of muscle function, this contraction paradigm is rather uncommon in most daily activities. Physicians and physical therapists involved in rehabilitation, occupational medicine, and sport medicine are very interested in studying cyclic contractions, since these are prevalent in a variety of every day actions.

SURFACE MYOELECTRIC SIGNAL IN DYNAMIC CONTRACTIONS

The surface myoelectric signal detected during repetitive non-isometric contractions usually consists of relatively short bursts of signal whose statistical moments change remarkably over time, within each single burst as well as from burst to burst. Changes occurring within a specific burst are mainly related to changes in muscle force, muscle length, and displacement between active muscle fibers and the detection probe, while changes occurring from burst to burst are primarily related to the progression of muscle fatigue. The assessment of muscle fatigue during cyclic non-isometric contractions requires: a) the choice of a spectral estimation technique suitable to processing nonstationary signals, b) the availability of spectral variables that effectively describe spectral changes due to fatigue, and c) the capability of rejecting spectral changes unrelated to the fatigue process.

Spectral estimation technique - Among different possible approaches to the spectral analysis of nonstationary signals, time-frequency distributions belonging to the Cohen class recently received considerable attention. At this time, the Choi-Williams transform is considered as the most suitable to the analysis of myoelectric signals. The time-frequency spectrum of a stochastic process may be defined as an extension of the time-frequency distribution of deterministic signals, although the extension of the kernel dependent properties is not trivial. Moreover, to track the electrical manifestations of muscle fatigue we need to increase the estimation stability by adopting measures that do not compromise unacceptably those properties that are particularly desirable in this specific application, such as time and frequency shift invariance, frequency marginal, and time and frequency resolution.

Spectral variables -After computing the time-frequency spectrum, it is important to define spectral variables that effectively track its variations due to fatigue. We redefined the mean and median frequencies of the power density function of a wide-sense stationary process by substituting the time-frequency spectrum to the power density function. Again, this step is not trivial, but we demonstrated that the instantaneous mean and median frequencies can be estimated with acceptable errors on processes whose characteristics are similar to those of the myoelectric signal detected during cyclical movements.

Rejection of confounding factors - When the myoelectric signal is detected during dynamic contractions its time-frequency spectrum, and hence the values assumed by the instantaneous spectral variables, depend on numerous factors that are not related to fatigue. These factors contribute to increasing the variability of the instantaneous spectral variables, thus masking their fatigue related changes. In order to reduce the effect of these confounding factors, different approaches may be followed, depending on the characteristics of the myoelectric signal to be processed. Restricting the study to cyclic movements only, two different cases have been considered: fast- and slow-fatiguing cyclic contractions.

Fast fatiguing contractions - An example of fast-fatiguing contractions are those performed on isokinetic devices. Generally, since the contraction is maximal, the subject is not able to perform more than $10 \div 20$ flexion-extension cycles, and the characteristics of the myoelectric signal change substantially from cycle to cycle. In this case, the only possibility to increase the stability of the estimates of the instantaneous spectral variables is to perform appropriate time-averages within each single burst, first observing the structure of its time-frequency spectrum and then choosing proper windowing and time-averaging. If the velocity of the movement is rather low (60 deg/s), within the isokinetic portion of the movement it is often evident a generalized scaling towards the lower frequencies from burst to burst. In this case, the computation of the spectral variables over the frequency marginal can be sufficient to observe the effects of localized muscle fatigue, but this approach is not sound and may be misleading if applied without taking into account the characteristics of the time-frequency spectrum of the signal.

Slow fatiguing *contractions* - Examples of slow-fatiguing cyclical contractions are activities such as biking, walking, lifting a light weight. If the progress of fatigue is slow enough to be considered as irrelevant over N successive cycles and if the biomechanics of the exercise is sufficiently repeatable, the signal may be considered as quasi-cyclostationary over the N considered cycles. In this case, after choosing the portion of the signal burst that best allows to track spectral changes due to fatigue, the stability of the estimates may be increased by time-averaging the time-frequency distributions or the realizations of spectral variables obtained by processing N consecutive bursts.

The problem of selecting the portion of each burst that best allows to track the electrical manifestations of muscle fatigue must be solved by taking into account the biomechanics of each specific exercise, the characteristics of the muscles involved, and the characteristics of the time-frequency distribution of the signal.

CONCLUSIONS

In conclusion, results obtained in pilot studies involving fast- and slow-fatiguing cyclical muscle contractions demonstrate that it is possible to extend to dynamic conditions procedures previously limited to isometric and constant force contractions. More study is needed to correlate spectral parameters derived by this new method to the physiological phenomena that underlie the progression of muscle fatigue, however we believe that this effort is worthy being carried out due to the many possible applications of this novel approach in the field of ergonomics, rehabilitation, and sports medicine.

Using Mean Frequency to Monitor Fatigue in Dynamic Muscle Contractions

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INTRODUCTION

Tracking the mean frequency (MF) of the power spectrum of a myoelectric signal (MES) in time is an accepted method of monitoring fatigue in static contractions, since the spectrum characteristically shifts downward with fatigue. The objective of this investigation was to determine the feasibility of using a similar index to monitor fatigue in dynamic contractions.

To track the frequency content of a signal in time, a time-frequency transform must be utilized. Since the process of fatigue is relatively slow, its effect on static contractions is the production of a pseudo-stationary signal, so the short-time Fourier transform (STFT) can be used to track MF in contractions of this kind. Monitoring fatigue in dynamic contractions is more challenging. Since such signals are non-stationary to begin with, researchers have assumed that the STFT is inadequate for power spectral estimation because of the pseudo-stationary constraint of the STFT However, it may be possible to extract relevant MF data through the STFT even when this constraint is not met.

Shwedyk, in 1977 modeled the non-stationary MES produced from dynamic contractions as a signal resulting from an amplitude modulating process between an original stationary MES and a non-stationary variance [11. If the frequency spectrum of the variance is sufficiently low-band, it is feasible that the spectrum of the outcome of this process will yield a characteristic frequency similar to the original stationary MES. If both of these hypotheses are valid, then the STFT may be productive. Therefore, it may be possible to monitor fatigue in dynamic contractions with the STFT.

METHODS

A matlab simulation was used to evaluate the sensitivity of MF to changes in the variance of a fabricated MES signal. A white Gaussian signal was fabricated and filtered according to Shwedyk [l] to generate a stationary MES. The Fourier transform of the autocorrelation of this signal was estimated to produce its power spectrum 3000 ensembles were averaged to reduce the variance of this estimation to less than 2%, and MF data was determined. Then a variance was multiplied to the MES in a way that fashioned amplitude modulation (var(t $= 1 + \mu\cos(w_0 t)$). Finally, MF data from this signal was obtained for μ ranging from 0.2-l and f_0 ranging from 1 - 9 Hz .

To test the validity of Shwedyk's dynamic contraction model, real MES data was collected. A single assembly of data consisted of MES and force data collected from 5 sets of biceps brachii contractions. Each set involved 30 seconds of contraction split into 3 phases: 10 sec. of dynamic contractions followed by 10 sec. of static contractions followed by 10 sec. of dynamic contractions. For dynamic phases, subjects continuously flexed and extended their elbow between $\pm 45°$ from 90" flexion at 1Hz to gradually increase and decrease the length (and tension) of a spring which was attached to their wrists. For stationary contractions, subjects held the spring at a constant length/tension in one of 5 specified positions. A complete assembly consisted of sets at about -45°, -23°, 0°, +23°, and +45°. MES data was collected through a set of Red Dot™ surface electrodes and processed through a Tektronix AM502 differential amplifier system Force data was collected with a Grass F10 force transducer. Both signals were sent to an oscilloscope for display and to an A/D system for sampling (f_s = 1000 Hz) and recording. The MES data was divided into its phases (based on the force data) and each phase was sub-divided into 20 ½ sec. epochs. The periodogram was estimated for each epoch and MF data was obtained.

RESULTS

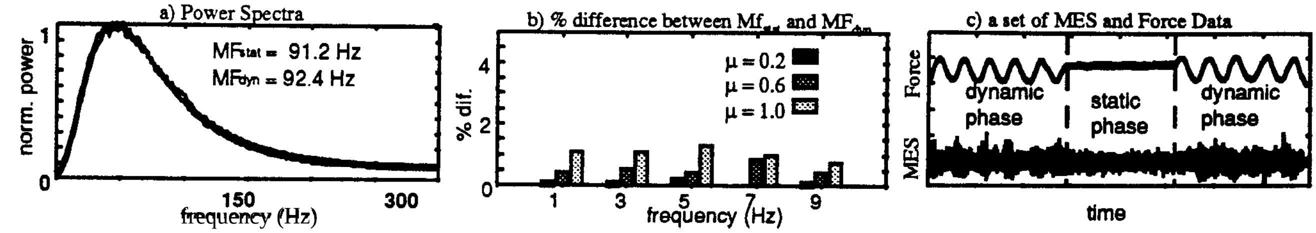

Figure 1: **a)** power spectra from 1 set of stationary and non-stationary simulated data **b)** % differences in static and dynamic MF in all simulated data c) 1 set of MES and force data (position = +22°).

	Position 1 (- 45°)			Position 2 (- 22°)			Position 3 (0°)			Position 4 (+22°)			Position 5 (+45°)		
	dyn	stat	dyn	dyn	stat	dyn	dyn	stat	dyn	dyn	stat	dyn	dyn	stat	dyn
mean MF (Hz)	69.6	68.7	65.3	69.2	69.1	65.6	66.2	76.3	65.7	68.6	78.1	67.5	70.6	71.4	66.2
SD	6.3	3.6	4.8	5.5	2.8	6.6	4.9	4.9	6.3	5.9	5.1	5.4	5.9	4.2	5.8
t	0.56*		2.54*	0.05*		2.2*	6.5		5.8	5.4		6.4	0.48*		3.26
mean MF (Hz)	76.6	74.2	75.0	76.2	76.7	76.3	75.4	84.1	72.0	76.7	89.6	75.7	76.9	89.1	75.5
SD	7.2	3.4	6.7	8.4	4.5	8.0	11.8	4.6	11.9	9.9	5.3	12.0	9.5	8.3	11.9
t	1.32*		0.52*	0.22*		0.18*	3.06		4.21	5.14		4.75	4.30		4.20

Table 1: Data obtained from 2 participants including: a) mean MF for each phase in each position **b)** Standard deviation of $\overline{MF}$ for each phase in each position c) t parameter for a 2 sample t-test between static and dynamic means for each position

Figure la depicts a sample set of results from the computer simulation and figure lb depicts graphically the differences between the stationary (Mf_{stat}) a:::! non-stationary (Mf_{dyn}) data for all sets of data. Since these difference are relatively small (< 2%), the hypothesis was accepted. That is, MF is relatively insensitive to non-stationarities modeled by amplitude modulation.

Figure lc depicts a sample set of results from the experiment designed to test the amplitude modulation model of dynamic contractions and the complete MF data set is tabulated in Table 1. According to the t-test (α=0.01) there are insignificant differences between the static phase means and the dynamic phase means indicated by an asterisk.

DISCUSSION

In the simulation, differences between stationary and non-stationary MF data were relatively small This was expected since the non-stationarity modeled in the simulation acted to amplitude modulate the signal. When a wideband signal such as MES is amplitude modulated by a low, narrowband signal, there is almost no effect on the MF of the signal This result was then used to test the amplitude modulation non-stationary signal model proposed for dynamic contractions. If the model is valid, then differences between static and dynamic MF data should be insignificant. Some of the data (indicated by asterisks in table 2) do verify this hypothesis, while other data do not. While the dynamic MF was consistent throughout all position trials, static MF increased with increasing position. Perhaps this is the result of a change in the geometry of the surface electrodes with respect to active motor units, or a recruitment of different motor units, or a combination of both, as position increases. At this time, data has only been obtained from 2 subjects. More data should yield conclusive results.

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ACKNOWLEDGMENTS

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Effects of Muscle Kinematics on Surface EMG Frequency during Fatiguing Dynamic Contractions

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Introduction

Most studies of the relationship between SEMG frequency characteristics and muscle fatigue have been done with prolonged, sub-maximal isometric conditions where fatigue is generally accompanied by shifts in the SEMG frequency spectrum to lower frequencies **(6).** The SEMG frequency content has been shown to be inversely related to muscle length (2). Consequently, the interpretation of SEMG signals from dynamic contractions is made more difficult because factors other than fatigue can influence the spectrum. Few studies have analysed the effects of muscle kinematics on the SEMG spectrum during dynamic contractions. Some have demonstrated SEMG frequency decreases during fatiguing dynamic contractions (4, 5). The purpose of this study was to further investigate the effects of muscle kinematics on biceps brachii SEMG frequency characteristics during a repetitive elbow flexion/extension task.

Methods

Fifteen performed repetitive flexion/extension movements of the right elbow with a hand held load (7 kg) until exhaustion. The rotations ranged from full extension (elbow angle of 0°) to full flexion (- 140”). Subjects were free to select the cycle frequency. SEMG signals were recorded from the biceps brachii. Elbow joint angle and velocity were monitored with a potentiometer. All signals were sampled at 1024 Hz and separated into 250 ms segments. The SEMG mean power frequency (MnPF), average angle and velocity were calculated for each dynamic segment.

MnPF data were sorted into two velocity directions (concentric and eccentric) with 7 angle ranges within each direction (O-20” through to 120-140” in increments of 20”). Segments with average EMG amplitudes below 5 %MVC were removed from the analysis. Regression analyses were used to determine average MnPF values at the start of the trial (when rested) and at the end (when fatigued) of the dynamic trial. A series of two-way repeated measures ANOVAs were used to determine the effects of velocity direction and joint angle range on rested MnPF values and the magnitude of fatigue induced decreases in MnPF. Orthogonal means comparisons were used to determine the significance of differences between individual means. Comparisons were made between concentric and eccentric values at each joint angle range and between angle ranges within a velocity direction. Significance was set at p<o.o5.

Results

The average number of elbow flexion-extension cycles was 47.9±18.7, the average duration of the dynamic trials was 155.3±64.0 s and the average cycle frequency was 19.1±4.1 cycles/minute. There was a progressive increase in concentric MPF with increased angle and decreased muscle length (Figure 1). There were no effect of joint angle on the rested eccentric MPF between 0” and 60” and then a progressive increase with further elbow flexion. Fatigue resulted in similar MPF decreases for the concentric and eccentric phases. MPF decreased in each angle range but the magnitude of this decrease became progressively larger as flexion angle increased and muscle length increased. Consequently, muscle length was observed to have no effect on MnPF when the muscle was fatigued.

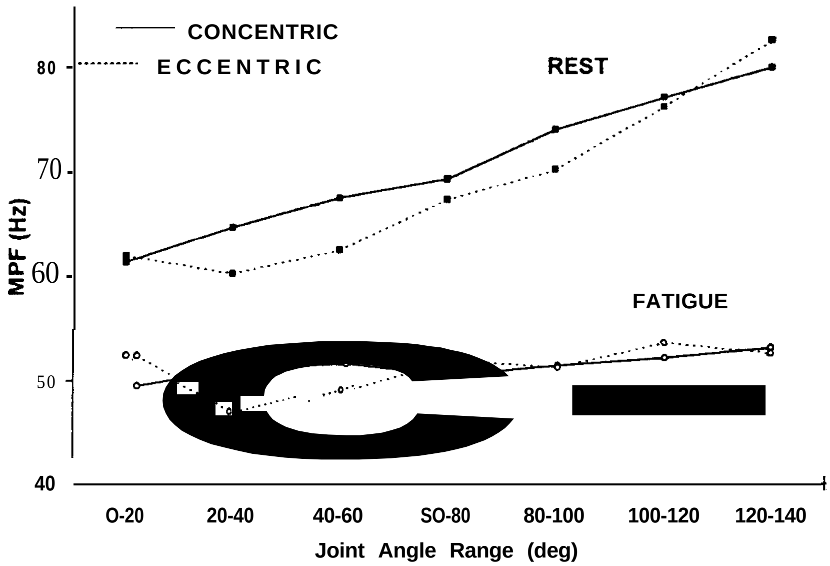

Figure 1: Average MnPF values at the start (Rest) and end (Fatigue) of the repetitive, dynamic elbow flexion-extension trial. Values have been pooled within joint angle range and movement direction (concentric and eccentric). (n=15 for each value).

Discussion

Some authors have proposed that the higher MnPFs associated with increased elbow flexion result from shorter muscle lengths having higher conduction velocities (1, 7). The current results are consistent with the proposed relationship between muscle length, conduction velocity and MnPF through the entire range of elbow flexion. The decrease in SEMG MnPF observed throughout the concentric phases (Figure 1) was consistent with previous dynamic studies of the biceps brachii **(4, 5).** Tesch *et al. (8)* observed MnPF decreases during concentric trials but no changes were observed during repeated eccentric muscle actions. In contrast, the current concentric and eccentric phases demonstrated similar fatigue induced decreases within each angle range.

The effect of length on MPF was observed to become progressively diminish over the course of the fatiguing trials (Figure 1). Fatigue caused larger MPF decreases in the higher flexion ranges and the MPFs were observed to be relatively independent of joint angle at the end of the trials. This convergence of MPF values with progressive muscle fatigue may indicate that there is some biological minimum for the action potential conduction velocities that dominate the EMG spectrum. Cupido *et a/* **(3)** propose that propagation failure occurs beyond the point where conduction velocity had decreased approximately 50%. With long fibre lengths the conduction velocity would already be relatively slow such that fatigue would only cause small decreases in velocity before transmission failure. However, in shorter fibres the initial conduction velocities would be much higher and could experience larger fatigue induced decreases before conduction would no longer be possible.

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Electrical Manifestations of Muscle Fatigue in Knee Flexors and Extensors During Isokinetic Exercise

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INTRODUCTION

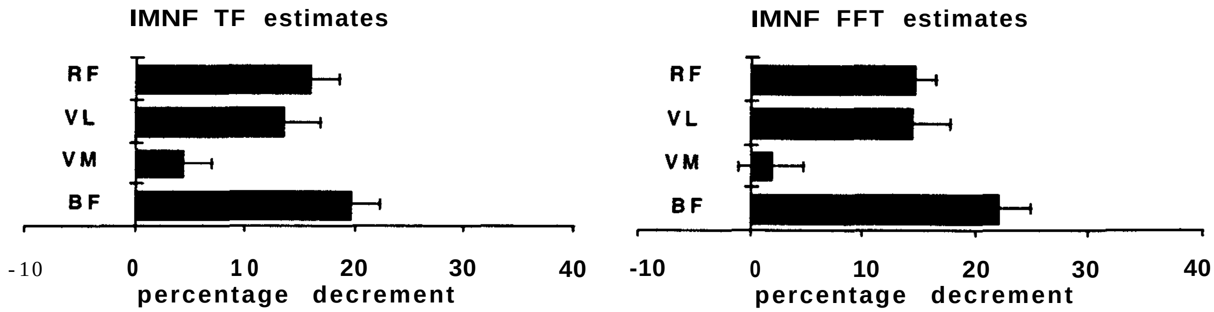
It is well known that during a sustained muscle contraction the power spectrum of the surface myoelectric signal (SMES) is progressively scaled towards the lower frequencies because of the progression of localized muscle fatigue. The spectral scaling is usually studied by means of spectral parameters, such as the mean and the median frequency of the instantaneous power spectrum of the signal, Previous studies analyzed the SMES recorded during static or constant-force isometric contractions, when the signal can be considered as a realization of a wide-sense stationary stochastic process with gaussian distribution of amplitude and zero mean. The analysis of dynamic contractions, during which the SMES is a nonstationary stochastic process, is still missing. Due to the nature of the considered contraction, there are several different factors, such as the change in the muscle length, the force exerted which is not constant within the angular range of movement, and the variations in the position of the body segments, which increase the variability of the estimated parameters. This paper describes the analysis of the SMES recorded during an isokinetic exercise (the leg flexo-extension) and presents the results obtained in terms of quantification of the electrical signs of muscle fatigue in four different leg muscles. The quantification of the progression of muscle fatigue during the exercise is obtained by two different approaches that lead to the determination of the fatigue pattern of the observed muscles.

METHODS

Ten healthy volunteers (nine men and a woman) with no previous knee disorders and age equal to 29 ± 16 years performed an isokinetic test consisting of 15 flexo-extension cycles of the leg with an angular velocity of 60deg/s. The range of motion was limited to 80 degrees. During the exercise, the SMESs were recorded by means of single-differential active probes from four thigh muscles: vastus medialis (VM), vastus lateralis (VL), rectus femoris (RF), and biceps femoris (BF). Myoelectric signals were sampled at 1024 Hz and were converted by a 12 bit A/D converter. Signal analysis was performed as follows: each signal burst was segmented in epochs corresponding to a range of motion of 2.5 degrees and for each epoch we estimated the instantaneous mean frequency of the signal (IMNF), using an estimator and based on the Choi-Williams distribution we recently developed. The IMNF estimates were averaged over the entire range of motion, in order to lessen their variability. This led to the estimate of a single mean frequency value for each signal burst. The least square regression line of the time course of the IMNF was then calculated, and from its slope we derived the percent decrement over the entire contraction (PDC). The PDCs of each specific muscle relative to the ten subjects were then averaged to achieve a quantification of the percent decrement typical of each muscle on our sample population. Moreover, the PDCs were also calculated starting from the frequency marginal of the time-frequency distribution computed over each signal burst. Results obtained on the sample population following the two different approaches were statistically compared.

RESULTS

The percent decrement of the IMNF estimates, obtained by means of time-frequency technique and averaged over the ten subjects, is reported by fig. la, while fig. lb shows the same parameter obtained from the frequency marginals of the distributions computed over each burst. The entire exercise, consisting of 15 flexion-extension cycles, was considered.



a) b)

DISCUSSION

Since the angular velocity fixed in this experimental protocol is relatively low, within the isokinetic portion of the movement the time-frequency based approach reveals a generalized scaling towards the lower frequencies of the time-frequency spectrum of the SMES. It follows that the IMNF estimates obtained by means of the time-frequency based approach and those obtained from the frequency marginals are expected to be similar. Then, also the IMNF percent decrements observed during the entire exercise are expected, in average, not to depend on the method utilized to estimate the IMNF values. Fig. 1 shows that our results confirm the cosiderations above and that both techniques give, in average, equivalent results. Specifically, if the purpose of a study is limited to observing the progression of the electrical manifestations of fatigue during a series of slow isokinetic flexion-extension cycles, the IMNF may be evaluated on the frequency marginal of the distribution, substantially reducing the computational cost. Nonetheless, the estimation of the time-frequency spectrun is essential a) to ascertain that the entire signal burst is scaled towards the lower frequencies and hence that the computation of the frequency marginal is sufficient, b) to investigate the evolution of the IMNF value within each single signal burst, and c) to develop methods that increase the capability of detecting limited effects of localized muscle fatigue that produce modest percent decrements of the IMNF during the entire exercise.

From a clinical point of view, fig. 1 shows that electrical manifestations of muscle fatigue consisting of a negative slope of the IMNF least square regression line are evident in each of the studied muscles. Moreover, it is also evident that while BF, RF, and VL show important IMNF decrements, the VM muscle exhibits the smallest change of SMES frequency content during the repeated cyclic contractions. This different behavior may be explained in terms of muscle architecture and recruitment modalities.

CONCLUSIONS

This paper presents a methodology to obtain the fatigue pattern of a muscle relative to a series of isokinetic flexion-extension cycles. We compare results obtained by using both a time-frequency based approach and the frequency marginal of the distribution for estimating the IMNF values relative to each cycle. We demonstrate that the two approaches are equivalent if only the electrical manifestations of muscle fatigue are to be studied. From a clinical point of view, we present the fatigue patterns obtained by studying four thigh muscles in ten healthy subjects. These preliminary data will be used to investigate the fatigability of knee flexors and extensors in subjects who underwent reconstructive knee surgery.

Fatigue of the Trunk Muscles in Isometric Trunk Rotation

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INTRODUCTION

Over 60% of back injuries are associated with trunk rotation (Manning et al. 1984). Trunk rotation being an asymmetric activity may cause differential fatigue among trunk muscles. The latter could result in overloading of some fatigued muscles resulting in injury. The objective of this study was to study the pattern of fatigue in trunk muscles during isometric axial rotation in neutral posture.

METHODS

Twelve young and healthy subjects (7 male, and 5 female) were recruited to the study after screening for musculoskeletal disorders. Fourteen pairs of disposable pregelled surface electrodes (HP 144445) were applied on 7 trunk muscles bilaterally at an interelectrode distance of 2 cm after suitable preparation of the skin. Electrodes were placed on erector Spinae levelled with spinous processes of T_{10} and L_3 vertebrae bilaterally 4 cm lateral to the tips of the spinous processes. Surface electrodes were also applied to the external and internal obliques, rectus abdominis, pectoralis major, and latissimus dorsi bilaterally. Such prepared subjects were seated in the chair of the axial rotation tester (AROT) (Kumar 1996). The seat was aligned with the AROT rotation axis and the harness fastened to the subjects shoulder allowing free rotary motion. The seated subjects were stabilized hip down using velcro straps. The AROT was brought to the subject's neutral posture and put in isometric mode. These subjects were instructed to attempt axial rotation to their left at 60% of their MVC (measured prior to the experimental session) with a visual feedback for a period of 2 minutes or until they could no longer hold at that force level.

These electrodes were connected to a fully isolated preamplifier with a gain of 100 through short cables near the body. These preamplifiers with low nonlinearity and high CMRR (135 dB) were connected to amplifiers which fed to a A to D board. All channels were sampled at 1 kHz for a period of 2.1s at the start and at every 10s interval for a period of 120 seconds.

The task was divided in 10% segments of the task cycle. The signals obtained were subjected to FFT analysis after Welch (1967). The median frequencies (MF) for each channel for each of the task 10 percentile vallues were plotted and the slope of the decline was calculated by fitting regression lines to the data. The slopes were subjected to analyses of variance.

RESULTS

The mean duration for which the isometric axial rotation could be held by male and female subjects was 102.05s and 113.13s respectively. A significant drop in median frequency of all muscles tested was clear. The left latissimus dorsi and thoracic erector Spinae demonstrated highest median frequencies and greatest drop with progression of time among dorsal muscles. Among ventral muscles the highest value and greatest drop in MF was observed in left pectoralis. There was no significant gender difference in MF. The MF of different muscles were significantly different ($p < 0.001$). The maximal MF for all muscles occurred in the first 10% task cycle and the minimum appeared between 60% to 100% of the task cycle. The MF declined from the beginning to the end, but the most precipitous drop was in the first 10% task cycle. The slope of the decline of the MF for different muscles was significantly different ($p < 0.001$). The slope of the decline of the agonists was steady and smoother compared to the antagonists. The slope of decline of the MF was significantly different between males and females ($p < 0.02$). Within each muscle the segmental slopes were significantly different ($p < 0.001$).

DISCUSSION

The prime movers for axial rotation were ipsilateral latissimus dorsi, erector Spinae and internal obliques and Contralateral external obliques. In sustained high level contraction the drop in MF of the prime movers was slow and steady whereas those muscles not primarily involved in the task was more variable. The role of the pectoralis muscle in this study was affected by its ability to exert on the harness as a complimentary muscle. The MF of different trunk muscles were significantly different. The slope of the decline of the MF of different muscles were significantly different. There was no significant difference between the two genders and the segmental slopes did not differ significantly. The latter implies a uniform strategy employed by the muscles with reliable patterns of fatigue. Since the same muscles are involved in active rotation and return to the neutral posture (Kumar et al. 1996) the rotational activities may cause mechanical failure of the repeatedly used muscles potentiating injury.

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Effects of Fatigue on the Contractile Properties and Behaviour of Motor Units in Human Muscle

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INTRODUCTION

Although muscle fatigue has been extensively studied in animals and humans, many questions remain open for discussion. There is for example a paucity of information at the motor unit (MU) level since most of our knowledge derives from animal studies during electrically induced fatiguing contraction, a condition which is different from that prevailing during voluntary muscle activation. Among the works performed in humans, studies were mainly focused on the effect of fatigue on the recruitment and firing rate of MUs during weak contractions (1,2, 3). The effect of fatigue on single MU contractile properties during voluntary contractions were investigated in only 3 studies and restricted to weak sustained effort (cf 5,6). The present work was designed to contribute to the discussion by recording simultaneously the contractile properties and behaviour of single MUs during intermittent fatiguing voluntary contractions.

MATERIALS AND METHODS

The experiments were performed on several occasions on 8 healthy subjects aged between 21 and 40 years and well accustomed to the experimental procedure. The force and surface electromyogram (EMG) of the first dorsal interosseous were monitored during the different tasks. In addition, single MU action potentials were recorded by selective wire electrodes (50 μm in diameter) inserted into the belly of the muscle by means of an hypodermic needle. The fatigue test consisted in intermittent contractions at 50% of the maximal voluntary contraction (MVC) recorded before the test. The task consisted to reach the target force in 3s, to maintain this level during 10s and then to slowly return to base line in 3s. A rest period of 4s was allowed between two successive contractions. The test was interrupted (endurance limit) when the subject was unable to fit the task criteria during 3 successive contractions. For each single MU, recruitment and derecruitment threshold forces were analysed. Mechanical properties of selected MUs were studied before and after the fatigue test by the "spike triggered averaging" method (4). The recovery from fatigue of these parameters was tested after 15 and 30 min rest period.

RESULTS

The endurance limit was variable from subject to subject and ranged from 6 to 15 min and a total of 63 MUs with recruitment thresholds ranging from 0.5 to 55% of MVC were analysed. During the fatigue tests, the recruitment order of MU did not change but there was a progressive recruitment of additional MUs. The previously active MUs showed a progressive mean decrease of their recruitment force ($-9.8 \pm 1.4\%$; $p < 0.001$), this change being more pronounced for high threshold MUs as compared with low threshold MUs. In contrast, MU derecruitment threshold exhibited a progressive increase over the course of the test. The analysis of the mechanical properties (twitch force and time course) of the pooled MUs before and after fatigue did not show significant change, however if MU were subdivided on the basis of their recruitment thresholds (cf 5), MUs of low recruitment threshold (< 15% MVC) showed a systematic increase in force which was accompanied by a slowing of their twitch time course. On the other hand, MUs of high recruitment threshold (> 25% MVC) exhibited a decrease in force and an acceleration of their twitch time course. No systematic trend was observed for MUs recruited between 15 and 25% of MVC.

DISCUSSION

This work, analysing fatigue during intermittent contractions, confirms and documents previous data on sustained contractions showing a progressive decrease of MU recruitment threshold and activation of additional MUs during the course of the test (2,3). In addition, our results enlarge these studies since they originally indicate that recruitment threshold decrease is specifically related to the MU type, high threshold MUs showing greater decrease than low threshold MUs. The opposite behaviour of the mechanical properties of different MU recruitment threshold emphasises the complexity of interpreting the underlying mechanisms. Because no enhanced MU synchronisation after fatigue appeared in our experimental conditions, the increase in twitch force of the low threshold MUs could be explained by mechanisms such as change in muscle stiffness or post-activation Potentiation. These adaptations should contribute to preserve the mechanical efficiency of low threshold MUs during longer time period.

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Can Motor Unit Force Potentiation Compensate for Declining Activation Rate During Submaximal Fatiguing Contractions ?

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Introduction

Paradoxically, during submaximal isometric contractions motor unit (MU) firing rates decline over periods up to 30 seconds, even when force is maintained at a constant level (1, 2). This has been interpreted as an expression of "muscle wisdom", i.e. a strategy to delay manifestation of fatigue or complete exhaustion of the muscle (3, 4). However, there is an ongoing debate as to whether for submaximal activation force can be maintained without delayed additional recruitment of non-fatigued MUs to stem the putative loss of force arising from fatigue and declining activation of early recruited MUs. Those, who maintain that there is no experimental evidence of additional recruitment during submaximal constant-force contractions lasting around 30 s (2), have argued that there is indeed no need for such compensatory recruitment, since force Potentiation could easily compensate for declining activation rates. Force Potentiation during the initial period of maintained tetanic stimulation (at constant rate) of large MUs (types FF and FR) is a well known phenomenon (5-7). However, it has never been examined experimentally whether, during muscle wisdom-like activation, force potentiation would indeed be sufficiently powerful and suitably timed to counteract the decline in force resulting from a progressively diminished stimulation rate. To put it succinctly, for submaximal contraction, the question is whether wisdom-like activation is an expression of damage control (i.e. minimization of fatigue) or of optimal control (i.e. matching of contractile properties).

Methods

The experiments were conducted on four cats deeply anaesthetized with pentobarbitone. The peroneus muscle was prepared with intact nerve supply in the otherwise completely denervated left hindlimb. After a lumbosacral laminectomy, the L7 and Sl ventral roots were exposed and functionally single motor axons to the muscle were isolated by microdissection. Special care was taken to attempt isolation and subsequent investigation of all MUs that could be detected, regardless of size and contractile properties. Individual MUs were characterized by their axonal conduction velocity and tetanic force, the latter measured during brief episodes of non-fatiguing stimulation at 30/s. Single MU force production was measured, at fixed intermediate muscle length, during stimulation at physiological rates over a 40 s period, using two separate stimulation profiles, constant or decaying. For the constant-activation profile mean stimulation rate was fixed (18/s); for the decaying-activation profile, stimulation rate gradually declined (from 25/s to 16/s) with a time constant of 9 s, so as to mimic the firing rate profiles of MUs recorded during voluntary constant-force contractions in humans (1, 2). In either case, the stimulation profiles were generated with customized software and stored as lists of inter-impulse intervals. Further, to imitate physiological discharge rate variability, Gaussian noise was added to the interval lists, so as to produce pseudo-random fluctuations of stimulation rate with a coefficient of variation of 15%. Force was recorded at the tendon, using a high sensitivity piezo-electric force transducer, low-pass filtered and digitized at 200/s. These force records were stored in computer files for off-line analysis. For each MU, a minimum of five minutes was allowed between successive stimulation episodes for recovery from fatigue or force Potentiation.

Results

Force profiles were recorded for 58 peroneus brevis MUs during both decaying and constant activation. Tetanic forces ranged from 10 to 300 mN, and their distribution was distinctly skewed towards small MUs, indicating that the present sample was not unduly biased in favour of large FF or FR motor units.

The force profiles elicited by stimulation of individual MUs showed a wide range of shapes. For constant stimulation the profiles included constant, predominantly decreasing (fatiguing), predominantly increasing (potentiating), waxing and waning, or waning and waxing patterns. In order to assess the impact of this variety of force transients on whole muscle force during recruitment of increasing numbers of MUs, estimates of forces generated by subpopulations of MUs were calculated, based on algebraic summation of individual force transients. MUs were rank ordered according to size (tetanic force) in order to mimic orderly recruitment. 12 overlapping subpopulations of progressively larger size were formed, so as to encompass the first 8.5%, 17%, ... 100% of the MU sample. The family of the subpopulation force profiles revealed consistent shapes and systematic trends: for constant stimulation, force was approximately constant for subpopulations up to about 45% of the MU pool. For larger subpopulations potentiation was increasingly manifest, amounting to an augmentation of force up to 65% (over 20 s) for the entire sample of MUs. Force build-up was gradual, with a time constant of about 8 s. For decaying activation, the subpopulation force transients followed the time course of activation (i.e. force declined), as long as only small MUs were activated, while for larger subpopulations (when the larger MUs were also recruited) the force transients approached approximately constant profiles.

Discussion

The present observations demonstrate that, at physiological rates of MU activation, force Potentiation may contribute significantly to force generation over periods up to 40 s. In the case **of** constant activation, about one third of the steady state force, which was reached after 20 s, was attributable to Potentiation.

With a profile of decaying activation, imitating the muscle wisdom-like firing profiles reported for voluntary contraction in man, force Potentiation approximately compensated for the declining ensemble activation rate, provided large MUs were also activated. In contrast, when only small MUs were recruited (up to 45% of the pool, generating about 25% of total force) potentiation was not observed and compensation for declining activation was insignificant.

Thus, for motor unit recruitment according to size, low-level constant-force contractions would require either activation of a fixed number of MUs at constant rates or additional recruitment if even the smallest MUs are activated at muscle wisdom-like declining rates. In contrast, it appears that for constant-force contractions above 30% of MVC, declining activation, especially of large MUs, is an optimal strategy for the generation of constant force for periods of activation up to at least 40 s. This supports the notion that for intermediate or high levels of contraction there is no need of additionaI recruitment to maintain constant force in the face of declining MU firing rates.

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Muscle Activity of ACL Deficient Knee Subject With and Without a Functional Knee Brace During Jogging

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INTRODUCTION

In the last ten years research has demonstrated that the ACL deficiency influences muscle activity. Some have concluded that the hamstring muscles become the first muscle acting in the stabilization of the knee joint [1-2]. Quadriceps acted directly on the antero-posterior displacement of the femur relative to tibia[3]. The gastrocnemius muscles action is still not clear. The functional knee brace for the ACL deficient knee is meant to increase the joint stability. How does the functional knee brace affect muscle activity? Only a few researchers have concentrated their work on this subject[1-2-4]. More studies are necessary to understand the effect of the brace on myoelectric activity of muscles implicated in movement of the knee.

The purpose of this study is to compare the muscle activity of the ACL deficient knee with and without a functional knee brace during jogging.

METHODS

Four ACL deficient males volunteered for this study. Active surface electrodes (ME 3000) were used to measure the myoelectric activity of the vastus (medial and lateral), the hamstrings (biceps femoris and semitendinosis), the gastrocnemius (medial and lateral). Each subject performed ten jogging trials with and without the brace on a five meters surface. The speed was not controlled to permit at the subject to feel safe and secure during the activity. Vertical forces were collected for all trials with a Kistler forceplate. The subjects was also filmed with a VHS video camera (Panasonic Camcorder filmed at 60 Hz) to control the portion of the foot on the forceplate. The sequence of EMG analyzed corresponds to the period from heel strike to toe off. EMG data was filtered (high pass filter at 5 Hz) to remove cable artefacts. The linear envelope and the integral were calculated after which time and amplitude were normalized. Vertical force data was also reduced and normalized by time and amplitude. Both the electromyography and the vertical forces were compared between braced and unbraced conditions.

RESULTS AND DISCUSSION

The biceps femoris was out of phase in the braced compare to the unbraced condition (Figure 1). The semitendinosis muscle activity had a tendancy to increase during the braced condition (Figure 2). The lateral gastrocnemius showed a decrease in activity for the braced condition (Figure 3). The vertical forces also decreased in the braced condition (Figure 4). The other muscle did not show noticeable difference between both conditions. The data collected during this research seems to show that the brace will alter the muscle activity and the vertical force signature for a jogging type of activity. It is still not possible to quantity the difference in either EMG or vertical force between conditions.

In conclusion, in a dynamic movement like jogging, the functional knee brace tend to alter both the myoelectric activity of specific muscles and the vertical force. A more in-depth research study with large sample size is necessary in order to obtain inferential statistics on the effect of FKB during jogging.

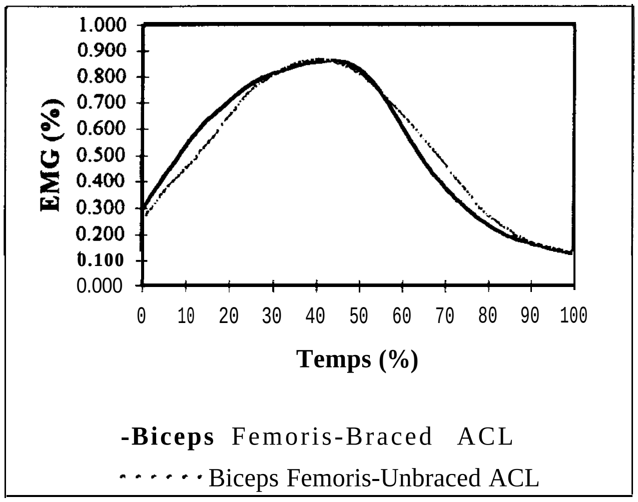

Figure 1 EMG curves of the biceps femoris with and without the brace.

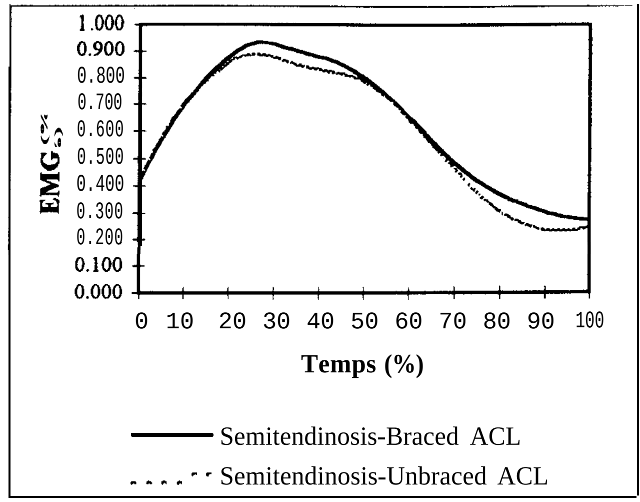

Figure 2 EMG curves of the semitendinosis with and without the brace.

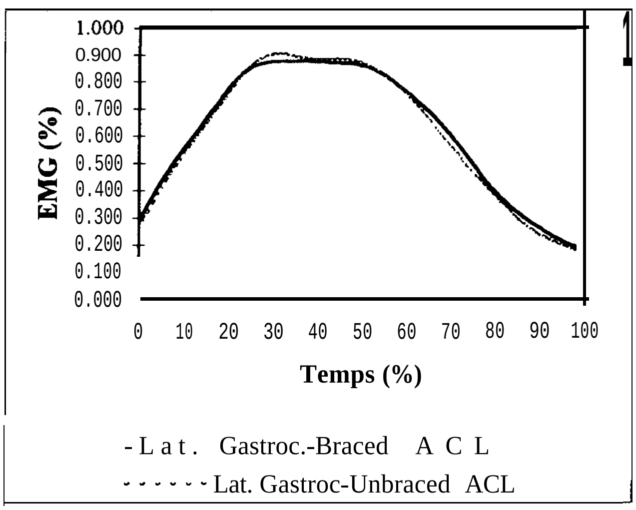

Figure 3 EMG curves of the lateral gastrocnemius with and without the brace.

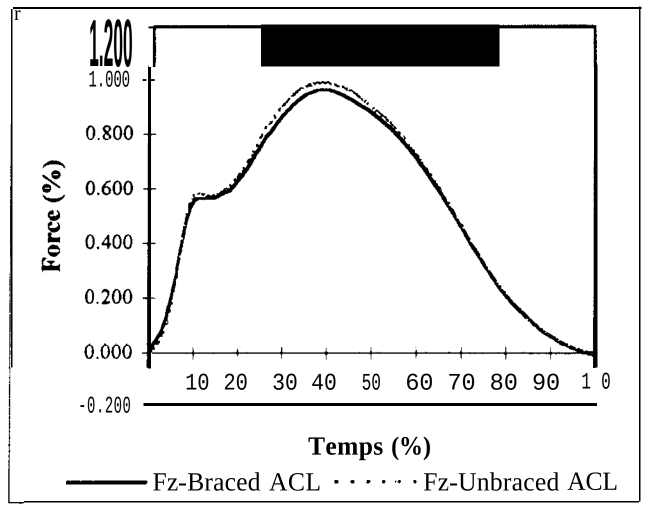

Figure 4 Vertical forces curves with and without the brace.

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The Natural Sciences and Engineering Research Council has partly funded this project. The knee braces were provided by Donjoy, Inc. Smith & Nephew, Carlsbad, California, USA.

Effect of Functional Knee Brace in Knee Joint Stiffness for Control and ACL Deficient Subjects

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INTRODUCTION

The concept of joint stiffness and passive moments of force was mainly introduced in the early sixties in connection with another research problem; the quantification of joint stiffness in rheumatoid arthritis and other related pathologies. The elastic moments of force are the moments of force that resist joint movement due to the deformation of all tissues such as the joint capsule, tendons, and ligaments that cross the joint when the muscles are inactive [1-3].

The purpose of the study is to measure the knee joint stiffness of both the normal and the ACL deficient knee patients with and without a functional knee brace (FKB) (Legend Brace: Smith & Nephew Donjoy Inc.).

METHODS

Seven ACL deficient males and ten males with normal knee joint volunteered for this study. The passive elastic moment function and joint stiffness were measured by moving the lower leg in a passive mode in both supine and prone positions[4]. This passive motion of the lower leg was performed on an isokinetic device (KIN-COM 500H) at a fixed velocity of 5°/s from full extension to 105 degree of knee flexion. This procedure was performed with and without the FKB on both injured knee and normal subjects. For the normal knee subjects, five subjects were tested on the lefl leg and five with the right leg under both braced and unbraced conditions. The knee joint stiffness was calculated from both the change in the passive elastic moment of the knee joint and the viscous damping values.

RESULTS AND DISCUSSION

For both groups of subjects the elastic moments of force were increased by about 20% at 10 degrees of flexion with the use of the functional knee brace. However, at 105 degrees of flexion, the change in elastic moment of force with the braced condition was more marked. The increase was 27.3% for the normal subjects and 33% for the ACL deficient subjects respectively. An ANOVA was used to test for significance of the elastic moment of force ($p<0.05$) between the normal and ACL-deficient groups. No significant difference was detected between the groups. However there was significant difference ($p<0.05$) between the braced and unbraced conditions. The knee joint stiffness (Nm/rad) values were not significantly different between groups but highly statistically different between conditions ($p<0.0001$) (Figure 1). The functional knee brace significantly increases the passive moments of force at the knee joint. That was a greater passive moment of force at the more extreme angles (10 and 105°).

In summary, the functional knee brace significantly increases the net passive joint moment of force (NPJMF) and knee joint stiffness. The functional knee brace has a mechanical impedance on the NPJMF irrespective of the degree of injury to the knee joint. The functional knee brace increases the knee joint stiffness. Thus the functional knee brace seems to have a direct mechanical effect on the knee joint moment of force.

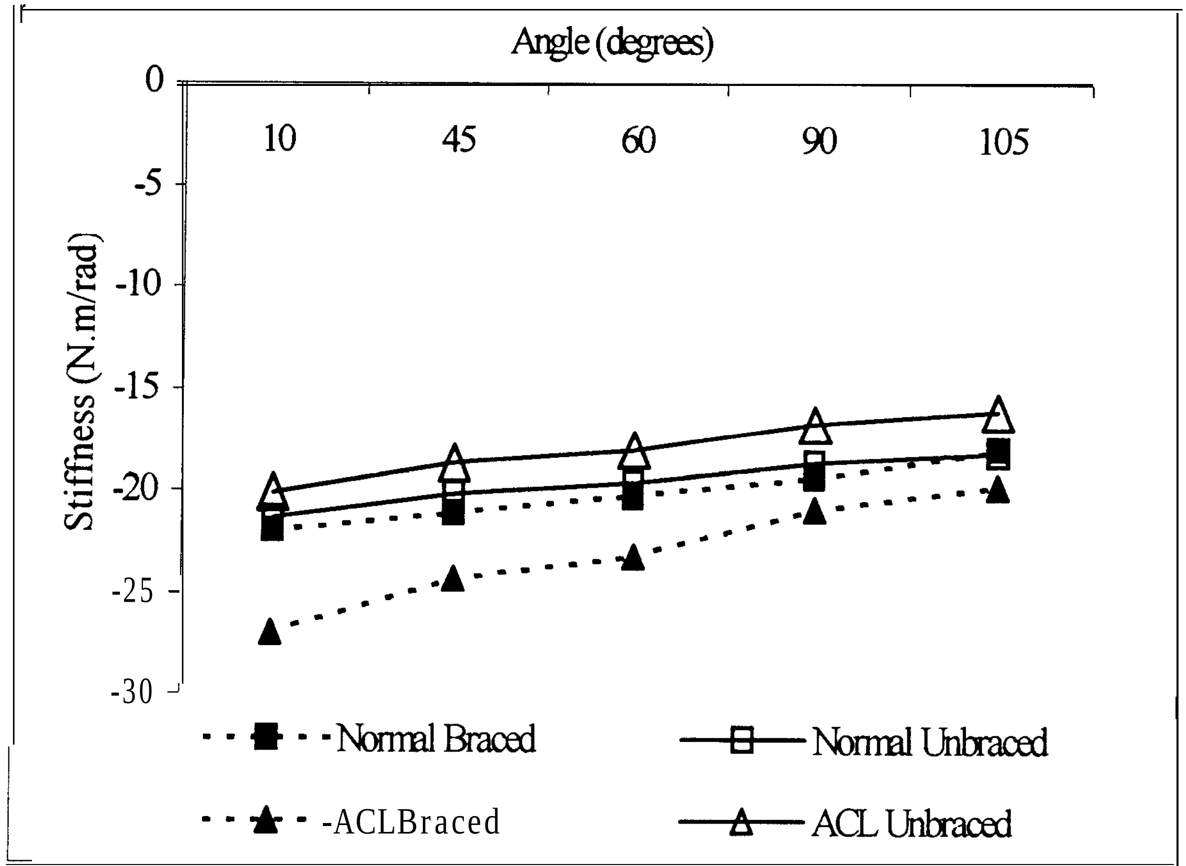

Figure 1 Knee Joint Stiffness as a Function of Knee Joint Angle.

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ACKNOWLEDGEMENTS

This research was supported by Natural Sciences and Engineering Council. The knee braces were provided by Donjoy, Inc. Smith & Nephew, Carlsbad, California, USA.

Stair Ascent Strategies Four Months After Total Knee Arthroplasty in Persons with Osteoarthritis

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INTRODUCTION. Stair climbing is a test of choice to detect locomotor abnormalities and identify adaptive strategies[1,2] The stance phase of stair ascent requires a coordinated and powerful action from the lower limb's chain of extensor muscles in order to support the upright posture and progress to the next step. The support moment, which summarises the support task by summing the relative contributions of each joint's moment of force, has been shown to remain quite consistent in spite of variation in the individual joint moment[3,4]. Although particular changes in the joint moment profiles adaptations have been described as a result of pathology[3,4], no reports using the support moment theory to explain walking or stair climbing strategies of subjects with total knee arthroplasty (TKA)[1,5,6] were found. Hence, the purposes of this study were to: 1) identify the locomotor strategies used to maintain an efficient support moment 4 months after TKA and 2) verify whether subjects with different locomotor strategies showed support moment consistency.

METHODS. Fifteen persons who underwent a TKA with a mean age of 65 ±11 years and weight of 79 ±18 kg participated in the study. They were recruited in 4 hospitals and performed locomotor and clinical tests at 5 different stages: preoperatively and 2, 4, 6 and 12 months postoperatively. This report focuses on the stair ascent test performed at 4 months after TKA. A trial of the stair ascent test consisted of climbing a staircase of 4 steps (17.8 cm high, 25.3 cm deep, slope 38°) at free speed with a banister available on the non-evaluated side. Five to seven trials were performed consecutively. Sagittal movements of the hip, knee and ankle joints were recorded using a 2-D video system and reflexive markers. Temporal parameters of the stair ascent cycle were calculated from footswitches taped under the heel, mid-foot and toe. Moment of force and mechanical power for each joint were calculated from the ground reaction forces recorded with a force platform (AMTI). During each trial, activations of 4 muscles (medial gastrocnemius, medial hamstrings, vastus lateralis, tibialis anterior) on both limbs were picked up with surface electrodes (Meditrace silver-silver chloride, 10 mm). The pre-amplified EMG signals were filtered, rectified and time averaged with a time constant of 0.08 s. They were recorded simultaneously with temporal, kinematic and kinetic data (sampling frequency 60 Hz), processed by a Pentium computer and stored for further analysis. From the recorded trials, 3 cycles were retained for analysis. A stair ascent cycle began with the initial contact of the evaluated foot on the first step and ended when the same foot contacted the third step. The individual knee moment profiles were used to classify subjects into different stair ascent strategies according to these two criteria: 1) the peak value of the first knee extension moment (KEM) phase and 2) the point in the cycle (in percent) when the KEM reverses to a knee flexion moment (KFM)[1]. At first, the classification was made by visual observation of the knee moment profiles and then by a cluster analysis (SPSS) using the same criteria. Data from 21 healthy subjects of similar age (mean 67±8 years) and weight (71± 13 kg) provided normative values (Table 1 and Figure 1).

RESULTS. Three knee moment strategies were identified. These strategies are illustrated in Figure 1. Six subjects were classified in the first strategy (Sl) called «low knee *extension moment (KEM) magnitude or hip strategy)). In* Sl, the timing in the knee extension and flexion moment phases was similar to normal but the peak magnitude of the KEM was reduced (peak value 0.39 N.m/kg Vs 0.88 in normal). The zero crossing point occurred at 40% of the cycle as in the normal group (44%). An increase in the hip extensor moment and the medial hamstring activity was also observed, most likely to stabilise the forward trunk lean and to compensate for the lower ankle and knee extension moments of force. Six subjects adopted the second strategy (S2) called <<*late crossing point or knee and ankle strategy*>>. This strategy is also characterised by a reduced KEM peak value (0.37 N.m/kg) but more specifically, by a late crossing point occurring at 58% of the cycle. This implies that the KEM is prolonged for almost the entire stance period and that from mid stance to early push-off a KEM is generated instead of a knee flexion moment KFM). The KEM from 35% to 55% of the cycle is associated to decreased moments of force at the other joints, especially the hip. This subgroup ended the stance phase with the three joints flexed and pushed the body over the next step using mainly ankle extensor muscles. The third strategy (S3), the *((early crossingpoint or hip and ankle strategy)),* was observed in three subjects. The crossing point in S3 occurred at 24% of the cycle and the KEM peak value was the lowest (0.24 N.m/kg). From 24% to 44% of cycle this subgroup presented a KFM while the others had a KEM. Changes in knee moment profiles were compensated by: 1) above normal hip moment values and medial hamstrings activity and 2) a near-normal ankle extensor moment which helps prevent knee collapse. In spite of different knee moment profiles, the support moment was found to be consistent between the three subgroups of subjects (Fig. 1). Moreover, the cycle duration of the three subgroups was not significantly different. When an extreme value was removed, the S2 mean cycle duration become 2110 ± 772ms (Table 1).

Table 1. Temporal characteristics of stair ascent

GROUP	Cycle duration (ms)	Stance (% of cycle)	Banister used
Normal	1331±206[a]	66.6±2.1[a]	o/21
TKA[b]	2566±1615*	66.4±5.2	7/15
Strategy l(6)	2276±987	67.6±5.4	216
Strategy 2 (6)	3082±2497	66.4i5.5	3/6
Strategy 3 (3)	2094±405	63.3±5.7	2/3

[a]mean± 1 SD, [b]Total Knee Arthroplasty, '(number of subjects) * signiticantly longer compared to normal group.

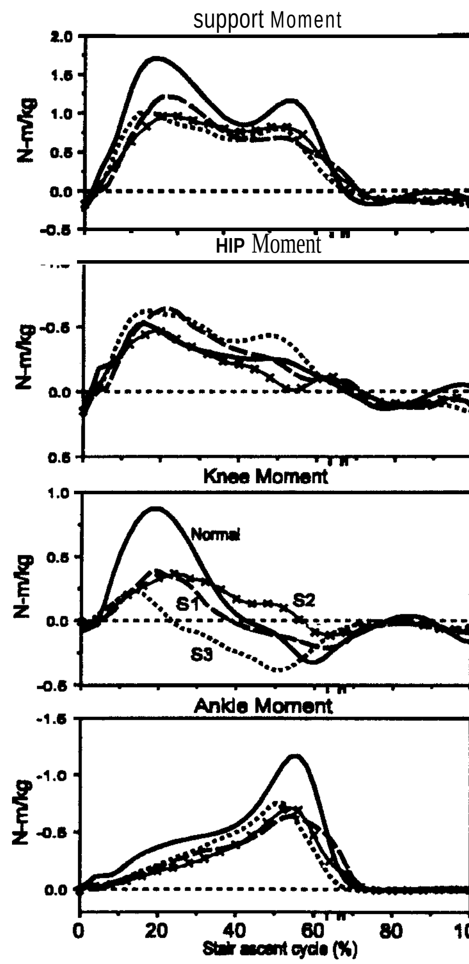

Figure 1 Moment profiles in the three strategies

DISCUSSION. Many factors have been suggested to explain why people use different strategies for level walking and stair ascent after TKA [1,5]. The extents of the deficits in joint proprioception, strength, range of motion, the preoperative stair ascent level of performance, duration between the onset and surgery and pluriarticular involvement have been implicated. In our study, subjects from the S2 strategy had decreased knee extension during stair ascent, which may explain the prolonged KEM in this group'. We could not verify the effect of the other factors in the present study. It is also not known whether these strategies change over time with recovery or which factors are linked to the stability or change in the strategy.

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Postural Strategies During Transition from Double to Single Limb Standing following Hip Arthroplasty

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Introduction

The transition from double to single limb standing (DSLS) requires the coordination of horizontal ground reaction forces (GRFs) under the motor (lifting) and postural (stance) limbs. In normal subjects, the motor (M) limb contributes a larger proportion of the lateral impulse for weight transfer than the postural (P) limb, yielding a M/P ratio of 4: 1. This ratio has been shown to be velocity dependent’ and to be modified in patients with a hemiparesis.[2] In addition, it has been reported that greater force rate development under the motor limb results in exaggerated oscillations of the center of mass during one leg-standing in children with sensori-motor disability.[3] Thus, the M/P ratio and the force rate are complementary measures that should help to better understand the role of horizontal GRFs during DSLS. In the present study, we used these two variables to characterize strategies of weight transfer during the DSLS task after a total hip arthroplasty. Given the residual muscle weakness and changes in proprioception after hip arthroplasty, it was expected that patients would use different motor strategies to transfer weight in DSLS.

Methods

Ten women who had a hip arthroplasty 10.8 (± 3.6) months earlier (EXP) and 10 age-matched women (CTL) were tested. Subjects stood on two separate AMTI forceplates with their weight evenly distributed, and lifted the operated limb (condition I) or the non-operated limb (condition II) to an auditory cue. The recording of GRFs and kinematics (3D Optotrak) started 500 ms before the auditory cue (baseline). Data from 4 to 5 successful trials (one leg standing > 1.5 s) were averaged. Comparisons between groups (Mann-Whitney U test) were made for the following variables: change in the lateral impulse during loading of the M limb (from T, to T,), ratio of M/P change in the lateral impulse, horizontal peak force rate, absolute and relative duration of sub-phases of the DSLS, hip local moment and power (frontal plane). and changes in hip and trunk angles. Correlation coefficients were computed for the peak force rate of the M limb (from T_1.to T_2) and the P limb (from T_2 to T_3) using the Spearman test; the level of significance was set at p=O.O5.

Relationship Between Peak Force Rate During Loading of Motor and Postural Limbs in condition II

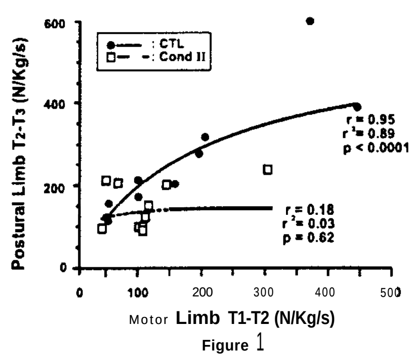

Figure 1

Results

For both conditions, there was no difference between groups in the change in lateral impulse nor in the M/P ratio. Lower (p=0.03) peak horizontal force rates were found, however, during the loading of the operated P limb (from T_2 to T_3) in condition II. The latter resulted in a longer duration of the loading phase (from T_2 to T_3) on the operated P limb. Moreover, the force rate under the P limb (from T_2 to T_3) was

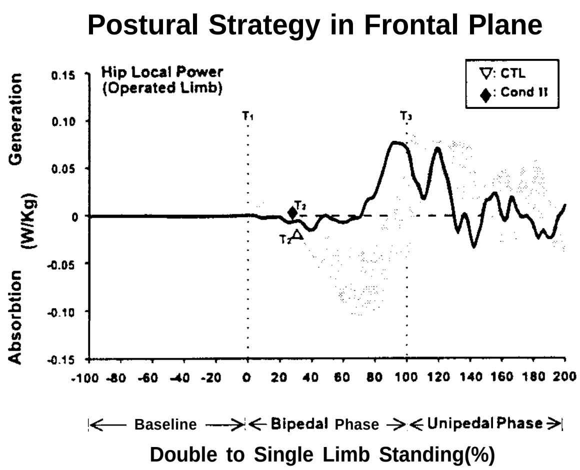

Figure 2

weakly correlated (r=0.18) to that under the M limb (from T, to T,), in contrast to CTL subjects (r=O.95) (fig 1). In addition, although both groups developed similar abductor moments, patients used a different postural strategy. The hip power curves (fig 2) indicate that the patients generated instead of absorbed energy by the hip abductors, and this was accompanied by a relative abduction at the operated P hip instead of the adduction seen in the CTLs. Lastly, patients had an earlier and larger lateral bending of the trunk towards the side of the P limb.

Discussion

The main finding is that when the operated limb was the P limb, patients used a different postural strategy to achieve their weight transfer. The impaired peak force rate under the operated P limb was associated with a longer loading time on that limb, resulting in a temporal reorganization of the task. In fact, on the operated P limb, patients failed to develop a force rate that was scaled to that of the non-operated M limb as found in CTLs. The latter resulted in more involvement from the trunk to decrease the mechanical demand on hip abductors. Therefore, patients combined larger trunk bending toward the side of the operated P limb with concomittant hip abduction as a compensatory strategy.

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A New Perspective on Dynamic Stability in the ACL Deficient Knee

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INTRODUCTION:

Not all people with anterior cruciate ligament (ACL) deficiency are created equal. Work in our laboratory has definitively shown that the knee joint laxity resulting from ACL rupture is not related to functional ability. Some individuals with ACL deficiency have significant knee joint laxity and are able to participate in athletic activities involving hard cutting and pivoting with no symptoms of instability (copers). Studies reported in the literature show few consistent differences between the movement and muscle activity patterns of ACL deficient and uninjured people. We believe that the inclusion of copers in samples of "ACL deficient subjects" may be responsible for the lack of consistent results. This study compares movement and muscle activity patterns of uninjured individuals to two groups of ACL deficient individuals (copers and non-capers).

METHODS:

Twenty four subjects have been tested; the results from 15 subjects are reported here. Five subjects were uninjured, five were copers (no more than 1 episode of giving way with regular participation in pivoting sports) and five subjects were non-capers (2 or more episodes of giving way with daily activities and are unable to return to sports). Subjects were tested using a 120 Hz, 3-D motion analysis system (Vicon 370, Oxford Metrics, London, England). Force and EMG data were collected simultaneously at a rate of 960 Hz using a force platform (Bertec Corporation, Worthington. OH) and an eight channel FM radio telemetry electromyography (EMG) system with active surface electrodes (B & L Engineering, Santa Fe Springs, CA). Subjects walked and jogged along a 13 meter walkway at a self-selected speed which was kept to within 5% of the average speed determined during practice trials. Sagittal plane kinematics and kinetics were calculated using Move3D rigid body analysis software (NIH Biomechanics Laboratory, Bethesda, MD). EMG data were recorded from the tibialis anterior (TA), soleus (SOL), medial gastrocnemius (MG). vastus lateralis (VL), hamstrings (HS),and gluteus maximus (GM). Linear envelopes were produced (low pass, phase corrected, 2nd order Butterworth filter with 20 Hz cut off). Timing of muscle onset (>2.5 times resting) and peak activity were found with respect to initial contact (IC). The involved limb of the ACL deficient subjects were compared to the left limb of the control subjects. Data were analyzed using analyses of variance with repeated measures and post hoc paired t-tests ($p < 0.05$).

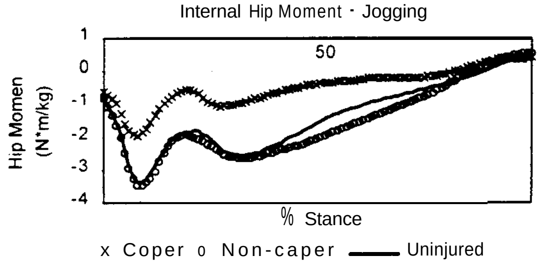

Figure I Internal hip moment. Copers have lower hip extensor moment at peak knee flexion than non-copers or uninjured subjects (p=O.O07).

RESULTS:

Kinetic and EMG data showed clear differences between copers and non-capers during both activities. In walking, immediately following initial contact (IC) copers had greater knee power generation (p=O.O25) than non-capers and uninjured subjects. Non-capers had delayed peak HS activity compared to copers and uninjured subjects (p=O.O5O). The peak HS activity occurred following IC in the non-capers only. At peak knee flexion, the hip extensor moment was lower in the copers than non-caper and uninjured subjects during walking (p=O.O24) and jogging (Figure 1). Both ACL deficient groups showed lower knee extensor moments (p=0.O09) during walking, but only non-capers had lower knee extensor moments during jogging (Figure 2). In jogging, EMG data showed trends in the copers toward early onset and more rapid onset to peak HS activity and longer duration onset to peak in GM (Figure 3). Non-capers had a trend toward early onset of VL and more rapid time to peak than uninjured subjects.

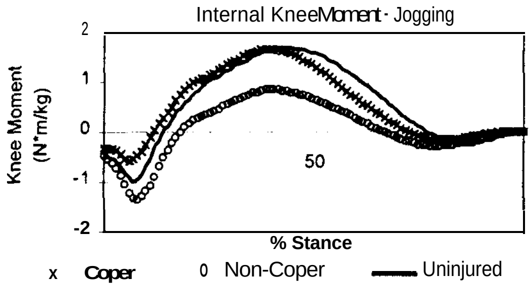

Figure 2. Internal knee moment, non-capers have lower knee extensor moments than copers and uninjured subjects (p=O.O66)

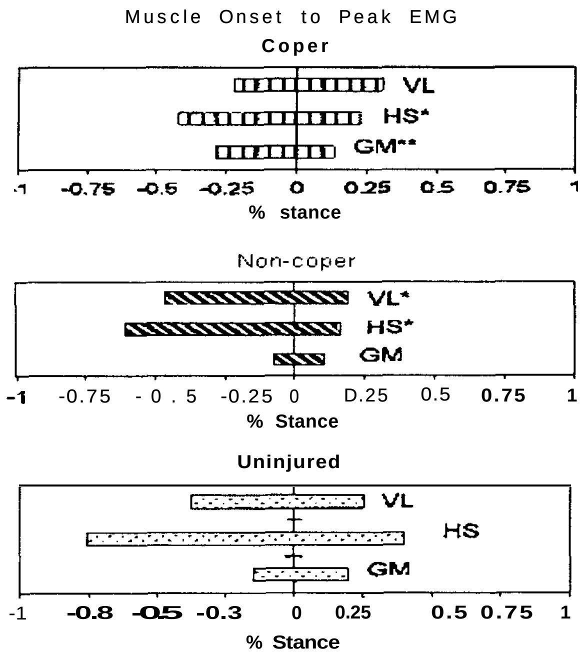

Figure 3 Muscle onset to peak EMG, Jogging. Trends showed both copers and non-capers had shorter duration from onset too peak EMG in the HS* (p=O.007). Copers had earlier onset HS* activity (p=O.073) and longer duration from onset to peak GM** EMG (0.089) than non-capers. Non-capers tended to have earlier onset VL* than copers (p=O. 166)

The total support moment was lower in the copers than in non-copers and uninjured subjects during walking (p=O.O21) and jogging (p=O.O25). The distribution of support moment about the hip, knee and ankle revealed a greater contribution of hip extensor moment to the total support moment in non-capers during walking (p=O.O40) and jogging (p=O.O07).

DISCUSSION:

The appearance of different movement and muscle activity patterns between uninjured subjects, copers and non-capers suggests different knee stabilization strategies in these ACL deficient subjects. The large knee flexor power generation in the copers during walking appeared in conjunction with peak hamstring activity which occurred prior to IC in the copers and control subjects and following IC in the non-capers. This appears to be a successful stabilization strategy used by the copers in order to allow more normal weight acceptance. During weight acceptance a strong quadriceps contraction could lead to anterior tibia1 translation in the absence of an ACL. Individuals with knee instability might attempt to control the limb by relying on the hip or ankle to support the body to a greater extent, thus avoiding the sensation of knee "givign way". This is indeed what is observed in the non-copers who had lower knee and higher hip support moments than copers or control subjects. The higher hip support moment in the non-capers appeared in conjunction with a more rapid time to peak in the GM and early onset of VL activity. The more rapid time to peak GM activity may indicate that the GM has insufficient time to contribute significantly to the hip extensor moment. This may lead the non-capers to use the biarticular HS muscles to control the hip rather than the knee, however, this can not be investigated without further data analysis involving the magnitude of the recorded EMG. Copers had lower hip extensor moments at peak knee flexion along with longer time to peak GM activity. This appeared along with delayed HS onset and more rapid time to peak in the HS muscles. This indicates a more precisely coordinated knee stabilization strategy involving the VL, HS and GM muscles. The GM may be providing control at the knee through its actions at the proximal femur. This could reduce the need for as forceful a quadriceps contraction and the concommittent anterior tibia1 translation which could follow. It appears that copers do indeed move differently than non-capers making it important to distinguish the functional abilities of ACL deficient subjects when selecting samples representing "all ACL deficient subjects". The copers in this study provide the unique opportunity to study a successful knee stabilization strategy in hopes of developing rehabilitation techniques to benefit individuals following ACL rupture and reconstruction.

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Contralateral Influence of a Healthy Quadriceps Muscle on the Maximal Volontary Isometric Contraction of the Atrophied Quadriceps

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INTRODUCTION: Orthopedic patients presenting a knee lesion often suffer from significant quadriceps weakness for they must wait a certain period of time before they are able to safely start a muscle strengthening regimen. During this inactive period, it is possible that strengthening the Contralateral muscle can have a positive effect on the affected muscle. The latter is known as cross education and was introduced more than a century ago in normal subjects (SCRIPTURE et al., 1894, and COLEMAN, 1969). In fact, the strengthening of a muscle in one extremity can directly affect the same Contralateral muscle without any direct strengthening. The fore-mentioned has often been explained by the Overflow theory (Henri and Smith, 1961). However, the cross education is rarely used in rehabilitation. In order to make it more applicable in clinical settings, a better understanding concerning the acute effects that the strengthening of the healthy quadriceps (HQ) has on the affected quadriceps (AQ) is needed. The goal of this study is to research the Contralateral influences, whether inhibitory or facilatory, on the maximum voluntary isometric contraction (MVIC) of the AQ.

METHODS: The control group was composed of 15 voluntary healthy men (27+/- 5 years). The experimental group was composed of 7 patients (35+/- 7 years) who underwent physiotherapy treatment at the Ottawa General Hospital. These subjects were also volunteers who recently suffered (81+/- 20 days) a knee injury. In order to participate in the study, the subjects had to meet several inclusion criterion (i.e. were relatively pain free, had quadriceps weakness, had more than 100 degrees of ROM).

The following bilateral measurements were taken:

-Thigh circumference (1 Ocm above the base of the patella)

- Q torque was measured at 90 degrees/sec and at 0 degrees/sec(MVIC) to 90 degrees by using the KIN-COM 550
- EMG activity of the biceps femoris muscle (BF) was measured (NEUROPACK 2 XPD, NICHON KOHDEN)
- Twitch Interpolation Technic: The M-response was recorded (lmsec square wave) on the femoral nerve in the inguinal groove during the MVIC of the Q (Morton, 1954).
- Tetanic electrical stimulation was measured (TES) (5OHz, 6 sec, symmetrical byphasic wave)

Contralateral stimulation of the HQ: The following tests were done in a random order with an average of 5 to 10 trials:

MVIC of the HQ; MVIC of the Hamstring muscle; M-reflex of the HQ; TES; H-reflex of the HQ; Patellar tendon reflex (PTR); Patellar tendon vibration (60Hz);Cutaneous stimulation (1.1 x sensitive threshold).

Statistics: Descriptive, student’s t-test, and the Wilcoxon non-parametric test was used for the smaller groups.

RESULTS: During the study, the patients ROM for the affected knee was 85%+/-5 of the healthy knee, the average circumference of the thigh decreased by 2.3+/-.5cm, and the average torque also decreased by 49.4+/-l 5%.

The MVIC of the ipsilateral AQ was not influenced by the following trials:

1) MVIC of the HQ (as in the control group)
2) MVIC of the Hamstrings (as in the control group)
3) M-response (as in the control group)
4) Vibration (as in the control group)
5) Cutaneous stimulation (as in the control group)
6) H-reflex of the HQ decreased by 8+/-4.5% (there was a significant decrease of 12+/-4% in the control group)
7) None of these stimulations modified the EMG activity of the ipsilateral BF

Contralateral inhibitory influences on the MVIC of the ipsilateral AQ:

1) PTR of 16+/-4 (19+/-4.2% in the control group)
2) TES was 9+/-3% (12+/-4% in the control group)

The following facilatory influences which arise due to ipsilateral stimulation affect the MVIC of the AQ:

1 M-response (twitch interpolation technic) increases the patients MVIC of the ipsilateral AQ by IO+/-3% (by 4.5+/-2% in the control group or in the patients HQ)
2) The ipsilateral TES increases the MVIC of the AQ by 15+/-4% (12+/-5% of the ipsilateral MVIC of the control group)

DISCUSSION: The Contralateral extremity of the HQ has an influence on the maximal ipsilateral recruitment of the AQ. These influences are inhibitory in nature and cannot be explained by the EMG activity of the antagonist muscle (BF). The latter are mostly proprioceptive influences (H-reflex and the PTR). These influences are not mechanical for the Contralateral M-response and the MVIC do not cause this inhibition. These results do not support the Overflow theory for it seems that there is an initial lack of recruitment (especially in the patients) or a deactivation of certain motor units during a few Contralateral stimulations. Only the ipsilateral stimulations (M-response and the TES) seemed to have facilatory influences on the MVIC of the AQ. In light of the results, it would seem inappropriate to prescribe a strengthening program for the healthy muscle in order to facilitate a contraction in the affected muscle. However, it does seem apparent that the lack of MVIC found in the AQ is partly due to the difficulties encountered when trying to recruit the motor units of the AQ.

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Human Soleus Muscle Phonomyogram During Twitches Evoked by Hoffmann Reflexes

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INTRODUCTION: Recent studies have demonstrated the reliability of skeletal muscle sounds recordings, or phonomyogram (PMG), as a good index of muscular force during twitches (4). Previous work (2) has indicated that muscle fiber composition influence PMG picked up from two different muscles. However, in this experimental situation we cannot distinguish the respective influence of muscle fibre composition and muscle geometry. How motor unit mechanical characteristics determine elementary PMG (3) remains unknown. In the present study, we bypassed this difficulty by recording PMG from human soleus during twitches evoked by Hoffmann reflexes. H and M responses, obtained during increasing stimulation intensities, give ((size principle)) and ((inverse size principle)) recruitment respectively, making possible to selectively record evoked PMG from either the smaller slowest soleus motor units or the larger fastest ones within the same muscle, without any regard for muscle architecture or dimensions.

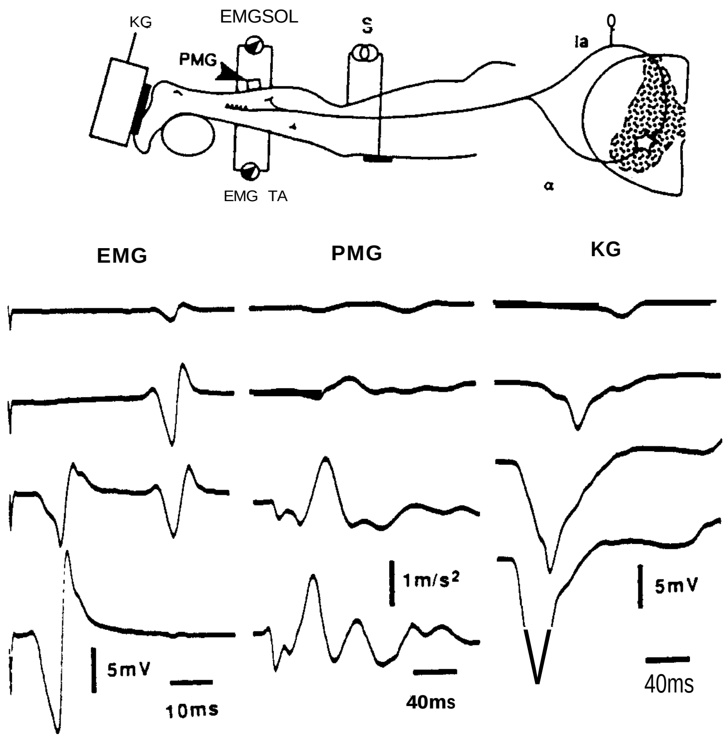

Figure 1 : Experimental set up. Below : raw EMGs, PMGs and KGs from one representative subject during electrical stimulation from threshold (upper traces) to maximal (lower traces).

METHODS: We recorded soleus electromyogram (EMG) and PMG in ten healthy subjects with surface electrodes (Dantec) and accelerometer (Nihon-Kohden) respectively, during twitches evoked by electrical stimulation single shock of the tibia1 nerve. We used also a kinemometer (KG) to record foot sole motion speed, giving the acceleration by calculating the initial slope of the signal. Electrical stimulation was performed with a cathod at the poplitea fossea and anod on the knee, with a constant current stimulator providing single shocks 0,l ms in duration, at various intensities (O-50 mA) from threshold H response to maximal M response. Signals were amplified, band-pass filtered (1 OHz-1kHz for EMG, lHz-100Hz for PMG and KG), and digitized for off-line amplitude measurements (peak-to-peak, root mean square). The initial slope of KG was calculated during the first 20 ms of the speed wave.

RESULTS: Figure 1 provides an example of typical EMG, PMG and KG traces obtained with increasing electrical stimulation intensities. For EMG, long latency H response firstly appears and progressively fades whereas short latency M response amplitude increases with intensity up to maximum. PMG amplitude increased during M responses mainly, whereas KG initial wave became steeper with stimulation intensity increase. The corresponding relationships (Figure 2, left column) between H (open squares) and M (dark squares) EMG responses and intensity allows to define the point (dot line) beyond which M response is higher than H response. PMG and KG initial slopes increase more or less linearly. Recruitment curves between PMG or KG and EMG (=H+M responses) show that the rate of increasing was higher after the critical point corresponding to H=M. For all the subjects, the slope values were significantly (p<O.OOl) higher for H<M than for H>M. It also indicated that PMG and KG were significantly correlated (p<O.Ol).

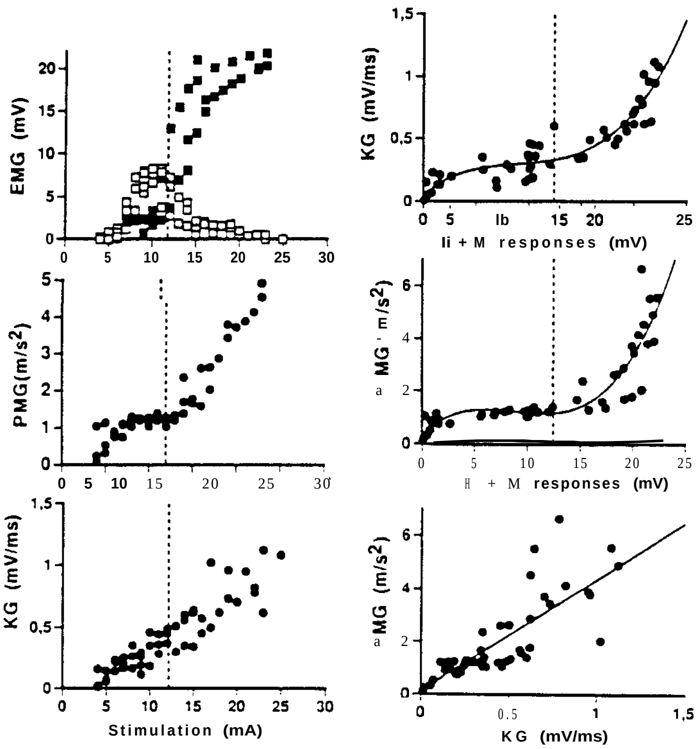

Figure 2 : **Left column** : recruitment curves for EMG, PMG amplitudes and KG initial slope for the same subject as in figure 1. The vertical line indicates the critical value of stimulation intensity corresponding to H=M. **Right column** : the two upper graphs show the PMG amplitude or KG initial slope over EMG (=H+M). The vertical line gives the EMG value for which H=M. The lower graph gives the PMG amplitude/KG initial slope correlation.

DISCUSSION: This study mainly demonstrates that the predominant type of EMG responses (M or H) during Hoffmann reflexes strongly determines the slope of the PMG/EMG relationships. We interpret these results as the consequence of the two motor unit recruitment principles, H responses corresponding to the slowest motor unit recruitment, whereas the fastest being recruited during motor evoked responses. Despite the fact that soleus is mainly composed of slow motor units, their contraction times are wide spread (1). PMG is thereby sensitive to motor unit mechanical characteristics. The significant increase of PMG/EMG relationship slopes with M responses also suggests that the fastest motor units produce higher elementary PMGs (3) than the slowest ones. Correlation between PMG and KG initial slope also indicates that PMG, namely lateral vibrations, reflect overall muscle mechanical change related to motor unit recruitment.

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Motor Performance After Non-Invasive, Selective Large-Fibre Deafferentation in the Cat

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Introduction

The role of sensory feedback in the control of movement can be examined by investigating the motor deficits that arise after deafferentation. Traditional deafferentation by dorsal rhizotomy is non-selective for different sensory modalities and is complicated by the possibility of compensatory adaptation during recovery from surgical trauma. Recently, we have used overdoses of pyridoxine (vitamin B,) to deafferent cats non-invasively and selectively for large sensory fibres. Excesses of pyridoxine have been reported to produce axonal degeneration of large sensory fibres in rat (l), dog (2) and human (3). However, the resulting motor deficits have not been investigated thoroughly. Here, we describe the motor abnormalities associated with pyridoxine-induced deafferentation and electrophysiological evidence for its large-fibre selectivity.

Methods

Eight male cats (3.5-4.5 kg) were trained to walk on an enclosed treadmill over a period of several weeks for food reward and verbal encouragement. Chronic recordings were taken before (d -3 to 0), during (d 0 to 3 or 4) and after (d 3 or 4 to 20) administration of py-ridoxine. A terminal electrophysiological experiment was conducted on day 20. The animals received 3 to 4 consecutive daily intraperitoneal injections of pyridoxine hydrochloride (350 mg/kg).

Kinematic recordings were collected during walking with and without harness support at 3 treadmill speeds (0.2, 0.4, 0.8 m/s) for 10-15, 4 sec trials at each speed. Reflective markers were attached to the shaved skin over a thoracic vertebra (Th4), shoulder, elbow and wrist of the right forelimb, and over the 7th lumbar vertebra, hip, knee, ankle and toe of the right hindlimb. The position of these markers was digitized in three dimensions at 60 Hz using a Kintrac Motion Analysis System. These data were analyzed off-line using commercial software (Kintrac) to generate files of three dimensional marker coordinates and to calculate basic parameters. In addition, kinematic recordings were taken for a 2 second episode centred around foot contact when animals landed from short falls (15 cm).

Terminal experiments were performed in a medial gastrocnemius (MG) nerve-muscle preparation under pentobarbitone anaesthesia. The left hindlimb was widely denervated except for the medial gastrocnemius and common and superficial peroneal nerves. A laminectomy was performed to dissect free the L7 and Sl dorsal and ventral roots. Compound evoked potentials (CEPs) were recorded from intact dorsal (DR) and ventral roots (VR) during graded electrical stimulation of the MG nerve and the superficial peroneal nerve. Longitudinal vibration (100 Hz, 100 micron) was applied to the MG tendon, while recording compound responses in dorsal roots. Single afferent units from MG were microdissected in dorsal roots and identified (as spindle Ia or II, Golgi tendon organ or group III or IV afferents) by axonal conduction velocity and characteristic responses to muscle stretch and whole muscle twitch. Monopolar recording and differential amplification was used for all root and nerve recordings.

Results

Following pyridoxine treatment, all animals developed an incapacitating motor syndrome characterized by unstable balance, crouched and unsteady gait or sheer inability to walk. Qualitative observations from withdrawal responses and measurements of tetanic force indicated that motor deficits were not due to a loss of strength arising from muscle or motoneurone dysfunction. Vestibular righting and roll responses were unaffected clinically.

Prominent features of ataxic gait after pyridoxine treatment include erratic coordination of limb and trunk movement in the frontal plane, which compromises balance, and poor coordination of fore- and hindlimb movement, most clearly seen in the sagittal plane. Frontal plane limb movement was analyzed by displaying wrist and hind-toe motion relative to trunk motion and by analyzing phase diagrams of relative toe vs. relative wrist motion. During ataxic gait, the toe-wrist displacement diagrams showed much increased individual variability of shape and occupied a much larger region in the toe-wrist displacement space.

Close observation of walking trials indicated that interlimb coordination was severely compromised after pyridoxine. Systematic measurements of forelimb cycle demarcation relative to current hindlimb cycle, expressed as a percentage of the normalized hindlimb step cycle, revealed strict phase locking between fore- and hindlimb for normal walking, as forelimb touchdown occurred between 20-50% of the hindlimb step cycle. In contrast, during ataxic gait, the coupling between fore- and hindlimb was strongly reduced, as revealed in much wider profiles of the cross-cycle phase histograms.

In response to short falls (15 cm), animals typically prepare for landing by fully extending all four limbs. Upon contact, they yield and absorb the inertial impact by partial flexion (100-150 ms) followed by moderate re-extension of the limbs. At the height of the motor syndrome both the duration and amplitude of the yield were significantly increased. The ataxic animals typically failed to absorb the inertial impact and landed on their trunk, unable to stand.

Compound evoked potentials (CEP) recorded in terminal acute experiments were normal in ventral roots but strongly reduced in dorsal roots. Antidromic ventral root CEPs, elicited by medial gastrocnemius (MG) nerve stimulation at 7x a-threshold had the same average latency and amplitude in pyridoxine treated and normal animals. Tetanic forces after pyridoxine were in the normal range. In contrast, orthodromic dorsal root CEPs elicited by MG or superficial peroneal (SP) nerve stimulation showed nearly complete abolition of the short latency components. Compound potentials elicited by vibration were abolished after pyridoxine treatment. Thus, on CEP evidence, pyridoxine intoxication caused a selective deficit in sensory axons, probably exclusively in large sensory axons.

Afferent units from the MG muscle were detected in fine dorsal root filaments by their response to stretch or electrical stimulation of the nerve. The goal was to isolate as many single units as possible to determine afferent fibre conduction velocity (CV) spectra. In pyridoxine treated animals, few units and practically no group I afferents, but appreciable numbers of group III and IV afferents, could be isolated. Thus, single unit recordings confirmed the occurrence of selective large-fibre deafferentation after pyridoxine. The extent of large fibre deafferentation varied between animals and appeared to be correlated with the severity of the clinical syndrome.

Discussion

Deafferentation by pyridoxine intoxication has major advantages compared with the traditional method of dorsal rhizotomy: it is non-invasive and has rapid onset, ruling out that sensorimotor deficits are masked by compensatory adaptation during recovery from surgical trauma.

The present observations provide the first direct electrophysiological evidence at single unit level that pyridoxine intoxication causes a selective large-fibre sensory neuropathy. Motor deficits can be attributed to sensory loss, since motoneurone function is not affected.

Acute large-fibre deafferentation by pyridoxine causes incapacitating motor deficits, including ataxia, locomotor inability and failure to adjust to sudden disturbances. The severity of the motor syndrome is much more significant than what has previously reported for deafferentation by rhizotomy (4). One possible reason is that, owing to the non-invasive nature of the intervention, motor function could be monitored immediately after treatment.

The disruption of interlimb coordination during weight bearing locomotion reveals a new role of large fibre sensory feedback in motor pattern generation as, traditionally, the coordination between forelimb and hindlimb movements is thought to be mediated by propriospinal circuits.

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Excitability of Ia Pre-Synaptic Inhibitory Pathways after the Application of Pressure Around the Leg

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INTRODUCTION: The application of pressure over a muscle tendon or belly has been shown to reduce motoneuron reflex excitability (MNE) in both healthy subjects and subjects with cerebral vascular accident (CVA) and spinal cord injury (SCI).[1-2] Although the central nervous system mechanism for this decrease is unknown, it is suggested that the mechanism responsible for reducing MNE following pressure application is spinal in origin.' This conclusion is based on the result that similar decreases in MNE occur in subjects with SCI when compared with healthy subjects and subjects with CVA Pre-synaptic Ia inhibition is one spinal cord inhibitory mechanism that may account for the decrease in MNE. This study investigated the effect of Ia pre-synaptic inhibition (IaPI) on the soleus H-reflex after the application of continuous circumferential pressure to the leg. SUBJECTS: Thirty-eight volunteers with no history of neurological disease participated. METHODS: For each subject, thirty 70% maximum H-reflexes were elicited by tibia1 nerve stimulation. These reflexes were recorded and their mean served as a baseline measurement ($H_{baseline}$). In the experimental phases ($H_{pre-pressure}$, $H_{pressure}$, $H_{post-pressure}$), soleus muscle IaPI was achieved by vibrating the tibialis anterior tendon with an amplitude of <lmm at 120 Hz for 1 Oms. Forty to 60 ms after vibration, the tibia1 nerve was stimulated and 30 H-reflexes were recorded.' During the $H_{pressure}$ phase, a pressure splint was inflated and maintained at 35-45mmHg. The H-reflex amplitudes from the experimental phases were compared with H-reflex baseline measurements and expressed as a percentage (i.e., $H_{pre-pressure}/H_{baseline}$ x 100). During the experimental phases, random checks of $H_{baseline}$ were performed to confirm $H_{baseline}$ did not change over time. RESULTS: Ten subjects were omitted from data analysis due to M wave configuration change or instability of $H_{baseline}$ amplitude. Data analyses were done on 28 subjects. A visual inspection of the $H_{pressure}$ data revealed two groups when compared with $H_{pre-pressure}$ values: subjects that exhibited an increase in IaPI with pressure (group 1) and subjects that exhibited a decrease in inhibition (group 2). Paired t tests (with Bonferroni' s correction factor for multiple comparisons) were used to compare $H_{pressure}$ group data with $H_{pre-pressure}$ group measurements ($p \leq .025$). Both $H_{pressure}$ groups were shown to be significantly different from $H_{pre-pressure}$ group values (figure 1). Multiple regression analysis showed no significance for the variables H/M,,, ratio. $H_{baseline}$, $H_{pre-pressure}$, gender. and age. CONCLUSIONS: These results suggest that pressure applied around the calf. does affect Soleus IaPI. however the effect varies. Unfortunately. the direction of IaPI response could not be predicted for any given subject. suggesting other individual differences not studied in this investigation caused the variable response. Type II error cannot be ruled out. however. since an n=28 may not have enough statistical power to do the multiple regression analysis. Lastly, this study' s results suggest inhibitory mechanisms besides IaPI are also involved in lowering MNE after pressure application. This conclusion is derived from research that shows Soleus MNE decreases in all subjects tested after pressure is applied around the leg. [1-2]

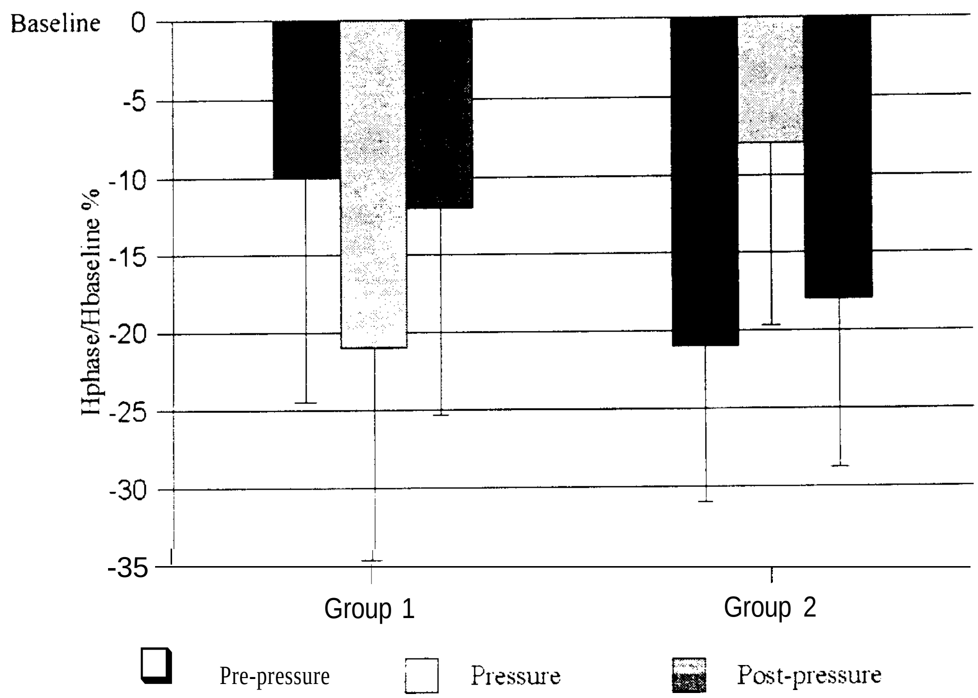

Figure 1:Mean percent change across test phases for each subject group. Post-pressure values were not used in the statistical analysis.

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Inhibitory Effects of Femoral Nerve Stimulation on Reflex and Voluntary Activity of Human Soleus Muscle

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Introduction

Sensory afferent information influences the excitability of homonymous motoneurones as well as heteronymous motoneurones. This short latency modulation of spinal pathways could be part of a basic coordination mechanism used in daily postural or locomotor activities. The goal of this study was to determine the intersegmental influence of sensory afferents from the knee extensor muscles onto an ankle extensor muscle (soleus). It has been previously documented in man, that a conditioning stimulus applied to the femoral nerve (FN) results in a strong inhibition of the soleus H-reflex (Meunier et al., 1990). However, the constancy of this finding in different postural positions and the influence of FN stimulation on a voluntary descending motor command onto the soleus motoneurones have not been previously investigated.

Methods

Fifteen subjects (27±4 years old) without orthopaedic or neurological deficits participated in this study. The effects of an electrical stimulation of the FN (conditioning stimulus:CS) on the amplitude of the soleus H-reflex (test response) and on the integrated EMG activity during voluntary contraction of the soleus in sitting and standing were studied. The CS had a fixed duration (0.5ms) and intensity (1 .O x quadriceps motor threshold). For the soleus H-reflex experiments, the reflex was evoked by stimulation of the PTN (0.5 ms duration) in the popliteal fossa at an intensity adjusted to elicit H-max/2. The conditioning - test intervals (CTI) varied from -6ms to 100 ms. The effect of varying the CS intensity (from quadriceps H-reflex threshold to H-max) was also investigated. Fifteen conditioned responses were compared with 15 tests (non conditioned) responses for each CTI. For the voluntary contraction experiments, the EMG signal was digitized at 5000 Hz after amplification and high and low passed filtering (1 Hz to 1 KHz) and recorded from 100 ms before to 400 ms after the CS. The integrated EMG activity within windows of 25ms were compared before (baseline) and up to 140ms after the CS. In sitting the subject had to press on a foot pedal to produce a baseline EMG level similar to the standing condition.

Results

The inhibitory effect started at a FN stimulation intensity corresponding to the quadriceps H-reflex threshold and maximal inhibition was reached at an intensity equal to Hmax/2. The following characteristics of the inhibition were the same for the reflex or voluntary contraction studies. 1) The central latency of the inhibition was less than 5 ms. 2) In sitting, the duration of the inhibition (- 50ms) was longer than in the standing condition (-30ms). 3) The amount of inhibition (40-60% of control values) was similar for the sitting and standing conditions within the first 25ms of inhibition. 4) The inhibition was absent when the position of the cathode was moved off the FN to produce a pure cutaneous CS with a similar sensation. Finally, stimulation of the FN of a deafferented patient deprived selectively of large sensory myelinated fibres but with intact motor fibres (Forget and Lamarre 1995) failed to inhibit the voluntary activity of the soleus muscle.

Discussion

It occurred in every subject tested and was very effective to decrease either the reflex or the voluntary contraction of the soleus. The inhibition of the soleus activity by quadriceps afferents was **also very** strong. The amplitude of the soleus H-reflex was decreased by 4060% and the integrated muscle activity of the soleus by approximately the same amount during voluntary contraction. In a previous report Pelletier et al., (1992) showed that the inhibition of the FN on the soleus H-reflex is 3-5 times stronger than the "reverse pathway", that is inhibition of the quadriceps H-reflex by posterior tibia1 nerve afferents.

Preliminary speculations as to the mechanisms involved can be make. The parallel characteristics between the inhibition of the reflex and of the voluntary activity point to a similar mechanism for the two phenomena. The duration of the inhibition (- 50 ms) and its efficacy to decrease the activity of a descending motor command does not favour the contribution of presynaptic inhibition of sensory afferents onto soleus motoneurones. Moreover, the absence of the inhibition in a deafferented subject tend to eliminate the contribution of heteronymous Renshaw activity (Hultbom et al., 1968) that should have been triggered by antidromic motor volleys in the femoral nerve. Finally, failure of the skin stimulation to produce the inhibition also eliminates the contribution of cutaneous afferents. The strength of the stimulation necessary to trigger the inhibition and the increasing inhibition with increasing stimulus intensity point to a mechanism that could be different than the one mediated by large Group I muscle afferents.

Conclusion

The results show a strong inhibitory influence of the quadriceps afferents onto the soleus muscle. This phenomena was present in all subjects in different postural conditions and was equally effective to decrease either reflex or voluntary activity of the soleus muscle. Even though the mechanisms involved are not yet well understood, the investigation of this inhibitory phenomena in patients with neurological deficits and showing co-contraction of the knee and ankle extensors could lead to a better understanding of the incoordination and pathological synergies observed in these patients.

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The Muscle Contraction Property and Repetitive Magnetic Stimulation

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INTRODUCTION

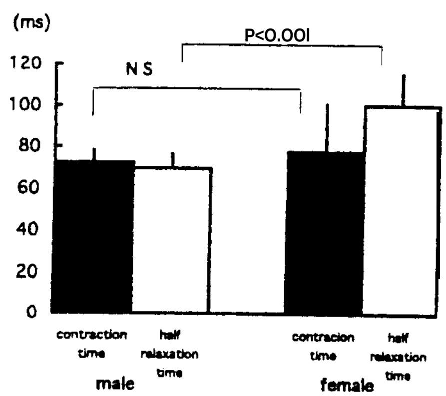
The pattern of muscle contraction have been ever studied, using electrical stimulation or voluntary movement. This time, we studied the pattern of human biceps muscle and the nature of muscle using magnetic stimulation.

METHODS

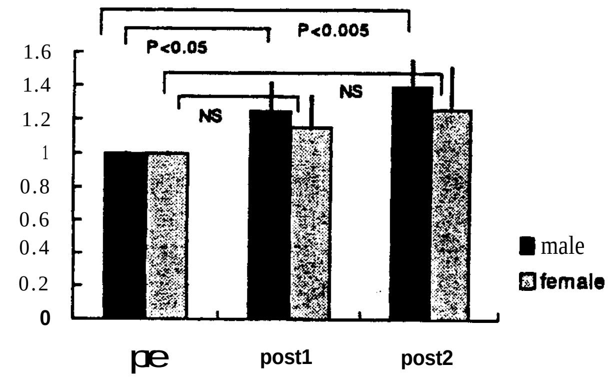
The subjects were 6 male volunteers and 7 female volunteers without neuromuscular disease. The mean of their age was each 24.8 years in male and 23.9 years in female. As a method, the forearm of sitting volunteer was fixed on the plate of strain gauge and was stimulated on the center of biceps muscle by magnetic stimulation. The output of magnetic stimulation increased each 10% from 40% to 100%. We measured contraction time,half relaxation time and twitch tension of single twitch by magnetic stimulation. Next, we elicited tetanic contraction by repetitive magnetic stimulation and studied the ratio of twitch tension, the single twitch after tetanic contraction to the single twitch before tetanic contraction. We used t-test to analyze the data.

RESULTS

The output of magnetic stimulator in maximum twitch tension was each 65.0% in male and 85.7% in female. The mean of contraction time in maximum twitch tension was each 76.3ms in male and 77.7ms in female. The mean of half relaxaion time in maximum twitch tension was each 69.7ms in male and 100.6ms in fmale(fig. 1). We found the significant difference between male and female in half relaxation time. Next,using repetitive magnetic stimulation, we measured the ratio of twitch tension. The twitch tension after tetanic contraction was increased cornpairing with the twitch tension before tetanic contraction(fig.2).



(fig. 1)The mean of each contraction time and half relaxation time in maximum twitch tension were illustrated



(fig.2)The ratio of twitch tension $\left[\frac{\text{twitch tensioin after tetanic contraction}}{\text{twitch tensioin before tetanic contraction}}\right]$
Was affected by tetanic contraction.
pre:twitch tension ratio before tetanic contraction
post l.twitch tension ratio after first tetanic contraction
postZ:hvitch tension ratio after second tetanic contraction

CONCLUSIONS

The output of magnetic stimulator to maximum twitch tension was different between male and female. It was thought to be related to the ratio of slow twitch units(1,2,3). Then, female would have more slow twitch units in biceps muscle compairing with male.

The half relaxaion time was affected by the ratio of slow twitch units, and it was different between male and female. The contraction time was not different between male and female. It was thought the contraction time might be affected by the presence of fast twitch units(4,5). The effect of tetanic contraction to single twitch was examined and was found amplification of twitch tension of single twitch after tetanic contraction. This was thought to be the post-tetanic potentiation(6).

Therefore, it is thought that the pattern of muscular contraction elicited single and repetitive magnetic stimulation will be able to measure the functional nature of human proximal muscles measuring contraction time, half relaxaion time and twitch tension.(7,8)

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The Effect of the Repetitive Magnetic Stimulation

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INTRODUCTION

In our previous study, single transcortical magnetic stimulation on the small animal induced the transient reversible change in several neurotransmitters. But the behavior of them was not affected significantly.

In this study the repetitive magnetic stimulation (rMS) was studied to observe the change of the behavior in normal mice and ataxic mice.

METHODS

The mice were composed of 32 normal male mice and 29 cytosine arabinoside injected male mice. The cytosine arabinoside 50 mg/ Kg was injected subcutaneously one a day for 3 days from 2 days to 4 days after birth of mice. The cytosine arabinoside injected mice showed the marked ataxic gait with falling in 2-3 weeks after injection. Repetitive magnetic stimulation (rMS) was given to the mice transcortically in 130% strength of cortical threshold, 5 days a week for 3 weeks. One group (11 normal mice and 10 ataxic mice) got the rMS in 10 Hz * 100 times. Another group (10 normal mice and 10 ataxic mice) got the rMS in 3Hz X 180 times. The third group (10 normal mice and 9 ataxic mice) did not get the rMS. The behavior of mice was observed by open-field method in 8 weeks after birth, just before the mice received the rMS. The second test was studied in llweeks, just after the mice finished to receive the rMS. And the third test was studied in 13 weeks. In open-field method, the mice moved in square cage in 7 cm X Sun (length X width). When they moved one cm in each direction, we counted it as its counts of movement. The falling ratio in ataxic mice was calculated as the ratio of the counts of falling to the counts of movement. During these procedure we did not observe the loss of conciousness or convulsion in any mice.

RESULTS

The movements of mice were not influenced by rMS statistically both in normal mice and ataxic mice, although some tendency to decrease in movement was observed in ataxic mice in 11 weeks, but not in 13 weeks, compaired with normal mice.

The falling ratio was decreased in 11 weeks and 13 weeks in both rMS ataxic mice and control ataxic mice. The mean value of the falling ratio of rMS (10Hz X 100) ataxic mice (35.7%) decreased more than it of control ataxic mice (59.2%) in 11 weeks. This tendencypersisted in 13 weeks, rMS (10HZ X 100) mice 40.5% vs control mice 49.0%. And rMS (3Hz X 180) ataxic mice showed its mean value between rMS (10Hz X 100) ataxic mice and control ataxic mice.

DISCUSSION

The low frequency electrical stimulation could make the kindling phenomenon, which was depending on the duration, strength and frequency of electrical stimulation. We used the 3Hz and 1OHz repetitive magnetic stimulation, which was low frequency stimulation. In our study each stimulation made the twitch in all muscles. But we did not observe the loss of conciousness or generalized convulsion after rMS.

In our previous study, magnetic stimulation could induce the marked changes of neurotransmitter in dopamine, serotonine and noradrenaline metabolism. In this study as the method of stimulation was the same as before, we believed the brain of the mice was stimulated by rMS. So I might suspect this type of rMS did not induce the convulsion like kindlingphenomenon, which was studied jigorously before. But this rMS might induce the effect on the behavior.

When cytosine arabinoside was injected to neonatal mice, the depletion of granule cell and functional glia was reported and clinically ataxia was observed. Our pathological preparation showed both the granule cell and Purkinje's cell were depleted, and the cell layer changed. These changes were anatomical changes, and we could not observe the gross anatomical difference between rMS mice and non-rMS mice. The decrease of falling ratio after rMS was recognized to be functional changes or synaptic changes, which would be induced by rMS.

These data need to study further, but they will help us in using the rMS to human as the method of investigation and treatment.

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Neural Control of Human Walking

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Introduction

The control of human walking by the nervous system is not well understood. While walking has been well described, the nervous pathways and neural mechanisms which produce the observed walking are obscure. Historically, study of the neural control of locomotion has focused on animals with a nervous system simpler than ours (e.g., dogfish, lamprey, tadpoles, salamanders, chick, rat, cat and dog, reviewed in Grillner & Wallen 1985; Grillner & Dubuc 1988; Rossignol 1996). In some cases, they have identified the neurons involved and their interconnections (e.g., reviewed in Grillner et al. 1995). There is a general expectation, among neurophysiologists, that many of the **structures** and mechanisms that control animal locomotion will control human locomotion as well. There is also reason to believe, however, that the control of human locomotion will differ in some way from other animals - after all, humans are bipedal, and they have a more complex nervous system.

The main obstacle to studying the control of human walking has been technical - the methods for studying neural control in animals are too invasive to use in the human. We have recently developed some strategies to get around some of these problems. These methods allowed us to clarify where animal and human control are similar, and where they are not. In this paper, I use three examples from work in our group, to show that we can gain considerable insight to the way the nervous system controls walking in the human. Insights gained from these studies together with parallel experiments in individuals with pathology help us understand the nature of changes that occur with pathology.

How is sensory input controlled during human walking?

Sensory information from the periphery is carefully controlled during walking. It contributes to walking when it is important, but not when it could interfere with progression (e.g., Capaday & Stein 1986; Crenna & Frigo 1987; Yang & Stein 1990; Duysens et al. 1990; Dietz et al. 1990). The flow of sensory input during walking, as measured by the magnitude of specific reflexes, is altered in some pathologies (Fung & Barbeau 1990; Yang et al. 1991; Jones & Yang 1994). Alterations in the control of sensory input presumably contribute to the walking problems experienced by these individuals. What are the neural mechanisms used to control the flow of sensory information during walking?

<u>a) Presvnaptic inhibition.</u> Presynaptic inhibition of primary afferent fibres is one way in which sensory information can be gated during walking in the cat (reviewed in Rossignol 1996). Presynaptic inhibition is estimated in animals by recording from cut dorsal roots, intracellularly from individual primary afferents, or from peripheral nerves during intraspinal stimulation, all of which are invasive. Capaday & Stein (1987) showed that, in the case of a monosynaptic reflex (such as the H-reflex), the postsynaptic excitability of the motoneuron pool can be controlled by matching the background EMG at the time the H-reflex is elicited. Under conditions of matched background EMG, if the amplitude of the H-reflex varies (for example, when performing different tasks), then the change is likely a result of presynaptic inhibition (reviewed in Stein & Capaday 1987). Within a walking cycle, the H-reflex of the soleus muscle increases in amplitude roughly in parallel with the activation of the

soleus muscle during the stance phase, and is completely suppressed during the swing phase (Capaday & Stein 1986). We wished to determine if the modulation within a step cycle is also a result of presynaptic inhibition. To do this, we used the fact that human subjects can learn to change their muscle activation in a task. Using the principle of matching the background activation of the muscle, we trained subjects to walk with activation of the soleus muscle during parts of the swing phase. Movement of the ankle joint was limited by a custom fitted splint. Soleus H-reflexes were elicited while the subject walked, as previously reported (Capaday & Stein 1986). We compared the H-reflexes elicited in the stance and the swing phases, at times when the background EMG levels were matched. The amplitude of the H-reflex remained very low in the swing phase compared to the stance phase (Yang & Whelan 1993). Thus, we concluded that depression of the H-reflex during the swing phase was a result of presynaptic inhibition.

In separate studies, we and others reported that in patients with spasticity (i.e, from traumatic incomplete spinal lesions), the soleus H-reflex is abnormally active during walking. In some patients, the reflex was very active throughout the walking cycle, with no inhibition during the swing phase. Since presynaptic inhibition in likely responsible for the inhibition during the swing phase, this mechanism may be altered in these patients. Fung & Barbeau (1994) have taken this further, and found a way to depress the hyperactive reflex during walking, at the appropriate times, using stimulation to the foot.

b) Gating of narallel pathwavs. Another form of reflex modulation during walking is characterized by changes in the direction of a response. For example, innocuous stimulation of certain cutaneous afferents in the foot leads to excitation of the tibialis anterior (TA) muscle in one phase of walking (swing) and inhibition in another phase (transition from swing to stance) (Yang & Stein 1990). Does this change in the direction of a response represent separate inhibitory and excitatory pathways to different motoneurons of a muscle (e.g., motoneurons that are active at different times in a step cycle)? Alternatively, do the pathways project to the same motoneuron, with the strength of the pathways modulated as a function of the walking cycle? Patients with incomplete spinal cord injuries showed predominantly excitatory responses to the same stimuli (Jones & Yang 1994). Understanding the mechanisms responsible for the normal modulation will assist us with determining the mechanisms responsible for changes in pathology. Three hypothetical configurations of reflex pathway could generate the responses recorded from healthy individuals. In animals, intracellular recording from motoneurons can distinguish between these possibilities (e.g., Schomburg et al. 1981). In humans, we determined the mechanism responsible by recording from single motor units during walking (De Serres et al. 1995). Fine-wire electrodes were inserted near the origin of the TA muscle. Movement of the muscle was limited by having subjects wear a custom-fitted splint. The activity of single motor units during undisturbed walking, and the response of units to stimuli applied to the posterior tibial nerve at the ankle were studied. The results indicated that the majority of motor units were active during both phases of TA activity. Moreover, in response to a stimulus, the motor unit activity was enhanced in one phase and suppressed in the other. Thus, the reflex-reversal was most likely a result of parallel excitatory and inhibitory pathways to the same motoneurons, which were alternately favoured during different periods of the walking cycle. Modulation of the reflex was present in most spinal cord injured subjects, but a predominance of the excitatory response was seen (Jones & Yang 1994). This would suggest that the abnormality may be an increase in the excitability of the excitatory pathways, rather than a problem with a specific population of motoneurons.

Is sensory information used to control the transition between the stance and the swing phase of walking in the human?

In the cat, sensory information from the periphery is important for controlling the transition from the stance to the swing phase. Two types of input are particularly important: 1) Extension of the hip in the stance limb (e.g., Grillner & Rossignol 1978), and 2) unloading of the stance limb (e.g., Duysens & Pearson 1980). Roth are needed for the limb to advance to the swing phase. Roth seem considerably less important for the transition from stance to swing in the human. For example, humans can clearly walk with large degrees of hip flexion, such as when walking through a low tunnel. Thus, hip extension is not a necessary condition for walking in the intact human. Our recent work suggests that load has a very small effect on the duration of the stance phase (Stephens & Yang 1996 a&b, Stephens et al. in press). Do these differences truly represent differences between the species?

One issue that remained unresolved from the earlier work was the difference between the preparations (e.g., decerebrate or spinal cats versus intact humans). It is possible that the underlying brainstem/spinal control of walking is similar between cats and humans, and that the differences represent the influence of the cerebrum. Thus, we sought to develop a way to study the human under conditions that were more analogous to the decerebrate cat. Human infants show a walking response at birth. This walking is thought to be largely independent of the cerebrum (Forssberg 1985). Thus, infants offer a way to study the brainstem/spinal control of walking in the human.

<u>Development of the infant model.</u> Initial work addressed the feasibility of using infants to study the sensory control of walking. Study of the newborn is very difficult, because while the stepping response is strong at this age, the infant is rarely in the appropriate state (awake and alert) for studying walking. Typically, infants lose the walking response between the ages of **2** to 7 months, a time that would be ideal to study the walking response, because the cerebral influences are likely still minimal but the infants are more alert and likely to walk. In the first study, we established that with brief periods of practice, infants can retain the walking response indefinitely, in support of Zelazo's early report (1972). Moreover, the stepping is sustained and repeatable, conditions that are necessary for studying the response to sensory disturbances (Yang et al. 1998).

<u>Does loading control the transition from the stance to the swine phase in infants?</u> Infants were supported to walk on a treadmill. Load was added transiently during the stance phase at random times. EMG's, foot contact, and video were recorded. In a few experiments, the load added was estimated with a force platform imbedded under the treadmill belt. Increasing the weight support by 22% of body weight on average resulted in a prolongation of the stance phase (increase of **30%** on average). The step cycle was also lengthened **(28%),** resulting in a resetting of the walking rhythm (Yang et al. in press). These results are in direct contrast to those in adults, where a similar amount of loading **(30%** body weight) resulted in a minimal change in the stance phase (7%) and step cycle (3%) (Stephens & Yang 1996b). Thus, load during the stance phase has an important contribution to the transition into the swing phase in the infant. In this case, the behaviour of the infant resembled that of the decerebrate cat. Much remains to be done to elucidate the neural mechanisms that control infant walking. Preliminary findings from this and other studies suggest that there are remarkable similarities in walking between the decerebrate cat and the human infant.

Summary

Many mechanisms for the control of walking appear to be conserved across the species, and much of what we know about other mammals may be transferable to the human. The human nervous system is considerably more complex, however. Evolutionary changes in the control between the carnivores (cats/dogs) and the human will be important to determine. These three examples illustrate that it is possible to study specific neural mechanisms in the human. It is also possible to studythe brainstem/spinal control of walking in the human, using the young infant. I believe there is a large potential of yet untapped territory for study in the human. This work will be essential for the development of better methods for retraining of walking after lesions to the central nervous system.

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Postural Adjustments and their Cognitive Appraisal Resulting from Gravito-inertial Force Changes in Standing Man

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INTRODUCTION

When an observer is submitted to changes of gravito-inertial forces such as those occurring in cars or planes or in various jumping or sliding sports, postural reactions develop to maintain or reset balance and equilibrium. Behavioral observations and verbal reports from plane pilots and from various athletes suggest that behavioral and cognitive errors may yield hazardous situations (loss of equilibrium and hand, head and eye movement inaccuracy) in tasks where pre-movement postural adjustments are critical for motor action. As a first step to analyze these problems, we have studied postural changes (behavioral) and their perceptive (cognitive) correlates in observers standing in an eccentric position on a plateform rotating around a vertical axis.

METHODS

Figure 1 describes the main features of the experimental apparatus. The observer stood in a radial plane, 33 cm from the center of rotation. The plateform was driven by a servo motor. The observer's task was to maintain his/her equilibrium while the plateform was set into rotation and brought slowly to the velocities tested in succession. The rotation occurred in darkness except for red diodes organized in pairs and used to code the observer's percept of the situation. While posturally stable at a given rotation velocity, the observer indicated successively his/her sense of subjective vertical (Mittelstaedt, 1995 a,b), body and plateform orientation. For this, the observer oriented a rod placed in front of him and carrying a red LED at each end. The rod was made to rotate around its center by a motor activated by a joystick placed at arm level in front of the observer.

The postural adjustments resulting from the rotation-induced gravito-inertial force changes were measured with a video device. For this, red LEDs were attached to the observer to monitor head and body inclination in the frontal and in the sagittal planes. LEDs positions before, during and after rotation were determined by means of a movement analysis program (ARIEL) from selected video frames for each conditions. From the Cartesian coordinates of the various points provided in absolute scale in the working space calibrated at the beginning of each experiment, various vector orientations and body segment position changes were calculated using Microsoft EXCEL. The eight observers were students from the Faculty of Sports Sciences, ranging from 22 to 30 years of age, 50 to 75 Kg of weight and 1.50 to 1.80 m of height. They gave formal consent to participate to the study.

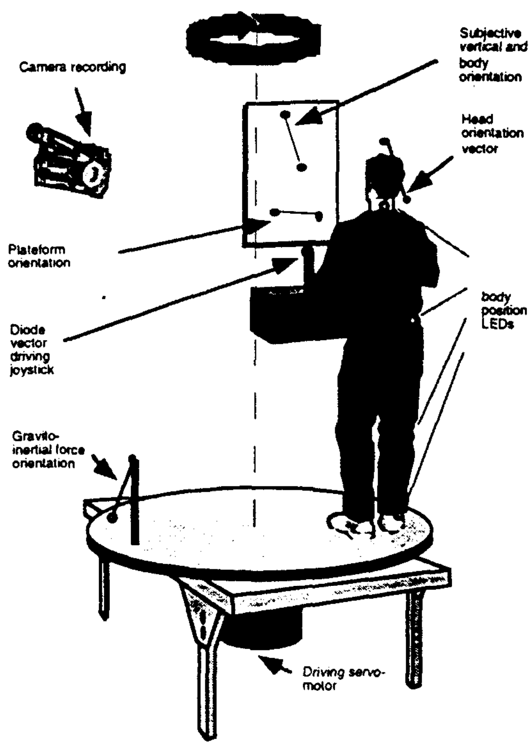

Figure 1 : Experimental setup showing the main arrangements for monitoring observers' postural changes and related cognitive correlates.

RESULTS

All tested subjects maintained proper postural equilibrium for the three tested rotation velocities (80, 120, 150°/s) but reported to have experienced difficulties, and a related fatigue, to develop the strength required in the outer leg. All subjects reported circular vection (an illusory sense of rotation, directed opposite to the real rotation), starting near the end of the deceleration period and lasting several seconds after the end of the rotation if the subject was maintained in darkness but ended immediately once the room was illuminated. All subjects made errors in their appreciation of the various subjective orientations they were supposed to code, especialy at the two highiest tested velocities. In particular they reported an erroneous sense of inclination, with respect to the horizontal, of the platefotm on which they were rotated. At the highest tested velocity (150°/s) all subjects reported to sense the plateforme as being inclined outward (as if standing on a conical surface) although all were aware that the plateform could not take that inclination. The readings show that on the average, the observers made an error of 1.7°, 3.3" and 10.5" in their appreciation of the subjective vertical (the error was in the direction of the orientation of the new gravito-inertial force) at 80, 120 and 150°/s, respectively. The observers sensed their body as being slightly but consistently more inclined towards the center of rotation that their actual leaning. The error was, for the same velocities, 2,4.8 and 6°, respectively. At the tested velocities, the observers felt the plateform as being inclined outward by 1.7, 3.4 and 1 1°, respectively. Average postural readings show that the observers were less inclined than what was requested for a centering of the direction of the new gravito-inertial force between the two feet (1.4 vs 3.5, 3.2 vs 7 and 5 vs 14°, at 80, 120, 150°/s, respectively). However, the head was more inclined than the body and sensibly aligned with the direction of the new force.

DISCUSSION

The data definitely show that observers can maintain appropriate postural equilibrium during transient and sustained rotation of a plateform while standing in an eccentric position. In our experiment, in spite of absence of visual control (except for small LEDs), the created tengential force, likely to disrupt the equilibrium, was fully compensated for by an inclination of the body towards the center of rotation, and an increase of force applied to the outer leg. One may also assume an increase of tension in the upper skeletal structure of the body, opposite to the outer leg so as to not only prevent leaning outward but also create body inclination toward the center of rotation. However, the equilibrium thus reached, although adequate to allow the observer to remain stable, was not achieved with a full centering of the direction of the compound gravito-inertial force between the two feet, but rather nearer the outer edge of the sustentation base. This resulted in a severe unbalance of the forces applied to the legs, responsible for the reported fatigue in the outer leg. The sensing of this unbalance, similar to the one that may be sensed while standing on an incline boad, may be responsible for the illusory sensing of being standing on a conical plateform. The overall underinclination of the body may be the result of the over inclination of the head (we have not attempted to determine the sensing by the observer of his sense of orientation of his head, which we will do next to solve that problem), or an inappropiate sensing by the Vestibular system of the tangential force applied to the body. Indeed, as the velocity increases, the body leans towards the axis of rotation with a resulting decrease of the distance between the actual position of the head and the center of rotation. The sensed centrifugal force by the Vestibular receptors is then underestimated with respect to the actual force distributed along the body, with an increasing factor from the head down to the feet. Other sources of information besides visual input are available particularly from neck trunk and leg muscle proprioceptive afferents and related efferent copies and from pressure receptors distributed under the feet and in the joints.

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Aging and Postural Control: Postural Perturbations Caused by Changing the Visual Anchor

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INTRODUCTION

With aging, the relative contribution of visual information for calibrating the body in space increases and elderly individuals are more dependent upon visual information for stabilizing their posture.’ When using elevator, the opening of the doors modifies rapidly the visual stable anchor. Such a change in the visual anchor could present a risk factor for falling because it induces a perceptual illusion of backward movement. The aim of this experiment was to determine if modifying the visual stable anchor by opening doors similar to that of an elevator, perturbs the postural stability of elderly persons and presents an important risk factor for falling. A biomechanical model using the inverted pendulum analogy will be presented to illustrate the importance of these effects.

METHOD

Eight elderly dwelling subjects without history of frequent falls (mean age = 69.8 years) and nine young subjects (mean age = 23.1 years) participated in this experiment. Sensory and clinical measurements were used to insure that all elderly individuals were free from any pathology affecting postural stability. Subjects were positioned barefoot with their feet IO-cm apart within a stable elevator cage and the visual environment was modified by opening the elevator doors within 2 to 5 s after the beginning of a trial. All trials included at least 4-s of data before and after the opening of the doors. Measures of balance before and after the opening of the doors were derived from the recordings of the center of foot pressure (range and speed of the COP).

RESULTS

Figure 1 shows a representative COP displacement. A greater COP displacement is observed after the opening of the doors and the effect is more important along the antero-posterior axis than the medio-lateral axis.

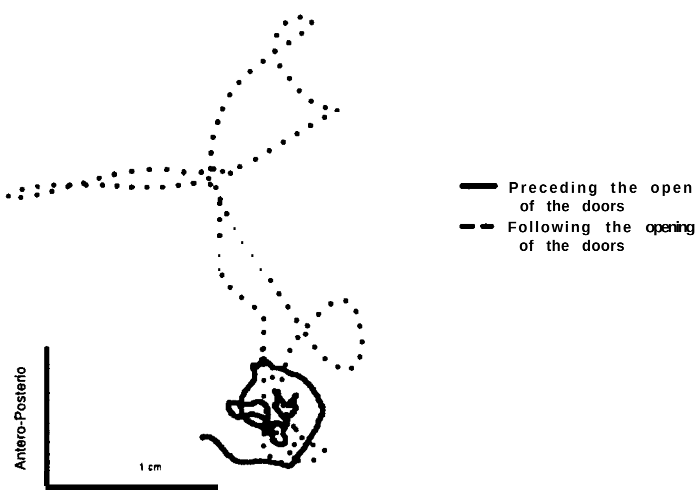

FIGURE 1:Representative displacement of the COP for elderly subject

Figure 2 shows that, on average, elderly individuals had greater range ($p<.01$) and speed ($p<.05$) after than before the opening of the doors and the magnitude was two to three times that observed for the young subjects.

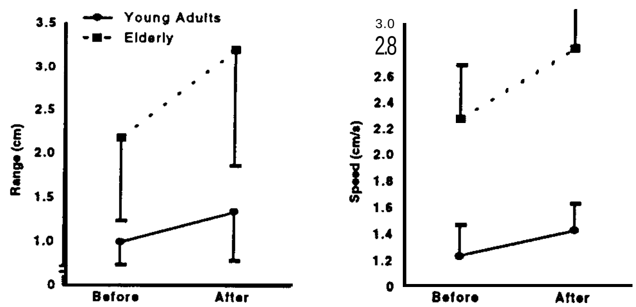

FIGURE 2:Range and speed of the center of foot pressure for young and elderly individuals for 4-s periods before and after the opening of the doors

DISCUSSION

A biomechanical inverted pendulum model was developed to determine the ankle muscular torque necessary for counteracting similar perturbation events. This general model provides a good approximation of posture control for upright standing condition.[2] The model is based on a dynamic equation of angular motion and three main torque parameters (temporal onset of the ankle muscular torque, maximum ankle muscular torque, and rate of torque development) were entered in a simulation to determine the boundary conditions of postural stability and to predict conditions leading to a fall. The simulation showed that a slower detection of a forward displacement needed a more rapid ankle muscular torque development and/or a greater ankle muscular torque production to avoid falling. Alternately, a rapid detection of a destabilizing event can allow a somewhat slower rate of torque development. Finally, small destabilizing postural event (such as those observed in the present experiment) do not require a maximal torque response to prevent a fall. Rather, the central nervous system tends to produce responses that are adapted to the context. Unfortunately, the motor responses are not deterministic but characterized by an inherent motor output variability. This variability, also labeled neural noise, increases with aging[3] and could lead to the production of an inappropriate ankle muscular torque (e.g., undershoot of the target torque necessary for preventing a fall).

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Coordination Between Posture and Pointing Movement in Predictable and Unpredictable Contexts

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INTRODUCTION

Simple and complex voluntary focal movements are performed with postural anticipatory responses.[1-2] Little is known, however, as to how these postural responses are regulated when goal-directed pointing or reaching movements are performed. Neither contextual uncertainty nor rapid visuomotor corrections have been used to assess the on-line dynamic functional link between posture and accurate focal movement. On the other hand, the control of aiming movements have been studied extensively[3] without concerns to the postural constraints. Posture and arm movement impose additional programming and control constraints on each other. In the present experiment, we investigated the dynamic interplay between posture and movement by asking subjects to perform rapid hand pointings to visual targets from a standing position. To examine the specific contribution of postural control to arm movements as well as the functional interaction between posture and movement, subjects performed hand pointings to visual targets that could change position at movement onset. The functional cost of the additional processing resulting from the preparation to this visual perturbation could inform about the degree of synergy (parallel processing) or dissociation (serial processing) of the interactive control processes of posture and movement.

METHODS

Seven volunteers aged 24 to 36 (all males and right-handed) participated on a voluntary basis. They were free from any visual and motor deficiency. In the initial position (fig. 1), the subject had his right hand in contact with the lower extremity of the sternum. The task required pointing with the right hand to one of two targets (light emitting diodes) presented in front of him along the sagittal axis. The two targets were 20 cm apart, 10 cm ahead and 10 cm beyond a distance corresponding to the length of the right arm extended in a pointing position. In the first condition (n=20), after a ready signal, one of the two targets was light up and the subject had to point as fast and as accurately as possible to the target. In the second condition (n=8O), a visual perturbation occurred randomly in 33% of the trials at hand movement onset, turning off the initial target and simultaneously lighting up the other one. In the present paper, trials with a perturbation are not reported and all movements reported are identical in terms of movement finality. More specifically, trials directed to the farthest target performed in a context with and without visual perturbation are presented. The 3D kinematics of arm and postural movements were obtained with a Selspot II system. Markers were placed on the right ankle, knee, ear lobe, index, and both hips and shoulders.

Fig. 1. Initial and terminal positions.

RESULTS

The analyses focused on the effect of contextual uncertainty on the control of posture and movement.

Visuomanual pointing analysis. Overall, less than 5% of the trials were rejected because of a lack of accuracy at target contact (radial error greater than 1.5 cm) and there was no difference across the two conditions presented in this paper. The hand movement was not affected by the contextual uncertainty. On average, the reaction time was 257 ms and the movement time was 365 ms. Similarly, the kinematics analyses of the arm movement did not reveal any effect of the contextual uncertainty (on average, peak velocity: 4.150 m/s; time to peak velocity: 167 ms; peak acceleration: 42 m/s^2; time to peak acceleration: 91 ms).

Postural analysis. The initial standing position before movement onset (as measured by the hip angle) was not affected by the uncertainty (on average, 174 deg). On the other hand, the hip angle (fig. 2) was more important and varied more across trials when there was a possibility of a visual perturbation (157 vs 164 deg; t = -3.2, p<O.O2).

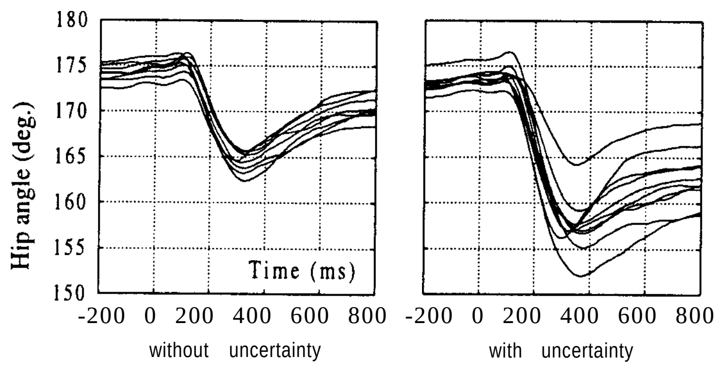

Fig. 1. Hip angle for one subject

DISCUSSION

It is important to emphasise that results presented in the present experiment are for movements without perturbation directed towards the same spatial goal. Thus, trials are differentiated only by the likeliness of a perturbation. Although the hand movements were not different, the control of posture was modified to prepare for the possibility of a visual perturbation. Increasing the trunk forward movements in the condition with uncertainty suggests the use of additional degrees of freedom at the hip joint. Hence, a less rigid on-line postural organisation was adopted when the environment was unpredictable than when it was predictable. This evokes the question of the flexibility of the feedforward processes[4] that defines postural mechanisms before a movement is performed. The increased forward movements of the trunk would indicate a predisposition of the central control processes to react to arm movement correction, suggesting that postural components of pointing movements are not simply set before arm movement but continuously updated according to the constraints imposed by the arm movement. These feedforward processes could serve also to maintain the stability of the whole body during a movement to avoid that focal task constraints exceed the reactive capability of the postural system.

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Postural Adjustments of Elbow and Shoulder Joints Associated with Fast Wrist Movements. Their Dependence on the Initial Postural Conditions

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INTRODUCTION

Anticipatory and reactional postural adjustments associated with voluntary movements have been extensively investigated in motor acts involving large perturbations of whole body balance: upper limb rising, trunk bending, elevation on tip-toe, initiation of gait...(1,2,4). These paradigms are hardly applicable to patients suffering from acute impairments of the body balance or/and who are easily tired. Thus, there is a need to define simple paradigms for accessing the coordination of posture and movement in these patients. The purpose of the present experiments was to investigate the postural adjustments of shoulder and elbow joints associated with fast wrist movements performed by healthy seated subjects.

METHODS

Six healthy subjects, aged from 23 to 54 years, participated in the experiments. The subject stood in front of a table, with his dominant arm and forearm flexed at 60 degrees (relative to full extension). The hand was supinated and strapped to a splint with the fingers fully extended. The subjects performed alternate wrist flexions and extensions, as fast as possible, in the sagittal plane. They were instructed to keep the posture of the upper limb as constant as possible without external feedback. This motor task was performed under two sets of experimental conditions: 1) with the elbow resting on the table, and 2) with no elbow support. Tangential accelerations of the hand, forearm and arm, and wrist and elbow angular displacement were recorded. Surface EMGs of Extensor carpi radialis (ECR), Flexor carpi ulnaris (FCU), Biceps brachii (BB), Triceps brachii (TB), and Deltoideus anterior (DA) were simultaneously recorded. All signals were digitized at 1 kHZ.

The onset of wrist movement was defined as the start of wrist acceleration and this time was used as “time zero” (t0) for the other measurements. The time of activation onset of each recorded muscle was defined visually for each trial. For each muscle, two measures of integrated EMG were made: 1) background EMG over 100 ms from 300 ms to 200 ms before t0, and 2) integrated EMG from the onset of the EMG burst to the peak velocity of wrist movement. The times of onset of shoulder and elbow movements were defined as the start of arm and forearm accelerations. The amplitude of the elbow and wrist rotations at the end of the wrist movements was assessed using the signals from the goniometers.

Student’s t and Mann and Whitney tests were used to investigate whether there were significant differences between the results obtained with and without elbow support. The significance level was set at $p \leq 0.05$.

RESULTS

The peak acceleration, peak velocity and peak displacement of wrist flexion and extension movements with elbow support did not differ in time or amplitude from those performed without elbow support. Similarly, there was no significant difference in the time of activation onset or in the integrated EMG values of ECR or FCU muscles for the two postural conditions.

Elbow and shoulder joints showed low-amplitude, mono- or multi-phasic oscillations, which started before t0. The maximum angle of the final elbow deviation was less than 10 degrees and that of the shoulder was too small to be measured. Interestingly, regardless of the direction of the wrist movement (flexion or extension), the initial direction of the first “postural” joints movements was either a flexion or an extension, depending on the subject. The mean values of onset time for elbow movements ranged from -17 ms to -24 ms and that of shoulder movements ranged from -11 ms to -22 ms. These onset times did not differ significantly for the two postural conditions. Nevertheless, elbow movements began later in some subjects and earlier in others, when the wrist movements were performed without elbow support.

The postural joint movements were associated with anticipatory muscle activations (mean values ranged from 1 to 41 ms before t0). Wrist flexion was associated with a chronological pattern of postural muscle activations, with muscles acivated in the order: BB, DA, TB without elbow support and DA, BB, TB with elbow support. The mean integrated EMG values for these muscular activations did not differ between the two postural conditions despite a significant increase in DA and BB iEMG for 4 subjects in the absence of elbow support. The chronology of muscle activations associated with wrist extension was TB, DA, BB for both postural conditions. DA iEMG was higher for 4 subjects in the absence of elbow support.

In the absence of elbow support The tonic initial iEMG of DA was always 8 to 10 times greater without elbow support than with elbow support, whereas BB and TB initial iEMG was significantly higher in only 3 subjects.

DISCUSSION

The experimental paradigm used in this study makes it possible to analyze posture-movement coordination and interaction without significantly disturbing the subject’s whole body balance. In contrast with Latash et al. data (3), our results show that anticipatory postural muscle activations are associated with fast wrist movements. This difference in the results obtained may be due to the less stable upper limb posture used in the present experiments. However, the anticipatory postural adjustments seem to have much less functional importance than reactional postural adjustments. We also found that, for a given postural condition, postural adjustments are associated with a stable chronological pattern of postural muscle activations and that this pattern and the degree of excitation of these muscles can be partly modified by the presence of an elbow support.

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Muscles Within Muscles: The Neuromotor Control of Intra-Muscular Segments

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INTRODUCTION

It has now been generally accepted that the CNS has the ability to independently control the intensity of motor unit activation within discrete segments (sub-volumes) of individual skeletal muscles (Paton and Brown, 1994). Furthermore, it is quite possible that during a single muscle's period of activation, these individual muscle segments may take on "prime mover", "synergist", "stabilisation" or "antagonist" functions; roles normally associated with the function of whole muscles within muscle groups. The current investigation, chose the human deltoid muscle to firstly clarify its segmental anatomy and to then determine (via EMG) what neuromotor control strategies were employed by the CNS to control these segments during the production of isometric - abduction and -adduction force impulses around the shoulder joint.

METHODS

For the anatomical analysis of the deltoid segments were individually identified and dissected away from the remaining deltoid muscle mass leaving 20 mm of the segment's muscle fiber's at both its origin and insertion. A segment was defined as a collection of muscle fiber's with a distinctly identifiable origin and insertion. To the center of each origin and insertion were then attached metal self tapping screws which enabled lines of action for each segment to be calculated from the subsequent radiographic imaging. The radiographs were taken with the cadaveric arm at 40 degrees of abduction in anterior and posterior views.

Eighteen subjects (mean age = 22.3 ± 3.26 yrs), with no history of shoulder injuries, volunteered to participate in the functional analysis of the deltoid. Miniature bipolar surface electrodes (lmm active plates; 7 mm interelectrode distance) were placed on the segments of the right deltoid in accordance to normalisation methods used in the dissection. Subjects were then seated in an experimental chair with their right arm placed in an arm cast at 40 degrees of abduction in the plane of the scapula. The subject was then instructed to match a computer simulated force-time curve that matched a 50% MVC contraction intensity with a movement time (time to peak isometric force) of 400 ms. For the analysis the temporal (EMG activation and deactivation) and intensity (%MVC and peak intensity) variables were calculated for each segment and statistically analysed ($p<0.05$) between each segment and across contraction modes.

RESULTS

The results of the anatomical analysis revealed that the deltoid muscle was composed of seven segments delineated by anatomically distinct origins and/or insertions [Figure 1]. The radiographic analysis indicated that the lines of action of segments Dl to D5 were above the joints axis of rotation, segment D6 had a line of action close to the joint's axis of rotation, while segment D7 had a line of action which suggested a shoulder adduction function.

The results of the functional analysis showed statistically significant ($p<0.05$) differences in both temporal and intensity variables for the segments of the deltoid during the two movements. Post hoc analysis suggested that segments D2 and D3 were activated significantly (p<O.O5) earlier than all other segments within the deltoid at -84.8 ± 22.5 ms and -83.2 ± 18.4 ms respectively before the rise in the force curve during abduction [Figure 2]. In contrast, during the adduction force impulses, in which only D6 and D7 were activated, segment D7 (-29.9 ± 52.7 ms) was activated significantly earlier than segment D6 (3.9 ± 52.1 ms) with the duration of activation of segment D6 (494.0 ± 72.6 ms) shown to be significantly (p<0.05) shorter than that found in segment D7 (550.0 ± 92.9 ms) [Figure 2] In generating an isometric abduction torque around the shoulder joint, all seven segments were activated between 22.8% and 33.5% of their values recorded during the MVCs In comparison only segments D6 and D7 were activated during the isometric shoulder adduction force impulses [Figure 3]

DISCUSSION

The aim of this investigation was to determine how the CNS controls individual segments of a single skeletal muscle. The results of this study clearly indicated that different segmental neuromotor control strategies were employed by the CNS dependent upon the position of the segment in relation to the axis of rotation and the direction of isometric force impulse generation (abduction or adductionj.

The results of the *abduction* force impulse indicate that the early activation (onset) and long durations for segments D2 and D3 could be considered to be the “prime mover” segments for that movement. With more oblique lines of action, later activation times and shorter periods of activity, segments D 1, D4 and D5 appear to have a more “synergist” role to help segments D2 and D3 generate, and then develop the isometric force impulse. With an activation time close to the initial rise of the force-time curve and a line of action across the axis of rotation, segment D6 appeared to play a “stabilising” role for the shoulder joint. Finally, segment D7, with the latest activation time (29ms after the initial rise of the force-time curve), lower intensity of motor unit activation (than segments D2 to D6) and a line of action below the axis of rotation, was characterised as an “antagonist” to the intended abduction motion. The results for the isometric *adduction* force impulse clearly indicated that the deltoid could be functionally differentiated since only segments D6 and D7 were activated during this isometric motion. With a line of action favourable to the direction of motion (adduction), an activation time 29.9ms prior to the initial rise of the force-time curve and a significantly (p<O.O5 j longer period of activation than segment D6, segment D7 appeared to have a role in both the generation, and subsequent development of the isometric adduction force impulse. In contrast, segment D6 with similar activation characteristics (timing and intensityj to that seen during the isometric abduction force impulse, again appeared to be playing a “stabilisation” role. In summary, the data suggests that the functional roles of skeletal muscle segments may be determined from the segments line of action and the direction of the intended motion. The possible existence of “muscles within muscles” eluded to in this study affords flexibility for the CNS to “fine tune” the activity of skeletal muscle to efficiently meet the demands of the imposed motor task.

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Closed-Loop Control of Isometric Force by Orderly Recruitment

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INTRODUCTION

Force feedback provided by Golgi tendon organs is used in the physiologic control of movement. It has been shown that this feedback mechanism operates as a simple gain of less than unity'. Muscle simulation studies suggest that this feedback design is sufficient to obtain excellent tracking if orderly recruitment is elicited electrically[2]. The purpose of this study is to explore experimentally the use of force feedback in a muscle control system incorporating orderly motor unit recruitment through high frequency stimulation. This muscle control system is intended as a sub-unit of a joint control system also involving agonist-antagonist interaction and joint position or torque feedback.

METHODS

The medial gastrocnemius of anesthetized cats were prepared by the classical isometric/sciatic preparation. A tri-polar cuff electrode was used to deliver pulses from a recruitment/firing rate stimulation system. Ramp, staircase, sinusoidal and band-limited pseudo-random signals were used to control motor unit recruitment. Force feedback was used on the recruitment stimulus at the highest gain which did not create oscillations (between 0.65- 1 .O), and compared against the open loop condition. The comparison was based on linear correlation for the linear and triangular signal, cross-correlation for sinusoidal and random signals, and by rise time and RMS error for the staircase signals. ANOVA was used for the satistical analysis of the data..

RESULTS

Figure 1 illustrates the dramatic improvements in tracking a variety of signals obtained by the addition of force feedback to the muscle control system. For linear signals, correlation coefficients increased from a mean of 0.955 to 0.993; there was a significant effet of feedback, but no effect of ramp time on correlation. In staircase inputs, mean steady state error decreased significantly from 17% to 4%, and mean settling time decreased from 763 mS to236 mS In sinusoidal, trangular and pseudo-random signals, mean maximum peak cross correlation improved from 0.842 at 124 mS to 0.975 at 23mS ($p < 0.0002$). An effect of sine wave frequency was found; increasing frequency tended to deteriorate tracking quality. Additionally, there was a significant effect of random input sequence on the peak cross correlation.

DISCUSSION AND CONCLUSIONS

A simple gain feedback improves dramatically the ability of isometric muscle to track various input signals as predicted by linear models, increasing correlation between input and output, and shortening settling and delay times. Also, the range over which the stimulation-muscle system behaves linearly can be extended to most of the active force range. However, the fact hat there are effects of random signal on the correlation parameters suggests that the system can't be though of as a simple time delay, but that muslce dynamics still play a rolew in the system's behavior. On a warning note, however, it was observed during the calibration process that if excessive feedback gains are used, instability in the form of oscillations similar to clonus can occur.

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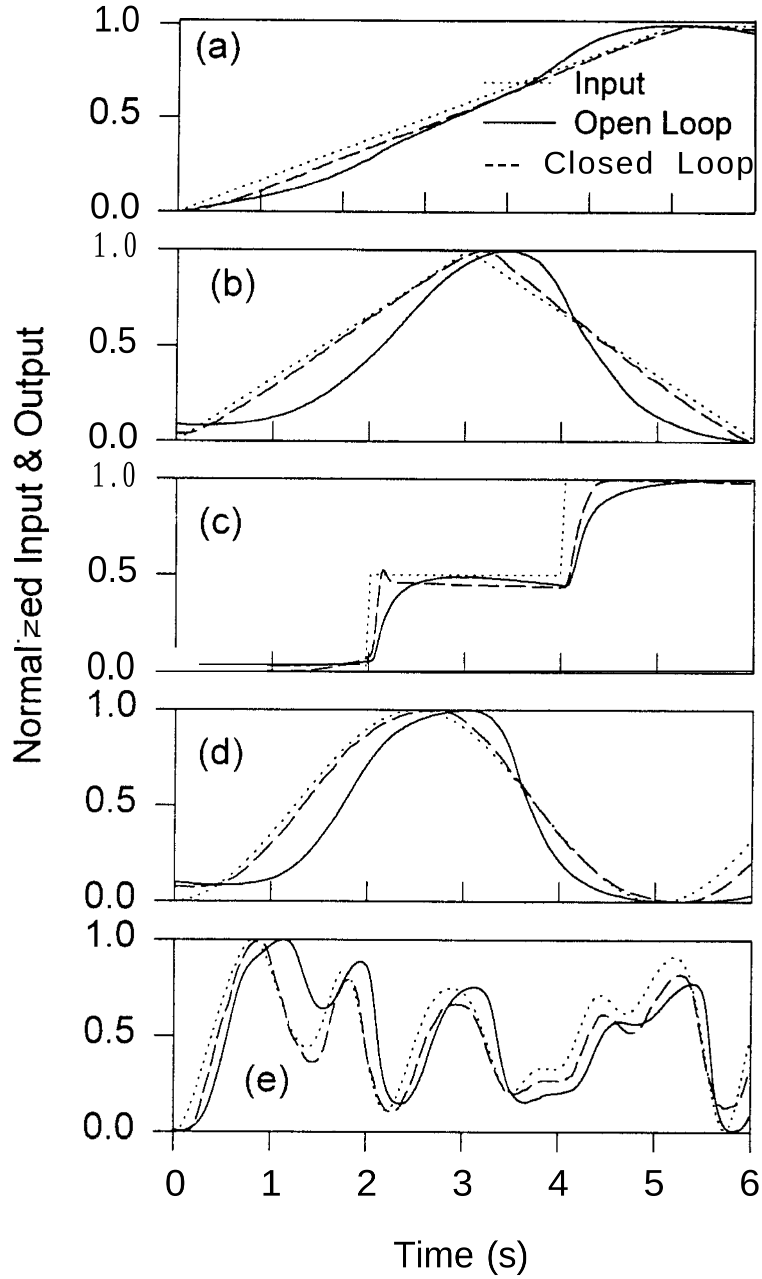
Figure 1: Sample input and output signals for a 6 second ramp (a), a 6 second triangular wave (b), a 3 step staircase (c), a 0.2 Hz sine (d) and a pseudo-random sequence (e).



The Bilateral Deficit, a Novel Approach

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INTRODUCTION

Several studies suggest that during bilateral activity of homonymous muscles maximum neural drive and maximum force output is reduced as compared to unilateral activity of the same muscle (e.g. (l-4)). It appears difficult for most subjects to maximally activate a muscle (5-7). This leads to a considerable between-trial variation in maximum contraction force (e.g.8,9) and maximum neural drive as studied by twitch interpolation (6, 7). Due to the nature of the EMG signal, variation in maximum EMG amplitude is even higher (8, 10). Howard and Enoka (11) did find a bilateral force deficit in leg extension, but no significant deficit in bilateral maximum EMG amplitudes. They attributed this apparent contradiction to the high between-trial variation of the EMG data. This argument might be extended to explain the fact that no or only a minor deficit in force or neural drive has been found in some studies (e.g. 7). In the present study we have tried to substantiate this assumption by introducing an experimental protocol which would exclude between-trial variation. In previous studies separate bilateral and unilateral contractions have been performed, thus necessitating comparison of separate trials to determine the presence or absence of a bilateral deficit. In the present study this approach was compared to using contractions starting at one side, with the Contralateral muscle contracting later, while the subject attempted to keep the first activated muscle maximally active.

METHODS

10 rjght-handed subjects (5 male, 5 female) performed 3 attempted maximum isometric unilateral finger flexions of 5 seconds with each hand separately and 3 bilateral contractions. EMG was recorded from the flexor digitorum superficialis muscles. The best trials (highest peak force) were selected for further analysis. The maximum force (F) was determined and the mean of the rectified EMG (E) during the period at which the force was above 90% of maximum was calculated to obtain an estimate of the peak neural drive. The bilateral deficits for force and EMG were determined as: $([F \text{ or } E]_{unilateral} - [F \text{ or } E]_{bilateral}) / [F \text{ or } E]_{unilateral} * 100$. This protocol will be referred to as protocol 1. In protocol 2, 5 subjects performed 3 exertions starting with the right hand and after 1 s following with the left hand. In the other 5 subjects the order was reversed. The force (F) and rectified EMG (E) were averaged over 0.5 s preceding the onset of force in the second hand and over 0.5 s after force onset. The bilateral deficit was defined as: $([F \text{ or } E]_{before} - [F \text{ or } E]_{after}) / [F \text{ or } E]_{before} * 100$.

RESULTS

A significant bilateral deficit of both force and EMG was found in all conditions (table 1). No significant differences between subjects starting with the right and those starting with the left hand in protocol 2 were found. Therefore, the data for this condition have been pooled. The bilateral deficit was found to be much more consistent when using protocol 2, as is evidenced by the much lower p-values at an equal number of observations (n=lO).

Table 1. The bilateral deficit (%) for left and right finger flexors in force and EMG as determined from the two experimenta1 protocols.

	Force			EMG		
	protocol 1		protocol 2	protocol 1		protocol 2
	left	right	pooled	left	right	pooled
mean	18.9	18.3	26.9	18.8	22.1	25.4
SD	16.4	11.9	10.7	22.6	21.2	11.5
p	< 0.005	< 0.005	< 0.0005	< 0.025	< 0.01	< 0.0005

DISCUSSION

The bilateral deficit in separate contractions is comparable to, though somewhat higher than, results obtained for finger flexion by Ohtsuki (3). The range obtained in the study by Ohtsuki (-25 to + 44%) underlines the wide scatter in the data obtained with this protocol. Also in the present data set for each of the deficits calculated in protocol 1 negative values were found for one or more subjects. In a previous experiment the bilateral deficit in finger flexion force and EMG did not reach signficance due to this scatter (unpublished data) and also Seki and Ohtsuki did not find a significant deficit in hand grip force (12). In contrast, the findings presented here obtained during the asynchronous contractions in protocol 2 appear to be much more consistent. In this protocol no negative deficits were found.

An alternative explanation for the deficit in protocol 2 might be a decline in force occurring over time. This is, however, contradicted by a much higher rate of force decline in protocol 2 as compared to protocol 1, and by the similiraty in the magnitude of the deficit in both protocols.

This finding suggests that the controversy in the literature on the bilateral deficit may be caused by the high variance in activation realized during maximal attempts. However, in view of the diverging results on the bilateral deficit presented in the literature so far, an attempt at replication of this finding seems necessary. An alternative explanations for the controversy was given by Herbert and Gandevia (7) who suggest that many positive findings on the bilateral deficit were found in larger muscle groups, where stabilization of the body during exertion of a bilateral force may be problematic. For this reason the present study focused on a relatively small muscle group. Another factor contributing to the disparate results in the literature is the fact that after specific training in bilateral activities the bilateral deficit disappears (11, 13).

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Effects of Aging on Synchronization of Motor Unit Firings

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INTRODUCTION

The present study was motivated by our observations in a larger study to investigate the effects of aging on motor unit (MU) control. In addition to providing support for the observations of depressed firing rates in elderly MUs, this study revealed two significant alterations to the relationship between the firing properties of MUs concurrently active in a contraction [1]: 1) The phenomenon of onion skin observed in the young, whereby MUs recruited earlier during increasing force have a higher average firing rate than MUs recruited later, was not observed in the elderly in whom the firing rate plots of motor units displayed "crossover" and; 2) the highly correlated common fluctuations in the firing rates of concurrently active MUs in the young, which have been termed "common drive" and interpreted as the unison control of MUs of in a given pool, are not observed as frequently in the elderly. The purpose of the present study was to investigate the effects of aging on the synchronization of MUs, a property reflecting more of the connectivity between MUs as compared to the common drive reflecting the control strategies. It was hypothesized that the neuronal remodeling that takes place in aging would result in changes in the synchronization among MU firings. In addition to providing information regarding the age-related changes on synchronization, it was hoped that this study would offer insight into the relationship between common drive and synchronization.

METHODS

Data were collected from 10 subjects aged 20-37 years (30.2 ± 5.66 yr., mean ± SD), classified as "young", and 10 subjects aged 65-88 years (76.9 ± 6.56 yr.), classified as "elderly". The elderly subjects were screened for neuromuscular disorders by a practicing physiatrist. Intramuscular electromyographic (EMG) and isometric force data were obtained from the first dorsal interosseous muscle of the dominant hand. The hand was placed in a restraining device to secure thumb and index finger at a 90 degree angle while a high stiffness force transducer positioned against the proximal interphalangeal joint measured index finger abduction force. A highly selective needle electrode was used to record three differential channels of EMG signals from the belly of the muscle. The signals were resolved into the individual MU firing trains using the Precision Decomposition technique [2]. The task presented to the subject involved tracing of a trapezoidal trajectory on the computer screen by abducting the index finger while visual feedback on the force was provided. The plateau of the trapezoidal trajectory was scaled to the subject's maximal voluntary contraction (MVC) force and the ramp parts of the trajectory had a slope of 10% MVC per second. Myoelectric data were acquired while the subject attempted the trajectories at plateau levels of 20% and 50% MVC lasting 60 and 20 seconds respectively.

A previously introduced technique for assessing synchronization [3] was modified by employing a strict criterion for accepting a peak in the cross-interval histogram as synchronized events. Two parameters are used to characterize synchronization: MU Synch, the percentage of all possible MU pairs recorded in a single contraction that exhibit synchronization, and Synch Index (SI), a measure of the degree of synchronization. The SI is computed as the percentage of reference MU firings synchronized with the alternate unit above chance [3].

RESULTS

A total of 675 MU pairs from 64 individual trials were tested for short-term (+/- 6ms latency) synchronization. Only MUs with more than 300 firings were analyzed. No significant differences (unpaired t-test, $p < 0.05$) were found for mean values of MU Synch or SI between young and elderly but the variability was higher in the older subjects at both contraction levels (Table 1).

Table 1 Percentage of concurrently firing motor units that show short-term synchronized events (Mu Synch) and level of synchronization in pairs of motor units (Synch Index) in young and aged subjects

	MU Synch (%)			Synch Index			
	Mean	*SD*	*c v*	*Mean*	*SD*	*c v*	*N*
Young 20% MVC	**70.17**	**31.96**	**45.55**	**7.28**	**3.03**	**41.61**	**19**
Aged 20% MVC	**72.50**	**43.97**	**60.65**	**8.89**	**5.35**	**60.13**	**14**
Young 50% MVC	**86.11**	**18.08**	**21.00**	**11.18**	**3.26**	**29.20**	**12**
Aged 50% MVC	**84.94**	**19.76**	**23.26**	**9.13**	**3.64**	**39.85**	**19**

Abbreviations: SD, standard deviation; CV, coefficient of variation; N, number of contractions

DISCUSSION

While further analyses are needed, specifically of cases with especially high and low synchronization and common drive, the observed patterns in synchronization in the aged can be a result of the remodeling of the MU pool. The decrease in common drive combined with the lack of age-related changes in SI and MU Synch at either force level as well as the higher variability of both parameters observed in the elderly, may suggest an age-induced alteration in the relative weights of common versus uncommon inputs. Such an alteration would result in decreased common drive. However, synchronization, being a shorter-term, smaller-scale phenomenon, may be dependent on a smaller set of common inputs. The loss of a few neurons may have a more drastic effect in this set, increasing or decreasing the weight of common inputs depending on whether the lost neurons were mostly common to the pair of MUs in question or not. Hence in some cases synchronization may increase slightly while in others it may decrease. This would cause the average values observed to remain unchanged while bringing about an increase in variability.

Our results also indicate that synchronization is not invoked as a compensatory mechanism to counter any diminutive effects of aging on muscle performance, which is in agreement with our previous conclusion that synchronization does not serve a physiological purpose and is simply a by-product of other mechanisms.

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Interpreting the Amplitude of Surface EMG Data

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INTRODUCTION

Under what circumstances may the absolute amplitude of surface EMG be useful? The well-known ill-posed nature of the inverse problem of electromyography led most investigators to discount any attempts to employ absolute measures in studies involving such recordings. Instead, it is customary to normalize *the data in* some way so as to retain useful patterns of activity yet ignore the actual voltage values recorded[6]. However, it may be difficult to determine an appropriate normalization standard in some circumstances.

Examination of records produced during studies of EMG in a defined protocol to document patterns of altered motor control in subjects with severe spinal cord injury (SCI) [4] led us to conclude that the recorded activity levels might provide useful insights into the exact status of the altered motor control, and that quantification of such levels could provide valuable quantitative data about the status of the subject's altered motor control.

We subsequently developed a recording protocol to allow digitization of this study in its entirety. We have shown that such recordings, when made in stable SCI subjects, are highly reproducible with test-retest correlations as high as 0.98 with sufficient smoothing of the data sets [S]. Thus, for sequential studies in the same subjects, it is reasonable to use an initial study as the baseline for "normalization", and to report results in terms of changes from that baseline study. as it is reasonable to assume that the biophysics of the recording do not change between studies conducted over relatively short intervals.

SUBJECTS

98 (2 female and 96 male) spinal cord injured subjects with lesions ranging from C2 to T12 (61 cervical and 37 thoracic lesions), sustained one month to 38 years prior to the examination (7.9 ± 7.7 years) were recruited from the Spinal Cord Injury Service of a Veteran's Affairs Medical Center (VAMC). Subjects' Ashworth scores **[l]** ranged from 0 (flaccid) to 3 (marked resistance to passive movement), with two exceptions having a score of 4.

METHODS

BMCA *Protocol* The overall purpose of the study was to assess the utility of the surface EMG data in studying altered motor control by comparing clinical and electrophysiological measures of spasticity or altered motor control after spinal cord injury. For this paper, only passive maneuvers are reported.

Subjects were placed in a comfortable, supine position and pairs of surface electrodes were placed centered over the long axis of the muscle bellies, 3 cm apart after preparing the skin to reduce electrode impedance below 5 kΩ. Recordings were taken from 5 major muscle groups of each lower limb (total 10 channels). SEMG data were amplified 5000 times with a (-3 db) bandwidth of 40-600 Hz, using Grass 12A6 amplifiers. An event mark to denote timing of protocol commands was recorded along with the SEMG data

After electrode placement, a physician carried out the clinical examination [3] using the Ashworth Scale **[l]** to assess muscle tone in the lower limb while moving the hip and knee together in a single maneuver, giving a single *score to* the combined maneuver *(flexion* and extension) for each limb.

For the recording, maneuvers were repeated three times for each side for hip and knee flexion together (first phase) then extension (second phase). Each phase of each maneuver was maintained for a minimum of 5 seconds.

Data Reduction Clinical data were scored according to published clinical scales [3]. The SEMG data were analyzed from the EMG envelope calculated post-hoc using a root mean square (RMS) algorithm [2]. Data were compressed in this way to an effective 20 samples/s. We used the average activity over a 5 second window [5] . The SEMG data for each maneuver were thereby reduced to a set of numbers, each of which represented the response of each muscle to each phase of each maneuver. This set of numbers was combined in various ways for the s subsequent data analysis. In order to maintain a consistent scaling throughout, data were summarized across muscles and across maneuvers as average activity in μV_{RMS}.

The data were divided into two classes dividing the group into slight or no increasedtone, and marked increase in tone (Ashworth score less than 2 vs. greater than or equal to 2). To compare clinical and electrophysiological data, three different methods were evaluated, top-down induction of decision trees, nearest neighbor and linear discriminant analysis. The latter (linear discriminant analysis) did not perform well, and hence was not considered in determining the best set of measures [7]. The former two methods gave nearly equivalent results, and were used in the following summary.

RESULTS

The comparison of the two data sets showed a satisfactory relationship overall. As would be expected, EMG activity recorded during passive movement of the ankle showed little relation to the Ashwortb score give by the clinician to hip and knee combined movement, regardless of whether these data were incorporated into an overall average score or analyzed independently.

When EMG activity in response to passive hip and knee movements was compared with the Ashworth classification however, accuracies approaching 75% were obtained with both methods.

Not all subjects were in good agreement between the two methods; therefore outliers for the passive hip and knee movement were considered separately. For subjects who had low average EMG activity to the maneuver but were given a large (>2) Ashworth score, the probable interpretation is that the subjects were exhibiting a mechanically, rather than neurally mediated increased resistance to passive stretch. Other subjects exhibited the opposite situation, low Ashworth score and *high* average EMG activity.

DISCUSSION

The analysis done to date indicates that the surface EMG data may be used to predict the clinical findings [7]. The results show that the expected thigh muscle activity was most informative in predicting the Ashworth scores. While it might be thought that subjects with increased tone in one joint might exhibitthe same phenomenon in other joints, ankle and hip/knee joints exhibit many differences in innervation and condition; thus the lack of relationship is not surprising. The observed outliers in the data set suggest that, for managing the neuromuscular problems of SCI subjects, surface EMG data is more useful than is the perceived net torque as expressed in the Ashworth score. In addition, the EMG data are more consistent than the Ashworth data, and are objective data.

The relationship of these two data sets suggests that the interpretation of amplitude of the surface EMG may have been unduly pessimistic in the past. While it is certainly true that the biophysics of the recording situation can greatly affect the amplitude of the recorded signal, nevertheless when considering real problems such as altered motor control or spasticity after spinal cord injury, the observed differencesbetween subjects contain substantial information related to the underlying pathophysiology when appropriately analyzed.

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Efficient Algorithm for Fatigue Monitoring from Dynamic Surface Myoelectric Signal Using a Complex Covariance Approach

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Introduction. Several studies in the literature show how the surface myoelectric signal can be used to detect the muscular fatigue revealed by a compression of the signal spectrum toward the low frequency band. Then, the median and the mean frequency of the spectrum have been extensively used as indexes to monitor the fatigue effect. Anyhow, the electrical effect due to the fatigue, even if well known in static conditions, is not yet enough clear in dynamic recordings. The reason lies in the difficulty of analyzing strongly non-stationary time series like the myoelectric signal recorded during dynamic protocols. Furthermore, another limitation on the monitoring of the dynamic fatigue is due to the unavailability of indexes which could be updated in real-time. This becomes a compulsory feature when the goal consists of demonstrating that the electrical effect of the fatigue anticipates the mechanic muscular deficit. In this work a method based on the estimate of the complex covariance has been optimized and tested on both synthetic and experimental dynamic myoelectric signals. The method allows robust estimate and seems to perform well even with high-variability signals.

Methods. The mean frequency of a spectrum can be expressed as the derivative (approximated by a finite difference) of the complex covariance function evaluated in the origin. The complex covariance function is $C_v(n) = \frac{1}{2N+1}\sum_{k=-N}^{N} x^+(k)x^{+*}(n+k)$, where $x^+(n) = x(n) + j\hat{x}(n)$ is the analytical signal of $x(n)$, obtained by using a Hilbert filter. The correct choice of the Hilbert filter length, affecting the computational cost of the estimator, the delay and the robustness of the estimates, represents a tradeoff to consider in the implementation of the method. An estimate of the complex covariance at the time point n can be obtained by $C_V(n) = w \cdot C_V(n-1) + R(n)$, where $R(n) = x^+(n) + x^{+*}(n-1)$ and w is a weight factor representing the number of samples N contributing to the covariance estimate (N = 1/ (1-w)). The mean frequency of the signal spectrum can be evaluated as $f_{mean} = \frac{1}{2\pi j}\frac{\dot{C}_v(0)}{C_v(0)} = \dot{\varphi}(0)$ ($C_v(n) = A(n)e^{j2\pi\varphi(n)}$ is the polar representation of the covariance function). The original method presents two main drawbacks: 1) high variability of the results depending on the Hilbert filter length, and 2) bad performance when applied on signal characterized by fast frequency variations. The first point has been resolved by windowing the Hilbert filter using a Harming window: this choice guarantees a higher stability of the filter almost independently of its length. In order to get over the second drawback, the algorithm has been modified by estimating the *R(n)* values using a weighting window around the point n. The n-th covariance sample is $C_v(n) = P_v(n) + F_v(n) + R(n)$, where $P_v(n) = \sum_{i=1}^{N/2} w^i R(n-i) = wP_v(n-1) - w^{N+1}R(n-N-1) + wR(n-1)$ and $F_v(n) = \sum_{i=1}^{N/2} w^i R(n+i) = \frac{F_v(n-1)}{w} - R(n) + w^N R(n+N)$ represent the contributions due to the samples contained in the window. Then the covariance and the mean frequency values can be recursively evaluated after a delay of N/2 samples due to the half window acquisition time.

Results. Experimental proofs testing the algorithm, have been based on the following signals: 1) chirp signal, with frequency varying in 1 second from 100 Hz to 140 Hz, to test the method during fast frequency variations (Hilbert filter length is 109 samples, and window length is 65 samples); 2) artificial myoelectric signal, 500ms long, simulating a muscular fatiguing exercise starting at 200ms and ending at 400ms, aiming at evaluating the algorithm applicability in real fatiguing conditions; 3) experimental dynamic myoelectric signals recorded during gait. The mean frequency of the chirp signal (test #l) has been estimated using both versions of the algorithm (the original one and the new proposal). The plots drawn in Fig. 1 underline how the proposed modifications are necessary to correctly follow fast frequency variations (the starting delay is due both to the Hilbert filter acquisition time and to the half window shift time). The decrease of the performance in the last samples of the mean frequency is due to border effects. The performance of the algorithm dealing with the signal simulating muscular fatigue (test #2) are shown in Fig. 2. It is worthwhile to notice the trend of the mean frequency which decreases from 200 ms to 400 ms revealing fatiguing effects. In the experimental protocol (test #3) the mean frequency has been evaluated for ten subjects using the signal recorded from four muscles: right rectus femoralis, right and left biceps femoralis and right tibialis. The results obtained during gait on a walking treadmill can be summarized in a histogram (Fig. 3) where averaged values of the mean frequency calculated for all subjects over the first and last recording of the exercise, are shown: the mean frequency decreases in the last acquisition (10÷15% less than in the first acquisition) revealing a fatiguing effect,

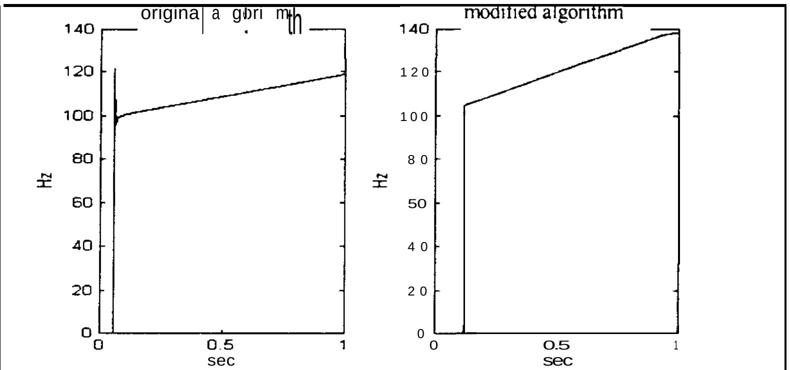

Fig.1. Mean frequency estimate of a chirp signal (test #l) using the original method (left) and the new proposal (right)

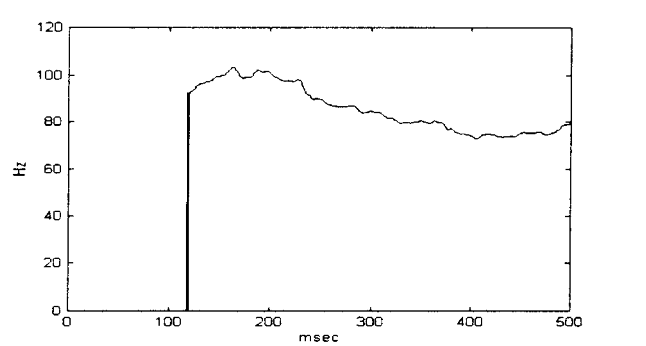

Fig. 2 Mean trequency estimate in an artifcial signal simulating dynamic exercise in fatiguing conditions.

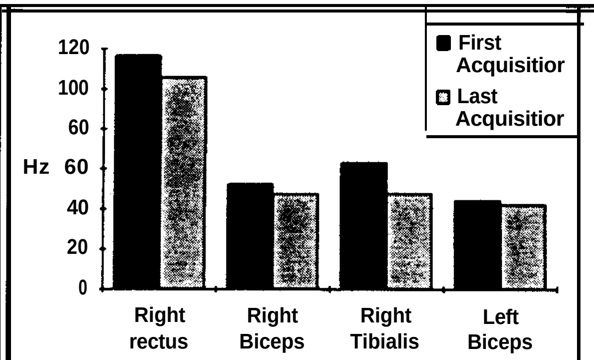

Fig.3. Mcan frequency evaluated for tour leg muscles at the begin and at the end of a walking exercise.

Moreover, the proposed method is more efficient than the Short Time Fourier Transform from the point of view of the computational complexity. The complexity for the proposed method is due to the number of multiplications for the analytical signal calculation and it is completely independent of the analysis window length. Otherwise, the method based on the Short Time Fourier Transform needs NlogN multiplications for the FFT, N multiplications for the power spectrum and N/2 multiplications to estimate the mean frequency. It appears strongly dependent of the analysis window length.

Discussion. The proposed method shows good performance in the monitoring of muscular fatigue phenomenon, qualitatively comparable to those offered by the Short Time Fourier Transform method. Otherwise the manageable computational complexity and the sequential implementation propose the complex covariance method as the best algorithm for the real time applicability.

Mean Frequency Cycling During Sustained Postural Contractions of the Cervical and Lumbar Extensor Muscles

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INTRODUCTION

The incidence of reported work-related cumulative trauma disorders (CTD's) is increasing. Occupations involving data entry, computer programming, and customer call centres, where employment is on the rise, place workers at risk of developing CTDs due to the maintenance of a static seated posture for several hours at a time. Static holding may cause chronic muscle strain, tendonitis, joint inflammation, or any combination of the above, especially in terms of the vertebral column. In order to prevent these injuries, it is important to understand their precipitating factors. One potential factor is muscle fatigue.

In moderate contractions (20-60% maximum voluntary contraction (MVC)), an increase in the myoelectric signal (MES) amplitude, and a shift to lower frequencies of the MES spectrum are thought to reflect muscle fatigue. Commonly used MES spectral parameters which reflect the muscle fatigue process are a decline in the mean frequency (MNF) and the median frequency (MDF). Time domain parameters are also used, such as an increase in the signal amplitude, or a decrease in the number of zero crossings of the signal, and a decrease in the number of EMG gaps in a given period of time.

Early researchers considered contractions below 1520% MVC to be non-fatiguing. Recent research, however, indicates that this is not the case. The endurance times for sustained isometric contractions of the elbow flexors and the knee extensors at 10% MVC is approximately one hour. Even at five and seven percent MVC, although the contractions could be maintained for one hour, there were both subjective and objective signs of fatigue'. Sjogaard et al. found that after one hour the MVC itself had fallen by ten to twelve percent of its original value'. Jorgensen et al. demonstrated that at seven percent MVC, the recorded MES from the triceps showed a marked decrease in the MNF (96.1 Hz to 69.9 Hz), but at the same effort level, the MNF of the biceps was virtually unchanged (77.3 to 75.4)[3]. Despite these findings, it is generally believed that in contractions less than 20% MVC, there is actually an increase in the MNF over time, and this has been attributed to motor unit recruitment. The work described in this paper may help to explain the discrepancies among research groups regarding the mean frequency behaviour of the MES during a sustained contraction below 15% MVC.

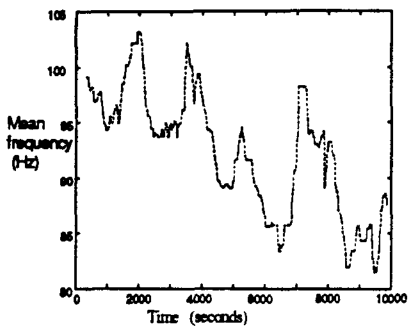
METHODS

Eighteen volunteers were recruited - three male and fifteen female participants with ages ranging form 28 to 60 years of age (median age 38). Volunteers were asked to remain seated at their workstation all morning, except for during their usual coffee break. Surface ECG electrodes (Red Dot ™ Model 2259 recessed Ag-AgCl electrodes by 3M Canada Ltd.) were applied along each subject's right lumbar erector Spinae (at L2-L5) and right cervical extensors (at C3-C6) for the bipolar measurement of MES. Estimates of the MNF of MES were recorded at two second intervals using portable data loggers developed at the University of New Brunswick Institute of Biomedical Engineering (described elsewhere)*. The data were smoothed using two sequential non-linear filters.

RESULTS

The purpose of this work was to determine whether there was an increase or a decrease in the mean frequency of the myoelectric signals recorded from the postural muscles of the neck and low back during prolonged computer terminal operation. Subjects demonstrated both increases and decreases in the slope of the MNF, however, most subjects consistently demonstrated a cycling between higher and lower mean frequencies, which occurred at a frequency of 1.5 to 4 per hour. This cycling was evident in at least one of two records at the cervical extensors in 17 of 18 subjects, and at the lumbar erector Spinae in 14 of 16 subjects (Two data sets were incomplete and removed from the analysis). Typical results are displayed in Figure 1.

FIGURE 1: a) neck b)back



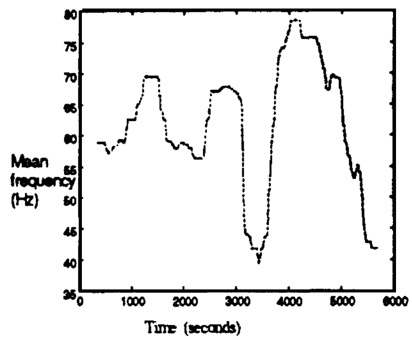


Figure 1: Typical results are shown for a) the cervical extensors and b) the lumbar erector Spinae during sustained postural contractions held while subjects performed computer terminal work for an extended period of time (in (a) 3 hours, and in (b) 2 hours).

DISCUSSION

Using surface recorded MES, it is difficult to determine what is happening at a single fiber or motor unit level. It is possible, however, to determine whether or not the shifts in MNF are accompanied by postural shifts. This has been examined using the biceps muscle, and it has been found that cycling of MNF can be induced by brief complete relaxation of the muscle, or by a temporary increase in muscle activation (a maximal voluntary contraction lasting 15 seconds or greater). This finding is consistent with recent work by DeLuca[5] who has found an increase in recruitment thresholds of single motor units performing low level activity once the activity has stopped for a short period, despite the reproduction of the exact position and contraction level Further work is required to determine the exact mechanism behind this cycling.

The authors are working toward a method for quantifying the cycling behaviour of the MNF. The most promising technique thus far is to count crossings of a least squares regression line of the overall MNF data, or similarly to count large inflections in the data.

ACKNOWLEDGMENTS

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Neuromuscular Activation and RPE During Near Maximal Steady State Contractions to Failure in the Quadriceps

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INTRODUCTION: The maintenance of a near-maximal steady state muscle contraction depends on an outflow of impulses from higher brain centers. Such an action may have a differential effect on neuromuscular activation and ratings of perceived exertion (RPE). Although an increase in RPE is often assumed to correspond with an increase in neuromuscular activation, conclusive evidence has yet to establish this notion. The purpose of this study was to examine changes in neuromuscular activation and RPE at a near maximal steady state contraction of the quadriceps.

METHODS: 17 healthy college aged male volunteers (age = 22.6 years, height = 178.9 cms, weight = 78.8 kg) were assessed for neuromuscular activation and RPE of the quadriceps. Isometric torque was measured on the Biodex System II Isokinetic Dynamometer with the each subject placed in an upright, seated position with the knee flexed 60 deg. Subjects performed isometric voluntary contractions of the quadriceps at lo%, 80%, and 100% MVC. Perceived exertion was measured with a modified category-ratio scale (CR-l 0) as developed by Borg'. In order to provide the subjects with a context through which sensations could be evaluated, 1 high and 1 low anchor were applied at 100% and 10% MVC, respectively. Subjects were asked to perform a 100% MVC (5 sec) of their quadriceps and were asked to "think about the feelings in their quadriceps at the end of the contraction and to assign a rating of maximal to those feelings". Following a brief period of volitional recovery (1-2 min) subjects were asked to contract at 10% MVC for5 sec and to assign a rating of 1 to those feelings. Following another brief period of volitional recovery, subjects were asked to contract at 80% MVC and to sustain this contraction for as long as they could. Subjects were asked to rate the feelings in their quadriceps every 5 sec during the 80% MVC by visually observing the CR-10 scale. In order to equate the different number of RPE values obtained for each subject, these measures were interpolated every $1/6^{th}$ during the duration of the 80% MVC. Neuromuscular activation was measured through the use of surface electromyography (EMG) for the vastus medialis (VM) and vastus lateralis (VL) muscles. Bi-polar circular surface electrodes (Ag/AgCl) were placed over the vastus medialis (20% of the distance from the medial joint line of the knee to the anterior superior iliac spine) and vastus lateralis (mid-way between the head of the greater trochanter and the lateral femoral epicondyle) muscles. The reference electrode was placed over the medial shaft of the tibia approximately 6-8 cms below the inferior pole of the patella EMG activity was collected and recorded via telemetry by an 8 channel FM transmitter (Noraxon Telemyo) worn by the subject in a beltpack. EMG signals were sampled at a rate of 1000 Hertz and broadcast to a FM receiver where they were bandpass filtered (16-500 Hertz) and underwent analog to digital conversion by a 16-bit A-D board interfaced to a Pentium microprocessor. Following full-wave rectification of the raw signals, EMG activity was integrated (IEMG) every $1/10^{th}$ ($\mu V \cdot sec$) of the 80% MVC. Each IEMG value was then expressed as a 1 sec average. A single factor ANOVA with repeated measures was performed on RPE. A 2 factor ANOVA (time by muscle) with repeated measures was performed on the IEMG values. All significance testing was conducted at an alpha of $p<0.05$.

RESULTS: The mean duration ($\pm$ SD) in which subjects could sustain the 80% MVC was 42.94 $\pm$ 10.23 sec. The results demonstrated a statistically significant increase in RPE during the duration of the 80% MVC ($F_{5,80} = 51.44$, $p<0.05$ – Figure 1) from 5.0 to 9.68 (mean increase = 48%). The results also demonstrated a significant increase in IEMG of both muscles ($F_{9,144} = 3.84$, $p<0.05$ -

Figure 2). The mean increase in IEMG during the 80% MVC for the VM was 18.6% and 16.3% for the VL.

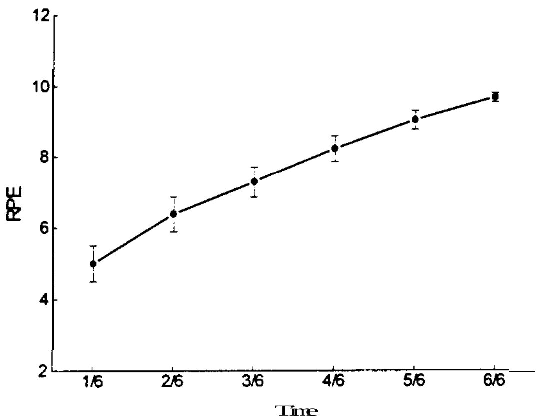

Figure 1: Changes in RPE during 80% MVC

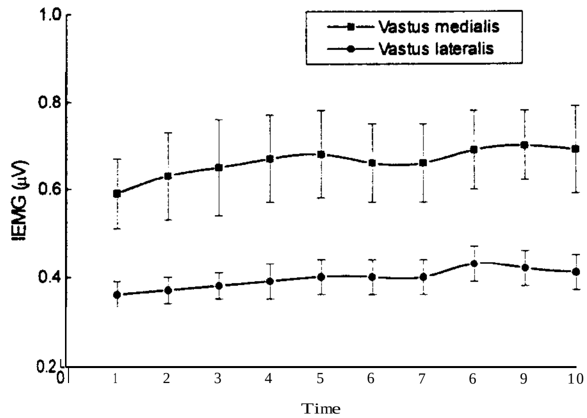

Figure 2: Changes in IEMG during 80% MVC of the VM and VL muscles

The results also revealed a significant difference in IEMG activity between the 2 muscles as the VM demonstrated significantly greater values than the VL ($F_{1,16} = 10.34$, $p<0.05$). The results did not demonstrate any significant group by time interactions ($F_{9,144} = 0.62$, p=O.78).

DISCUSSION: The major findings of this investigation demonstrate an apparent "uncoupling" of perceived exertion and neuromuscular activation during a high intensity sustained isometric contraction. Although a significant increase in RPE was observed during the 80% MVC, the increase in IEMG appears smaller in comparison. It may be thought that during a steady state isometric contraction, a continual and relatively constant flow of neural impulses to the working muscle is necessary. The degree of this outflow at high level contractions may, in fact, closely parallel force output. The feedforward hypothesis of Cafarelli (1982) suggests a corollary discharge of impulses from the motor cortex to the somatosensory cortex. In this manner, increases in neuromuscular activation, as evidenced in the EMG, would coincide with concomitant increases in perceived exertion. This notion, however, does not appear to be completely supported in this investigation. Alternatively, a feedback mechanism, stimulated by sensitization of group III and IV afferents by metabolite accumulation, may contribute significantly to RPE. It also appears evident that the relative contribution of the uni-articular VM and VL muscles to the 80% MVC is equivalent over time. However, the absolute activation of the VM is higher than the VL, suggesting the importance of the VM to quadriceps function. It is anticipated that future investigations may focus on the relative contribution of these muscles during fatiguing tasks of a dynamic nature.

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Biomechanical and Electromyographic Evaluation of Spinal Muscle Functions for Normal Subjects and the Scoliosis Patients

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INTODUCTION: Idiopathic scoliosis (IS) is a common spinal deformity and surgical correction followed by spinal fusion is the basic method of treatment. Biomechanically, the spine itself can not bear large loads, and has to be stabilized by ligaments and muscles. The question is after spine was fused, what is the function of the back muscle, are they function as usual? Or whether their functions are changed. Further more, are there spinal muscular atrophy due to the dysfunction of the muscle? In the literature, few studies looked at how spinal muscles maintain the functional interdependence after long-term spinal fusion. The purpose of this study is to examine the contributions and control strategies of the muscles in the thoracic and lumbar spine in normal subjects and scoliosis patients with spinal fusion using Electromyography (EMG).

METHODS: 15 normal subjects (mean age 21) and 15 patients (13 females, 2 male, mean age 18.5 years) with idiopathic scoliosis undergoing surgical correction at the Duchess of Kent Children's Hospital were evaluated. The IS subjects have a minimum of 2 years follow-up (mean 2.5 years). Eight pairs of surface EMG electrodes were placed on the muscles of latissumus dorsi, erector spinae, and superior traezius, over the thoracic and lumbar regions. Three tasks, which include forward, right, and left lateral bending, were tested. 5-Kg weights were held by the subjects during the tests. Three trials for each task were tested for each subject. To minimize fatigue, brief rests (2-3 min.) were given between trials. EMG and kinematics data were recorded during the tests. The EMG signals were full-wave rectified and low pass filtered at 4 Hz to produce a linear envelope (LE) EMG. In order to identify the relationship or interdependence of the EMG profiles between or within the subjects, the cross-correlation coefficients for LE EMG were calculated.

RESULTS: During the forward bending, the scoliotics shown higher EMG signals than the control group. The EMG displayed asymmetric profiles from the scoliotics The average cross-correlation coefficients between right and left side EMG was 0.83 for normal and 0.65 for scoliotics, the difference was significant ($P<0.05$). In comparison to normal subjects, almost all the muscles recorded during the testing were activated with longer duration, suggesting that there were no clear muscle synergies during forward bending. In addition, in lateral bending, more muscles were activated with higher amplitude within the scoliotics. There was no significant difference of the average cross-correlation coefficients between the normals and the scoliotics in the lateral bending, but the EMG activity patterns varied.

CONCLUSIONS: EMG profiles and the cross-correlation coefficients for back muscles have shown different muscle contraction patterns between the normal subjects and the scoliotics with spinal fusion.

Development of Gait of Individuals with Down Syndrome

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INTRODUCTION: The physiological development of mature gait is completed by the age of 7 years. During this period, children show a progressive passage from the digitigrade to the plantigrade walking (within 18 months of age) and a progressive maturation of muscle activations and movement patterns (Sutherland et al., 1980; Malouin, 1986). In spite of the fact that most of children with DS have gait problems, little is known about the development of gait in children with DS. Only two papers (Parker and Bronx, 1980; Parker et al., 1986) have described the kinesiological patterns of gait of children aged from 5.0 to 7.3 years. The aim of this study was, thus, to investigate the other components of gait such as the muscle activation patterns and the ground reaction forces (GRF) during the gait of children and adolescents with DS. METHODS: Individuals: thirty five children and adolescents with DS home reared (range of age 1.5 to 17.7 years) have been selected for their collaboration from a larger group of individuals. They did not show any radiological signs of atlanto-occipital dislocation or heart disease. According to the chronological age, they were divided in three groups: ten children aged from 1.5 to 4.8 years (group A), ten children of age ranging from 5.1 to 10.4 years (group B) and fifteen adolescents of age ranging from 11 .O and 17.7 years (group C). Kinematics of limb movements were monitored by an automatic TV image processor (ELITE, BTS, Milano) (50 Hz sampling frequency) which detected the instantaneous positions of passive retroflective markers applied onto skin landmarks of each children/adolescent (anterior-superior iliac spine, greater trochanter, knee, ankle and head of fifth metatarsus). In the younger children it was possible to apply the markers only in the knee, ankle and the head of fifth metatarsus. GRF were measured by means of a force platform (Kistler, sampling frequency: 100 Hz). EMGs were recorded from distal (tibialis anterior and gastrocnemius lateralis) and proximal (vastus lateralis and biceps femoralis) muscles of one limb by surface preamplified electrodes (sampling frequency: 500 Hz). After the setting of markers and electrodes, children and adolescents were requested to walk along the working space of the laboratory to ring a bell held by the mother. For each individual were recorded at least ten gait cycles. The GRF were normalized to the body weight. RESULTS: The spatiotemporal and movement parameters of children and adolescents with DS were in the range of the values reported in literature (Parker and Bronks, 1980) for the children with DS. In fact, we observed interindividual variability, low velocity and cadence, prolonged stance phase as well as increased values of flexion movements. The main EMG characteristics were represented, in the more younger children, by 1) coactivation of distal muscles during the stance phase 2) praecox activation of gastrocnemius lateralis beginning at the heel strike, 3) prolonged activation of biceps femoralis. The muscle activation patterns of older individuals of groups B and C still were characterized by a praecox activation of the gastrocnemius muscle and a prolonged activation of biceps femoralis. Regarding the GRF acting on the sagittal plane, 7/10 children in the group A showed only the first decelerative component. In the children of group B, we observed a progressive increment of the propulsive component, but still low. In the group C, the propulsive component was well represented. DISCUSSION: The development of gait is a milestone of motor behavior in individuals with DS which strictly influences the development of cognitive abilities. In a previous paper on the kinematics of gait of children with DS, Parker and Bronks (1980) have shown the presence of some specific kinesiological dysfunctions such as reduced ankle flexion at the swing phase, premature and prolonged plantar flexion during the stance phase prior to foot strike, reduced extension at the toe off (15 and 27% less than adult and infants respectively). These authors (Parker and Bronks, 1980) suggested the presence of an arrest or of a severely retarded development of the neural control of distal segments in these individuals. We have found, in **the** younger children (group A), the presence of a marked digitigrade walking with Coactivation of distal muscles and prolonged activation of biceps femoris. Moreover, in most of the younger and older children/adolescents we observed the presence of a praecox activation of gastrocnemius lateralis, which could be correlated both to a persistent immature pattern or to an eccentric activation of a shortened muscle.

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A Kinesiological and Electromyographic Analysis of Stair Descent of Individuals with Down Syndrome and Parkinson's Disease

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INTRODUCTION: The control of body position during stair descent is one of the more complex motor performances both in the younger children and older men as well as in individuals affected by diseases of the central nervous system. In children, stair descent is a milestone of motor development and its evaluation is included in the scale of Russell (1989). Moreover, several studies have reported about the difficulty of older people in the stair descent (Canavagh et al., 1997). In order to analyse the presence of a different neuromotor strategy between individuals affected by hypotonia and those with an extrapyramidal hypertonia, we have analysed and compared the electromyographic and kinesiological patterns of individuals with Parkinson's disease (PD) and with Down syndrome (DS) during the first step of stair descent. METHODS: Six individuals with PD (mean age 67.3 ± 5.4, 6 males and 1 female; 1 in the stage I of Hoehn and Yahr; 3 in the stage II; 2 in the stage III), six individuals with DS (mean age 18.3 ± 1.3; males 3 and 3 females, mild mental retardation) and 5 individuals without any apparent neuromuscular disease (mean age 46.0 ± 7; 3 males and 2 females) were requested to participate to this pilot study. The protocol consisted in a one step descent from a platform 17 cm high (4 1 x 36 cm) over a force platform hidden by a carpet. Kinematics of limb movements were monitored by an automatic TV image processor (ELITE, BTS, Milano, Italy) (50 Hz sampling frequency) which detected the instantaneous positions of passive retroflective markers applied onto skin landmarks of each individual (anterior-superior iliac spine, greater trochanter, knee, ankle and head of fifth metatarsus). Ground reaction forces (GRF) were measured by means of a force platform (Kistler, sampling frequency: 100 Hz). EMGs were recorded from muscles tibialis anterior (TA), gastrocnemius lateralis (GL), vastus lateralis (VL) and biceps femoralis (BF) by surface preamplified electrodes (500 Hz sampling frequency) of both inferior limbs. For each individual were recorded at least five descents with the same limb. The GRFs were normalised to the body weight. For the analysis, we have identified the following three main phases: A) 200 msec before lifting the swinging limb from the platform; B) by the lifting of the swinging limb from the platform until the ground contact; C) the stance phase. RESULTS: The analysis of temporal parameters has shown that the phase B was performed by the normal individuals in 580.0 ± 141.4 msec, by the individuals with DS in 600.0 ± 138.5 msec and by individuals with PD in 706.0 ± 180.0 msec. The phase C was performed by normal individuals in 670.0 ± 70.7 msec, by individual with DS in 670.0 ± 50.3 msec and by PD patients in 756.6 ± 215.5 msec. The analysis of GRF mainly showed, in the normal individuals, the presence of a decelerative phase followed by an propulsive component, with a very short transition component. Our preliminary results point out some variability in both groups of individuals with DS and with PD. The individuals with PD, and mainly

those more affected (stage III of Hoehn and Yahr, 1967) differ from the normal individuals for 1) a longer movement time of phase B (swinging phase) and C (stance phase); 2) an increased transition and a reduced acceleration phases of GRF 3) for a disturbed timing of muscle activation patterns, mainly during the phase A (before swinging phase) by activating the VL, the BF and the TA muscles instead of only the VL as we observed in the normals. Moreover, in our patients, the phase C (stance) was characterised by a marked activation of TA, VL and BF during the ¾ of the stance. On the other hand individual with DS showed a praecox and prolonged activation of GL and VL from the beginning of the phase A throughout the phase B. BF muscle was activated at the beginning of both phases B and C. DISCUSSION: Stair descent is a common action in the daily activities and its recovery is one of the goals of rehabilitative programs. Our results suggest that in individuals with PD an abnormal control of posture and an altered timing of muscle activations affect the execution of the requested performance, whereas in subjects with DS an hypotonic muscle, mainly defective during the eccentric contractions, could be responsible for the subtle differences in the muscle activation timing observed during stair descent.

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Stability of EMG Profiles During Gait, After Change of Gait Dynamics by an Ankle Foot Orthosis in Children with Cerebral Palsy

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INTRODUCTION

Ankle Foot Orthosis (AFO' s) are frequently used to correct equinus gait in Cerebral Palsy (CP). The AFO keeps the foot in dorsal flexion, thereby virtually shortening the leg. Since during midstance rise of the body is decreased, and during swing phase hip elevation can be lessened, the excursions of the center of mass are minimized when using an AFO. This will result in a higher walking speed and lower energy consumption. However, besides this biomechanical and physiological effects, also a change of the EMG profiles during walking is hypothesized, since propioception is changed as a consequence of the changed kinematics of the gait pattern.

METHOD

Nine CP-children (mean age: 6 years) were measured, which could all walk independently. They were walking on their shoes, in two conditions: with and without AFO. All children were using an individualized AFO (1). The following measurements were performed: surface EMG was recorded of the m. rectus femoris, m. semimembranosis, m. tibialis anterior and m. gastrocnemius (medial head), and processed to the linear envelope (i.e. the rectified EMG, smoothed at 2 Hz). Amplitude was normalized to the maximum of both conditions. Kinematics in the sagittal plane were derived using video and markers. The foot reaction force vector was measured by a force plate. From kinematics and foot reaction force the net internal moment of force was calculated for the knee and ankle. Also walking speed was measured and a heart rate monitor was used to measure the heart rate at rest and during walking, after proper stabilization. From these latter measurements the Physiological Cost Index (PCI) was calculated.

RESULTS

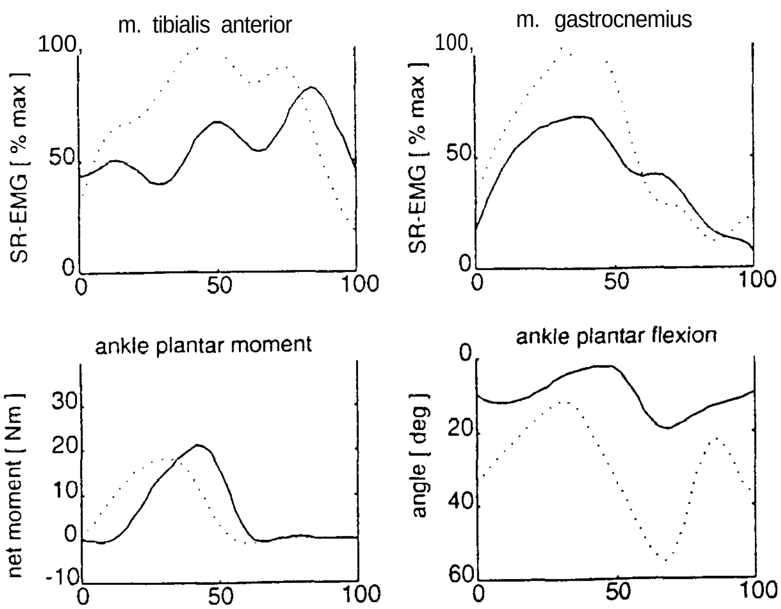

Figure I EMG profiles, kinematics and kinetics of the ankle(-muscles) during walking with (solid) and without (dotted) an AFO. Abscissa is the gait cycleji-om 0% at heelstrike to next heelstrike (100%)

The PCI significantly improved with a 34% lower heart rate per meter when using an AFO. Figure 1 and 2 show the EMG and biomechanical results for typical subject for ankle and knee respectively. The kinematics of the ankle changed towards a dorsal flexion position of the ankle. Also the range of motion is minimized, but not zero, indicating that the AFO still allows some motion. Knee flexion during early and mid-stance is enabled through the use of an AFO. Also the kinetics (net moments around knee and ankle) changed towards normal values, as a result of walking with an AFO, being consistent with the improved PCI. However the EMG profiles showed a variable response. Except for the m. rectus femoris, no significant changes in the profiles were seen, where in general the amplitude was lowered.

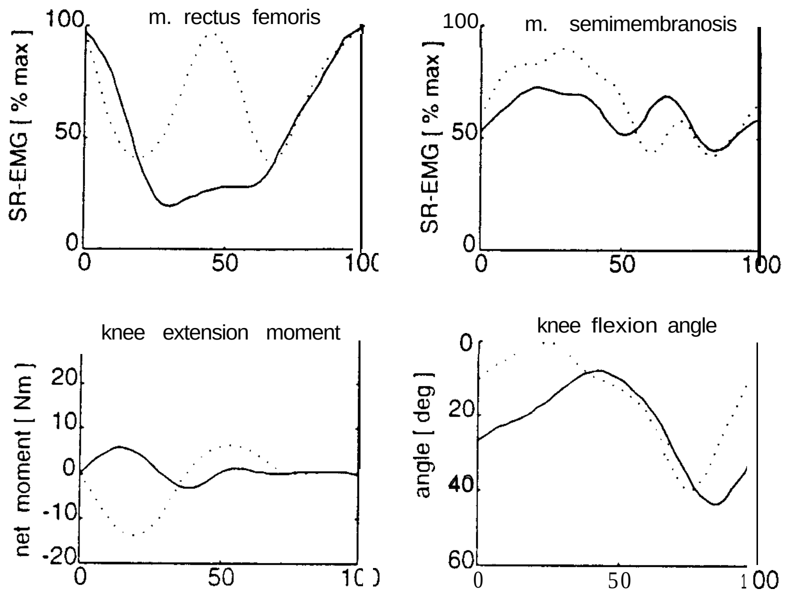

Figure 2 EMG profiles, *kinematics and kinetics of the knee(-muscles) during walking with (solid) and without (dotted) an AFO. Abscissa is the gait* **cycle** *from 0% at heelstrike to next heelstrike (100%).*

DISCUSSION

The different kinematics of walking with an AFO implies a change of proprioceptive input to the central nervous system. Also, the change in the pattern of joint kinetics must be compatible with changes in muscle forces, which requires a different activation pattern. Both mechanisms could explain the diminished burst of rectus femoris activity during the second part of the stance phase (2). For the muscles around the ankle it must be realized that stiffness of the AFO accounts for part of the internal moment. The relative stability of EMG activation patterns was also found in studies where muscle dynamics changed due to soft tissue surgery (3).

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The Effects of Serial Casting on the Forces About the Knee In Children with Spastic Hemi or Diplegia

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INTRODUCTION

Children with cerebral palsy present muscular imbalances, and typically develop limitations in range of motion (ROM) and muscle shortening. Clinically, muscle shortening most often occurs in the triceps surae muscle which is known to play an important role in ambulation(1). In normal gait, the position of the ankle at initial contact is that of slight dorsiflexion. This position results in a flexion moment at both the ankle and knee, the latter of which is controlled by a contraction of the quadriceps muscle (6) In the child with a shortened triceps surae, the ankle is consistently in a plantar flexed position (5). At initial contact, this ankle position tends to produce an extension moment at the ankle and knee joints (3). The resulting forces that act on the knee may eventually lead to a deformation known genu recurvatum (4).

The use of serial casting to lengthen the plantarflexors (triceps surae) in the child with spastic cerebral palsy is well established in the literature (2). To date however, no study has examined the mechanical changes produced by this technique on the knee, the joint which is potentially damaged when there is presence of gait with equinus. The purpose of this study is therefore to evaluate the effects of serial casting on the forces about the knee during gait in children with spastic hemiplegia or diplegia.

METHODS

Seven sujets (4 male, 3 female, 2 with diplegia, 5 with hemiplegia, mean age=7.39+/- 0.85 years) volunteered to participate in this study. Informed consent was obtained. All subjects participated in both a pre- and post-treatment evaluation which consisted of two parts: 1) a clinical evaluation by an experienced pediatric physical therapist; and 2) an evaluation in the Gait Laboratory of the Hopital Marie Enfant.

Clinical Testing- Evaluations consisted of the following elements (bilaterally): 1) active and passive ROM of all movements at ankle 2) muscle strength of the dorsiflexors 3) spasticity of the triceps surae [modified Ashworth scale] 4) a video tape recording of the child's gait in the sagittal and frontal planes.

Laboratory Testing-Simultaneous acquisition of kinematic and kinetic data were aquired at the hip, knee and ankle during gait using the ELITE 3-D motion analysis system. 15 trials were collected at each data collection session. Data was collected unilaterally, on the side that was more clinically involved. Time-distance parameters were calculated for stride length (cm), stride time (s), cadence (steps/min), stance time and swing time (% of cycle). Kinetic data were collected using a single force platform (AMTI). For all data, only values at 0, lo,20 and 30% of the gait cycle were used for statistical analysis.

DATA ANALYSIS

Clinical Testing-Clinical data were compared pre and post treatment using the Wilcoxon Signed Rank Test. For the analysis of the pre-recorded videotapes, the position of the knee and ankle at the 5 major phases of gait were compared (descriptive).

Laboratory Testing- 5 to7 gait cycles were selected for their consistency in time- distance parameters and were subsequently analyzed per evaluation. Means and standard deviations were calculated for each of the time-distance parameters, kinematic and kinetic data at each evaluation. Pre- and post-treatment values were compared using the Wilcoxon Signed Rank Test. All tests of significance were performed at the probability level of $p<0.05$.

RESULTS

Clinical Testing-As expected, clinical tests revealed a statistically significant increase in active (9.2” ± 6.4°, p=0.028) and passive (11.6° ± 4.1°, p=O.O18) ROM at the ankle. A significant change in strength was seen in the dorsiflexors post-casting (p= 0.028). Spasticity was decreased in 5 of the 7 subjects, but this reduction was not of statistical significance (p=O.O68). Analysis of the videotape data in the sagittal plane revealed a consistent improvement in the position of the ankle at heel strike, and at the knee during single support. No important differences were noted in the frontal plane.

Laboratory Testing- Post-casting, stride time decreased from 1.1 ± 0.18 to 1.03 ± 0.12(s) (p=O.O28) and cadence increased from 110.56 ± 15.91 to 118.53 ± 14.29 steps/sec (p=0.043). No significant changes were seen in the kinematic data despite a tendancy toward increased knee flexion and increased dorsiflexion during stance. The kinetic data remained relatively similar pre and post-casting except at the knee where a statistically significant increase in torque was noted at 10% of the gait cycle (p=O.O18).

DISCUSSION AND CONCLUSIONS

Clinical testing revealed results similar to those seen in the litterature with the exception of a predicted significant reduction in spasticity. There was a tendency toward an increase in knee flexion during stance, but this again, was not found to be significant. Contrary to our initial hypothesis, the laboratory testing demonstated that there appeared to be no initial excessive extention forces acting about the knee. Post-casting, the subjects presented an increase in knee flexion concurrent with an increased extention moment at the knee. This change in ambulatory mechanics may have been related to a weakness in the plantar flexors and /or knee extensors (quadriceps). The initial stiffness encounted at the ankle may have masked this weakness. Therefore, despite it’s effectiveness for increasing mobility at the ankle, serial casting did not decrease the extention forces acting on the knee in our subjects. The ankle-knee unit plays an important role in gait and must be studied further in order to permit the optimization of the use of techinical aids in children with cerebral palsy. This study was financed in part by a grant from the Reseau de recherche de Montreal et de l’ouest du Quebec.

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Effects of Stabilization Athletic Taping on Control Ankle

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INTRODUCTION:

A common treatment used on athletes ankle joint for improving joint stability is through the application of athletic taping (AT). Our previous studies have shown that ankle AT of healthy subjects offers a mechanical support (MS) during ankle inversion which quickly decreases over the time following passive standardized mobilizations (NS after 20 minutes) with KIN-COM machine. No correlation was found between the results of MS and the perception of stability. Almost of the athletes (N=77) who received AT as a means treatments were pleased with MS they received and did not find that ankle movements where affected in any way. If MS seems to be less than effective, but the consumer are still satisfied, there must be a reason for it's popularity? The study has been conducted to measure how the control of ankle movement is affected by this stabilization device.

METHODS:

Forty healthy subjects participated in this study (23+/-2,4yrs) Sense of position (SP), sensitivity threshold (ST) and the reaction time (RT) of AT (N=30) were randomly measured in pre and post standardized fashion without vision and audition. The M-Response and the soleus H-Reflex (H-Sol) was measured with vision (N=lO) Ss were seated with their right hip flexed at 40-60 degrees and the knee supported and flexed at 40-60 degrees such that the subject's lower leg was parallel to the ground. The subject's right foot was secured into the KIN-COM ankle plate with the ankle in neutral position. SP: the ankle was passively moved into inversion by the KIN-COM to a pre-set angle of 15 degrees at 5 degrees/sec. Ss were instructed to remember the angle. Upon completion of three cycles, the evaluator passively moved the ankle plate of the KIN-COM. The angle at which the participant stopped was recorded. ST: was detected by a square wave stimulus of 1 msec. These electrodes were placed between the first and second metatarsal area on the dorsal aspect of the right foot. RT: the peroneus brevis EMG was measure using and EMG with a standard protocol. A pedal switch acting as the trigger for a 1.2 threshold electrical intensity stimulus and they were asked to execute a rapid contraction of the right evertors immediately upon sensing the stimulus. A practice session was imposed. This testing block consisted of five trials in each condition. The M max soleus response and H-Sol were elicited on the tibia1 nerve and measured an experimental set-up using the classical methodology describe in DESMEDT (1973). The experiment was composed of 10 trials. The statistics significance was determined by paired two-tailed tests.

RESULTS:

Paired t-tests showed as significant increased of RT in a stabilization condition by 8% (235+/-34 and 253+/-38 msec, P<O.05) and a significant decrease by 21% in ST (2.2+/-.6 and 1.84+/-. 5 mA, P<O.01). However, there was no significant difference in the SP (14.1+/-1.9 and 13.6+/-l. 1 degrees) and the trend in decreasing precision was that of 4%.

DISCUSSION:

In fact, the simple RT increased by 8%, the inhibition of H-Sol by 12% and no positive change were noted on the SP (proprioception). However, the ST showed a **21%** improvement. This result alone cannot explain the popularity of taping. The decreased ST can be due to an improved conduction which is explained by a decreased skin impedance, increased sweat and improved direct contact of electrodes due to the application of tape. Unless the cutaneous information's consists of a compensatory feedback mechanism for the ankle motor control movement. The taping did not improve the SP. This result was different than some literature reviews (JEROSCH et al. 1995, PERLEAU et al. 1995). The popularity of taping is apparent, however it does not seems to enhance, improve or facilitate the function of the parameters used for the control of ankle movements.

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The Separation of Neurological and Physiological Strength Gains During Isometric Training

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Introduction:

The adaptations caused by training that result in increased muscular strength have been classified as physiological (increased muscle size) and as neurological (increased muscular activation). Moritani (1979) has shown a method that allows the separate quantification of these two types of strength gains with two assumptions. The first assumption is that the EMG-to-force relation is linear, and that EMG values have high day-to-day reliability. -The purpose of this study was to examine the utility of this method when the EMG-to-force relationship was not linear and when day-to-day reliability of EMG was not high.

Methods:

Nine healthy volunteers participated in an eight week training program that consisted isometric contractions. Ten MVC contractions were held for four seconds with a rest of 8 seconds between each repetition. Three sets of these 10 repetitions were performed three times per week for eight weeks. Integrated EMG (IEMG) of the biceps brachii and average isometric elbow torque were measured at levels of 0%, 20%, 40%, 60%, 80%, and 100% MVC for both the trained and untrained limb during five testing sessions. The testing sessions occurred prior to the training and following training weeks 2, 4, 6, and 8. The IEMG-to-torque relationship was plotted for each subject for each testing session with the torque values normalized to the MVC value in the first testing session. The mean torque and IEMG values were also plotted for the population and is partially shown in Figure 1. Best fitting nonlinear equations were fit to the IEMG and torque values to describe the relationships. The strength gains were partitioned into neural and physiologic components using the method of Moritani (1979) and also using the nonlinear equations.

Results:

All subjects increased their MVCs during the training period with a mean final MVC that was 136% of the initial MVC (strength gain of 36%). The mean data for all subjects for the first and last testing days is shown in Figure 1. The relationship between IEMG and elbow torque was nonlinear for each subject with EMG increasing at a greater rate than torque. This phenomenon resulted in an over-estimation of the neural strength gain when the method of Moritani (1979) was used and a linear relationship was assumed (Figure 1: Top). The fitting of a nonlinear equation revealed that only 17% of the strength gain could be attributed to neural factors. This study also found considerable day-to-day variability in the IEMG (not shown in Figure 1) compared to the earlier study which made estimations of reliable EMG-to-torque equations difficult in this study. The best fitting nonlinear equation to the data after two weeks of training was different than the pre-training equation which would be evidence of physiological strength gains very early in the training program.

Conclusions:

It is suggested that, in situations with poor day-to-day EMG reliability and nonlinear EMG-to-force relations, the IEMG be normalized to a maximum M-wave elicited by electrical stimulation. This would hopefully achieve more reliable estimates of changes in motor unit activation, more reliable nonlinear equations and more accurate estimates of neural strength gains.

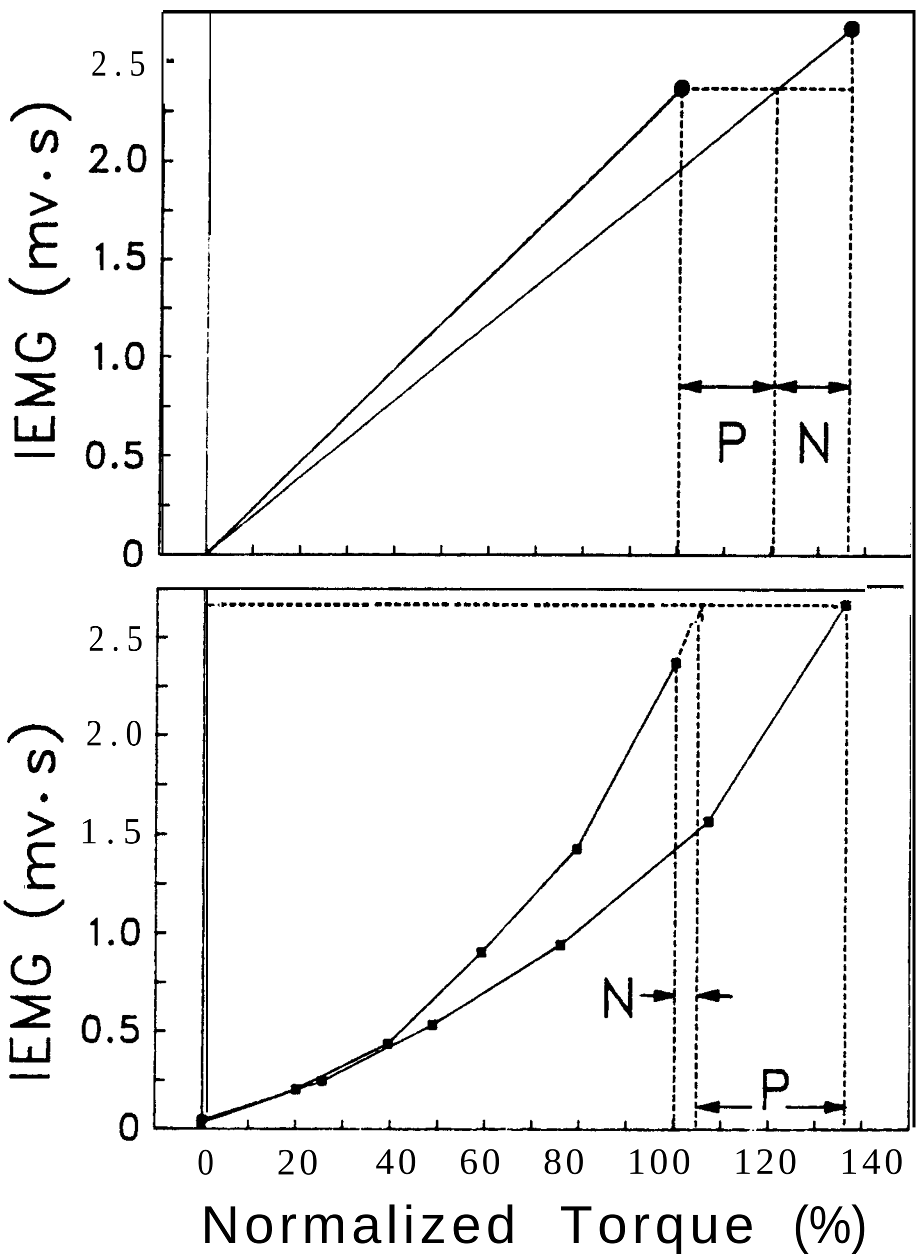

Figure 1. TOP: Separation of neural(N) and physiologic (P) strength according to the method of Moritani (1979). BOTTOM: Separation of neural and physiologic strength gain using nonlinear equations.

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Relationship Between the Medial Hamstrings and Knee Function During Jogging in Subjects with Deficient Anterior Cruciate Ligaments

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INTRODUCTION

Damage to the anterior cruciate ligament (ACL) is one of the most frequent knee related injuries (1). There is some controversy over the role of the hamstrings and quadriceps in this particular population (1, 2, 3,) Some researchers have shown an increase in muscular activity to compensate for the ACL deficiency, while others have suggested that there is a decrease in muscular activity. These inconsistent results may be due to different control strategies being used. None of the above studies, however, have distinguished subjects based on their level of function Self-assessment scores suggest some ACL deficient subjects are able to perform at higher levels compared to other ACL deficient subjects **(4).** Different motor control strategies may be linked to the different functional levels within this population. The purpose of this study was to compare EMG activity in ACL deficient subjects who were high functioning to those who were low functioning and also to determine if any differences between involved and uninvolved sides were evident.

METHODOLOGY

Eleven ACL deficient subjects participated in this study, 5 subjects in the high functioning group and 6 subjects in the low functioning group. ACL rupture was confirmed with arthroscopy or MRI. To assess function, a modified Noyes Questionnaire (4) was used where high scores indicated increased function. The subjects in the high functioning group had a symptom score of 90.4 $\pm$ 6.7 % and a function score of 90.4 $\pm$ 7.7 %. In contrast, the low functioning group had a symptom score of 52 $\pm$ 16.6 % and a function score of 68 $\pm$ 11.5 %. A manual laxity test, (maximal resistance) using the KT-1000, was performed with a mean anterior displacement of the tibia on the femur of 6.4 $\pm$ 2.7 mm for the high functioning group and 6.8 $\pm$ 4.6 mm. EMG activity was recorded using bipolar pre-amplified surface electrodes (Therapeutics Unlimited, Iowa City, IA) which *were* placed over the muscle bellies of the lateral gastrocnemius, medial and lateral hamstrings and the vastus lateralis. Pressure sensitive foot switches were used to determine heel strike and toe off. Subjects were asked to jog at 5 mph (controlled with the use of a tracking system) along an **11** meter walkway. Seven to 10 strides were ensemble averaged and normalized to the mean value and expressed as a percent of the gait cycle.

RESULTS

Results for the medial and lateral hamstring group demonstrated an emergence of two distinct patterns of EMG activity. In the high functioning group, both the involved and uninvolved sides demonstrated activity between 20 and 40 % of the gait cycle, as seen in Figure 1. However, in the low functioning group, 4 of the 6 subjects demonstrated minimal activity or no activity during the same time period, which is also depicted in figure 1. With respect to the other muscles tested, none of the other muscles appeared to show any significant differences in EMG activity between the two groups. In both groups, and in all muscles tested, there were no differences between the involved side and the uninvolved side.

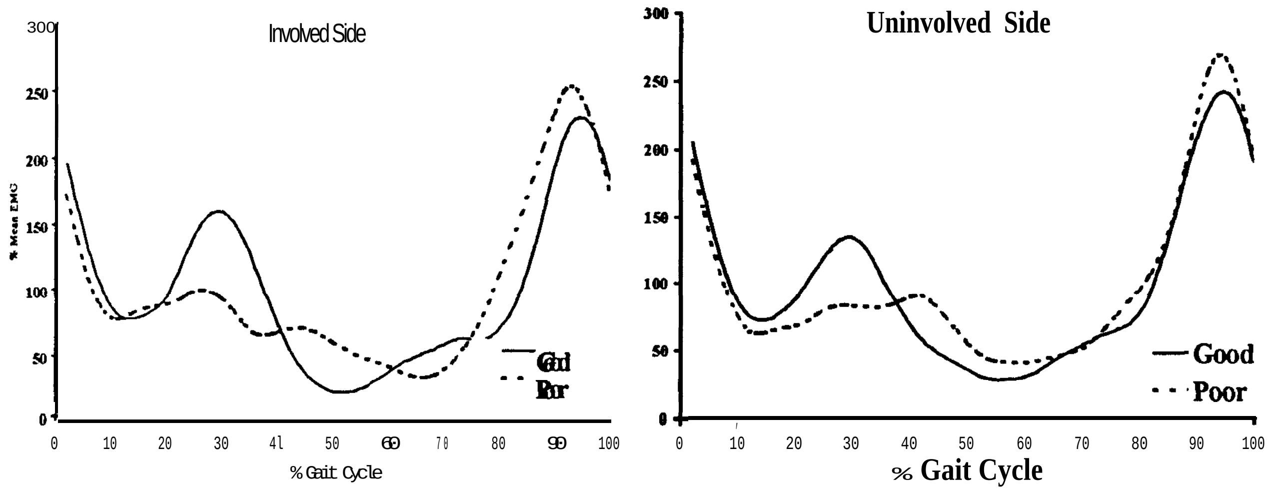

Figure 1. Normalized EMG activity for the Medial Hamstrings during jogging at 5 mph.

DISCUSSION

The results of this study suggest that there are at least two distinct patterns of EMG activity during jogging in this particular population. In the high functioning group, the data indicate that these subjects have adapted a control pattern that appears to be similar to non injured subjects. These subjects appear to use the hamstrings to assist with increasing the stability of the knee during the second half of stance phase. The emergence of different EMG patterns for subjects who are ACL deficient is in agreement with Shiavi et al (1992) who also demonstrated different EMG patterns for the hamstrings, as well as other muscle groups. Conversely, in the low functioning group, the vast majority of subjects appear to have difficulty adapting and exhibit EMG activity of the hamstrings which is different from the high functioning group and normal subjects. The data suggest that the activity of the hamstrings during the second half of stance phase is minimal, which suggests a different control strategy is being used which may be related to why these subjects are functioning poorly. The hamstrings are minimally active therefore may not be able to assist in providing much stability at the knee. Of interest was the fact that the data also demonstrate a high degree of symmetry between the involved and uninvolved sides. This is also seen in Figure 1. This symmetry was seen in all muscles and in both groups. From a control mechanism, it may be more efficient to try to develop some degree of symmetry during walking or jogging. The EMG profiles appear to be symmetrical, however, the amplitude information may provide more insight to this phenomenon.

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Time-Course of natural locomotor recovery in the first year following knee arthroplasty

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INTRODUCTION: Locomotor deficits following total knee arthroplasty (TKA) are characterized by a slower walking speed and abnormalities in muscle activation, joint movement and moment of force patterns [1-8]. Although large improvements in symptomatology and function occur following TKA, locomotor deficits have been shown to persist for more than nine years [4]. Few studies have, however, studied the time-course of natural locomotor recovery in the first year following TKA using longitudinal designs[2,6,7]. None have described concomittant changes in gait speed, kinematic and kinetic parameters and their relationships. The purposes of this study are to describe changes in speed, kinematic and kinetic variables of gait before and 2, 4, 6, and 12 months after TKA and to verify which kinematic and kinetic variables are related to speed.

METHODS: A cohort of sixteen patients (mean age: 67±10 years; 7 men, 9 women) with severe knee osteoarthritis who were scheduled for TKA, participated in the study. They gave their informed consent and performed a gait test before the surgery and 2, 4, 6 and 12 months after TKA. The gait test was carried out at free speed along a 10 m walkway. During the test, the sagittal movements of the hip, knee and ankle were recorded using a 2D-video system. Lower limb moments of force and mechanical powers were calculated using the inverse dynamic approach. Pressure-sensitive footswitches were taped to the heel, mid-foot and toe of both shoes to determine spatiotemporal variables. All signals were recorded simultaneously with a sampling frequency of 60 Hz and stored by a computer. For each patient, the mean performance of 3 gait cycles) was retained for analysis. Movement, moment of force and mechanical power profiles were normalized to 100% of gait cycle. Thereafter, mean values (n=l6 patients except for the last evaluation at 12 months, n=9 patients) were calculated at each 2% of the cycle and recorded in graphical form (Figure 2) to illustrate changes over gait cycle. The mean profiles (± 2SE) of 18 healthy subjects (mean age: 60±8 years; 9 men, 9 women) were used as normative data. Furthermore, the relationship between speed and kinematic and kinetic variables was studied with Spearman correlation coefficients. The kinematic and kinetic variables chosen for this latter analysis are the minimal and maximal values at the main transition points in the data as illustrated by the placement of the "r" values in Figure 2.

RESULTS: The locomotor performance, as measured by speed, kinematic and kinetic variables, improved from the first postoperative (PO) testing session (2 months PO) to the last one (12 months PO) (Figures 1 and 2). In general, the preoperative level of performance was reached only after 6 to 12 months PO except for a few variables (see hip movement and moment of force profile in Figure 2).
The largest improvements occurred in the first 4 months although slower but progressive changes were observed until the 12th month. Clinically important deficits persisted, however, at the last evaluation especially at the knee in the stance phase and the ankle in the energy generation period (around 45% to 70% of the cycle). Changes in gait speed

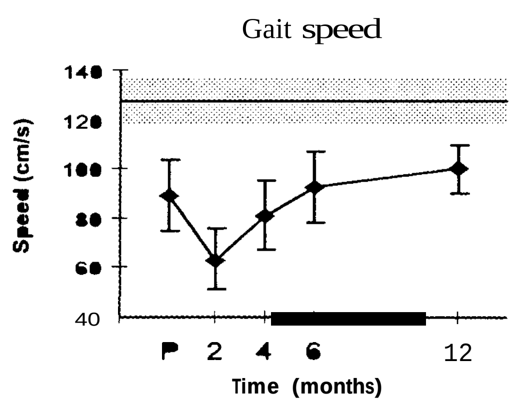

Figure 1. Mean gait speed values (±2SE) for a group of patients before (P) and 2, 4, 6 and 12 months after total knee arthroplasty compared to normative data (shaded area: mean± 2 SE).

were highly correlated with certain changes in kinematic and kinetic variables (for details see correlation coefficients reported in Figure 2) suggesting that the improvements in the former are mostly explained by changes in the latter. Changes in some relevant variables such as the maximum angle of knee flexion at 14-16% of the cycle (r=O. 14), the peak value of the knee extensor moment of force at 10%-l 6% of the cycle (r=.60), and the peak values of the knee mechanical power **(Kl :** 12%- 14% of the cycle, r==-.64; K2: 22%-24% of the cycle, r=.60) were not, however, related to changes in speed.

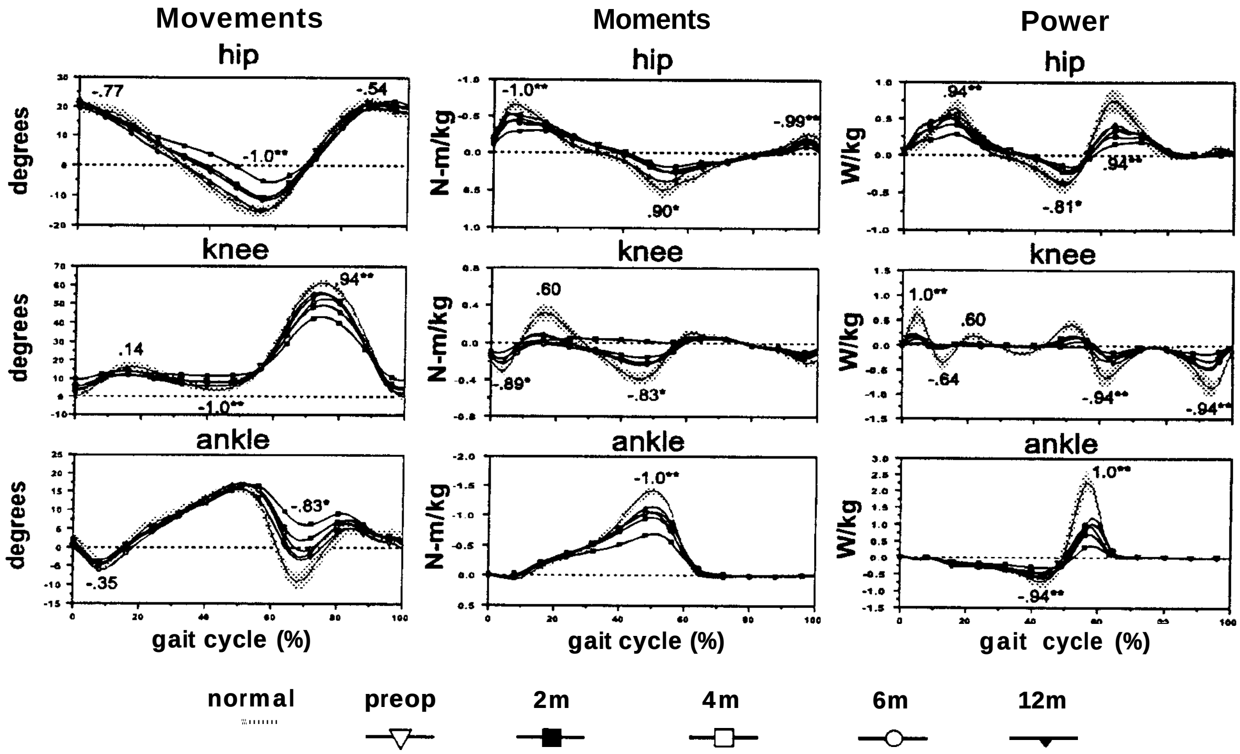

Figure 2. Mean movement, moment of force and mechanical power profiles of the gait cycle for a group of patients before (preop) and 2, 4, 6 and 12 months after a total knee arthroplasty are superimposed on the mean (±2 SE) of a group of normal subjects (shaded area). Spearman correlation coefficients (r values) between speed and the corresponding kinematic and kinetic variables are shown in the graph (*p<O.O5, **p<O.Ol).

DISCUSSION: These results confirm that locomotor deficits are still present one year after TKA. For the first time, the locomotor performance in the early stage after TKA was quantified and showed that at least 4 to 6 months are necessary to reach the preoperative level of locomotor function. Otherwise, the high levels of correlation between speed and the kinetic and kinematic variables confirmed that speed is a good indicator of global locomotor function. Kinematic and kinetic variables are, however, essential to pinpoint the cause of the residual deficits in speed. In this study, it was clearly shown that the operated knee did not recover, in the single limb support phase, in parallel with speed. As expected, the knee is a limiting factor to full speed recovery. Finally, the remaining deficits in the other joints, for example the deficits in H3, K3, A2 mechanical power bursts, are closely related to speed and perhaps less to the knee deficit.

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Active Stabilisation After a Lesion of the Anterior Cruciate Ligament of the Knee

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INTRODUCTION.

A lesion of the anterior cruciate ligament (ACL) is a major trauma of the knee. It frequently occurs in sports and is treated either surgically or conservatively. The rehabilitation program is mainly focused on regaining a stable knee. Previous research showed that neither the passive laxity nor the isokinetic strength of the knee muscles is a good predictor of functional outcome [Walla et al., 1985; Kramer et al., 1993]. However, clinical and biomechanical evidence is present for the hypothesis that higher levels of co-contraction of quadriceps and hamstrings provide an active Stabilisation of the knee to compensate for the lost ACL [e.g. Baratta et al, 1988; O'Connor, 1993]. To assess the level of co-contraction quantitatively, an individual EMG-force model is required in order to estimate the contribution of the moments exerted by each muscle (calibrated from the EMG signals) to the net joint moment. This EMG-force model includes information from literature, as well as data from subject- and trial specific calibration of the EMG-levels. In this paper, such a model is presented and evaluated in experiments with healthy and ACL-deficient subjects.

METHODS.

For the experiments, healthy subjects with no known history of knee problems and ACL-injured subjects participated. The ACL-deficient subjects minimally had to be able to perform a squat jump without any risk on recidive injury. The experimental protocol comprised two parts. First, subjects had to perform maximal knee extension and flexion contractions on an isokinetic dynamometer in order to calibrate the EMG-level of each muscle as a function of knee angle. Secondly, the subjects had to perform a one-legged vertical jump on a forceplate. For all trials, ground reaction forces, kinematics of the movement (with videorecordings) and surface EMG of seven leg muscles were registered. Off-line analysis of the isokinetic data comprised calculation of the calibration factors of each EMG-level as a function of knee angle. With these factors, the EMG-data during the jump were calibrated to muscle moments. With these values, the distribution of muscle moments to the net knee moment during the vertical jump could be obtained. Summation of these muscle moments gives an estimation of the net moment delivered about the knee during this movement Another way to calculate the net knee moment is the use of inverse dynamics by means of a linked segment model. For the latter method the input data are ground reaction force, position and antropometric data of the subject. In order to quantify the amount of co-contraction or active Stabilisation during the one-legged jump, the values of the moments of the agonist and antagonist muscles were used in an equation of Cocontraction Index (CCI) [Doorenbosch et al., 1995]. A schematic outline is shown in Figure 1.

DISCUSSION of RESULTS.

By means of the EMG-force model, the individual muscle moments were estimated for the vertical jump. In order to validate the calculation of the individual muscle moments during the jump with the EMG-force model, summation of the muscle moments was compared to the result of the net knee moment calculated by means of inverse dynamics. There appears a difference of about 15% between the net joint moment and the summated muscle moments. However, the model is in a preliminary stage and requires fine-tuning by more parameters than incorporated sofar. For example, the dependence of velocity on the EMG to force processing is subjected to further investigation.

The results of the experiments also show that the CCIs of the ACL-group was higher compared to the healthy subjects. This confirms the hypothesis that a higher amount of co-activation is developed as compensating mechanism to the loss of the passive stability.

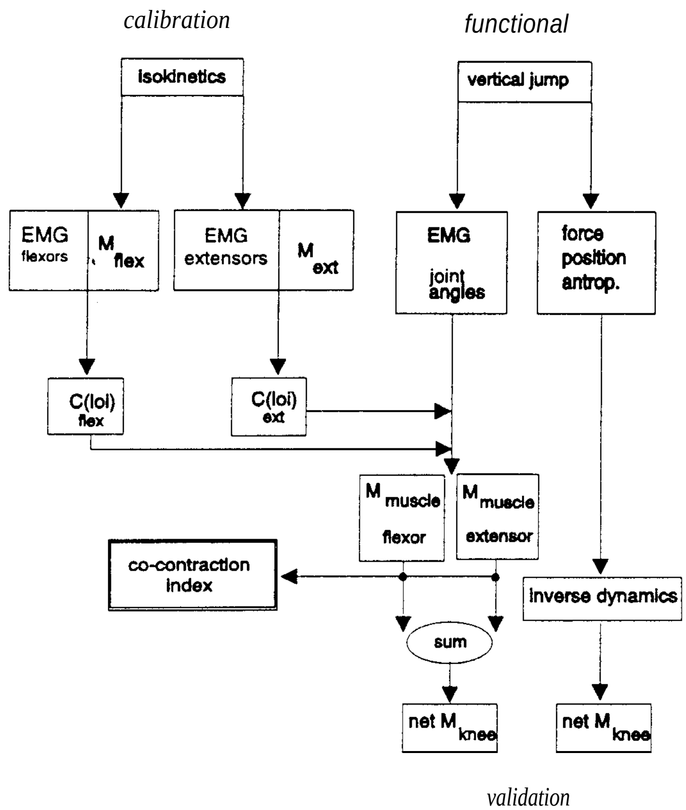

Figure 1. *Schematic outline of the EMG-force model used in this study. See text for further detail. Abbreviations: Mflex,Mext:moment generated by* ***flexor*** *and extensor muscles resp.; Cflex(Loi), Cext(Loi); calibration factors for corresponding muscle group as a function of muscle Iength(Loi); net Mknee: net knee moment*

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The Galvanic Stimulation of the Vestibular Apparatus as a Biofeedback for Improving the Posture Control in the Sensory Conflict Status

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The well known phenomenon of the galvanic stimulation effects of the Vestibular apparatus was studied in respect to give him a new possibility to transfer useful information able to correct the sensory interaction. This stimulation was used as a biofeedback signal in the posture stabilization task on the human additionally exposed to the visual disbalancing input. The biofeedback was specifically formed as an independent sensory channel giving the true information about the body sway from the local vertical using the precious accelerometer rigidly attached to the body. It was shown that this artificial independent sensory channel being mixed with the real Vestibular input can be accepted by the central nervous system as the real one. In conditions when the biofeedback gives the true information the summed multisensory input leads to the increased stability of the human in spite of the presence of the disbalancing visual input. The mathematical model of the interaction mechanism including the new appeared artificial sensory channel is given which explains experimental data concerning stabilization performance of the human in respect to different phase shifts in the biofeedback signal.

Sensory and Strength Training of Balance in Older Adults - A Preliminary Report

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INTRODUCTION

The ability to maintain balance decreases with age. It has been reported that older adults are less steady under sensory changing conditions. Furthermore, the ankle and thigh muscles strength is significantly weaker in the fallers than the nonfallers. According to the model of systems approach, it may be hypothesized that balance of older adults can be improved by optimizing the ability to integrate sensory information in a standing posture and by strengthening the ankle and thigh muscles. Previous study has shown that a multisensory balance-training program improved postural stability and the time to maintain one-leg stance balance in older adults. The multisensory balance training was conducted in an individualized manner. This study compared the changes in postural stability and weight shift ability in a training group and a control group from pretest to posttest. The exercise training program included both a multisensory component and a strengthening component.

METHODS

A total of 25 1 older adults were recruited for this study. Subjects were included in the training group if they (1) responded to the recruiting sessions for balance training, (2) were more than 65 years old, (3) had no history of stroke, Parkinsonism, or lower extremity orthopedic surgery, and (4) were able to attend the exercise classes. The control group subjects met the same recruiting criteria except that they were not informed of the balance training and that they agreed to participate in balance evaluation only. The results of 98 subjects who have completed both a pretest and a posttest are included in this report. The posttest was conducted at least 2 months after the pretest. Between tests, the 61 training group subjects (28 male, 33 female; averaged 70.8 years old) attended exercise classes whereas the 37 control group subjects (20 male, 17 female; averaged 7 1.2 years old) remained in their previous lifestyle. Each exercise class had a size of approximately 20 participants and 1 coach. The class met l-hour each session, twice a week, for 8 weeks. The exercise program included three parts: (1) Warm-up flexibility exercise. (2) Multisensory balance exercise with eyes open or closed, head still or moving while standing on a firm support surface or on a foam pad (8 cm thick, medium density). (3) Strengthening exercise for the thigh and ankle muscles using sandbags (0.5-2 kg) or Therabands.

The balance evaluation was conducted on a dual-plate force platform (Smart Balance Master, NeuroCorn Int Inc, Clackmas, Oregon, USA). Each evaluation session included the following tests: (1) sensory organization test, (2) rhythmic weight shift test, and (3) limits of stability test.

RESULTS

It was found that the postural stability (% maximal sway area) improved significantly more in the training group than the control group (p< .OOl). The postural stability significantly improved when the somatosensation and vision were simultaneously altered [Figure]. However, the control group also demonstrated significant improvement in postural stability from pretest to posttest (p < .05).

The rhythmic weight shift ability in the anteroposterior, but not the lateral, direction significantly increased their rhythmic shift weight amplitude at 1-, 2-, and 3-second paces (p< .005). The control group did not improve in this test. In the limits of stability test, both groups were able to shift weight and reach peripheral targets on the computer monitor faster during the posttest (p< .OOOl). However,

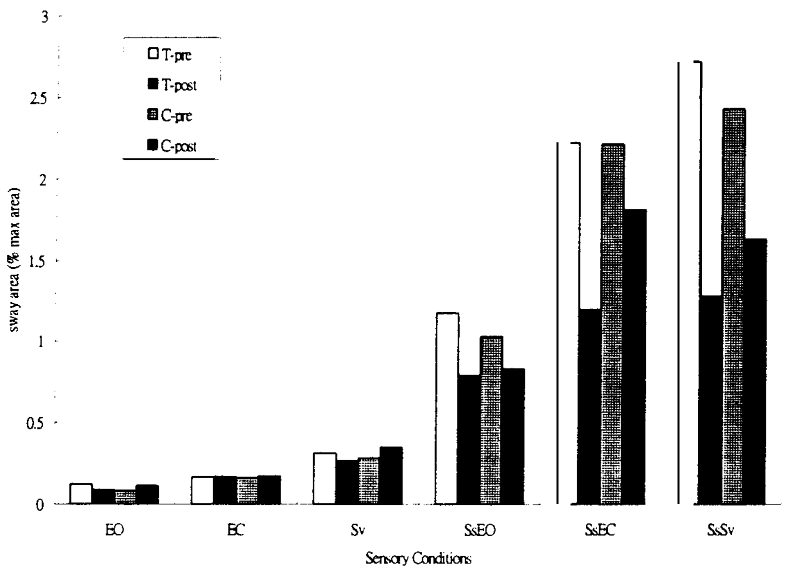

Figure. Sensory organization test results in the training group and the control group during pre- and posttests.

the path length of weight shift during this test did not show improvement in either group.

DISCUSSION

This study shows that multisensory balance training and strengthening exercise taught in a class are effective in reducing static postural sway under varying sensory conditions and increasing voluntary weight shift amplitude in the anteroposterior direction. It is suggested that the improvement of postural stability in the training group under varying sensory condition be due to the effect of multisensory training component of the exercise class. The improvement in weight shift amplitude anteroposteriorly may be related to the effects of strengthening exercise for the tibialis anterior muscle and of voluntary body swaying movement during the exercise-training program. The results support the model of systems approach to balance training and balance evaluation.

Acknowledgement. This study was supported by a research grant number DOH86-HR-6 18 from the National Health Research Institute to Dr. Hu.

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Postural Responses to Neck Muscle Vibration in Unilateral Vestibular Loss Patients and Normal Subjects

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INTRODUCTION

Muscle vibration is known to activate selectively Ia afferent giving rise to a false signal of stretch. In standing subjects, dorsal neck muscle vibration produces a body sway deviation forward [8.3.9]. Vibration of other posterior postural muscles induces a backward sway deviation [1,4]. It has been hypothesised that the difference in the direction of sway respective to the stimulated part of the body is due to the fact that neck proprioceptive signals are processed jointly with Vestibular input with the head taken as a reference [7]. On this hypothesis, one could expect that Vestibular dysfunction or Vestibular asymmetry would lead to a misinterpretation of neck proprioceptive signals. In the present study, we have investigated the postural responses during neck muscle vibration in unilateral Vestibular lesion patients healthy subjects with an asymmetric Vestibular tonus created by galvanisation of their vestibular system. Galvanic stimulation is known to activate the peripheral Vestibular afferent [2] and, in standing subjects, to produce a whole body tilt towards the anodal side [6].

METHODS

In the first part of the experiment, 19 normal subjects (24 to 49 years), and 13 unilateral vestibular loss patients (34 to 64 years; table 1) were tested with vibration. In the second part of the experiment, another 9 healthy subjects (28 to 49 years) were tested with vibration and galvanic stimulation. Sway was recorded with subjects standing on a force platform with their eyes closed. A small electromechanical vibrator (90Hz) was fixed over the medial superior aspect of the right or left trapezius muscle. The bi-aural galvanic stimuli were (0.5, 1 .O and 1 .5mA) applied with anode over the right mastoid. 5 bursts of vibration or galvanic stimulation of 4s duration were applied following a 4s baseline acquisition of 10s trial. Vibrations were applied unilaterally on both sides in the first part of the experiment and only on the right side in the second part. Galvanic stimulation was applied alone and combined to vibration. Parameters calculated were: Sway deviation, mean displacement from baseline level. Response direction, angle between response vector and sagittal plane (clockwise positive angles, 0°=forwards).

Patients	5	4	3	1
Disease	Acoustic Neuroma	Meniere' 's disease	Viral Neuronitis	Herpes Zoster
Nerve section	Yes	Yes	No	No
Caloric test	No caloric responses on the affected side to irrigation at 44" and 22°C			
History	From 4 months to 14 years for ail patients			

Table 1. Summary of patients.

RESULTS

Neck muscle vibration in normal subjects resulted in visible hole body sway forward. In patients, vibration on the side of intact labyrinth produced a forward sway similar to normals. With vibration on the side of affected labyrinth, patients swayed to the affected side. Normal subjects swayed -3.1±16.8° during right side vibration and 6.9±22.6° during left side vibration. Patients with normalised data (all patients has right lesion) deviated 23.1±40.5° during intact side vibration and 118.8±68.5° during affected side vibration. The response direction to vibration of intact side was normal whereas response direction to stimulation of affected side was significantly different from response to vibration of intact side (p<O.OOl) and from normals' response direction (p<O.OOl).

The postural response to simultaneous neck vibration and galvanic stimulation was an oblique sway deviation with a forward sagittal component and a rightward (toward the anode) frontal component. Angles of vector sway deviation are reported in table 2.

	Vibration	Galvanic stimulation			Vibration and Galvanic		
		0.5mA	1 .0mA	1 .5mA	0.5mA	1 .0mA	1 .5mA
Direction	-6±9	66±21	62±21	64±23	25±24	36±25	58±42

Table 2. Angle (in degrees) of vector sway deviation

There were no significant differences between response directions to galvanic stimuli at different current intensity. When both stimuli were combined, the response direction was significantly different only between 0.5 and 1 .5mA (p<O.05). Comparison of responses during galvanic stimulation and combined stimulation showed a significant difference at lower intensities only (0.5mA, p<O.OOl; 1 .OmA. p<O.05).

DISCUSSION

According to a previous hypothesis [7] the vibratory stimulus has decompensatory effect and could reveal a Vestibular asymmetry. In the case of unilateral loss of Vestibular function, the asymmetry of vestibular afference is similar to horizontal rotatory acceleration and tilt of the head to the intact side [5]. Vibration of neck muscle ipsilateral to the lesion implies a stretch of that muscle as if the head was rotating away from the lesion. This would confirm the Vestibular asymmetry and could induce a compensatory postural deviation towards the affected side. In contrast, vibration of the intact side would not confirm head rotation so there would be no lateral sway. To explain the interaction of neck vibration with asymmetry induced by galvanic stimulation, it is necessary to assume that galvanic stimulation gives a signal of head tilt so that compensation is predominately lateral with a little forward sway to compensate for the apparent extension of dorsal neck muscle.

CONCLUSION

This study showed that postural responses induced by neck muscle vibration were similar in the cases where a Vestibular asymmetry was present (unilateral Vestibular loss patients and galvanicaly stimulated normal subjects). Postural responses were directed laterally towards the affected or anodally stimulated labyrinth with small sagittal component.

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Antagonistic Co-contraction and Synergistic Co-activation During Closed Kinetic Chain Leg Extension at Two Pedal Height Positions

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INTRODUCTION. In the past few years, Closed Kinetic Chain (CKC) exercises have been advocated as a mean of promoting muscular co-contraction of both quadriceps and hamstring muscles at the knee joint (Lutz et al., 1990; Shelboume et al., 1992). During CKC exercises, leg extension involves simultaneous knee and hip joint extension. Extension of the knee is performed by the quadriceps while hip extension is performed by the hamstrings along with gluteal muscles. Thus, the biarticular hamstrings, a primary knee flexor, co-contracts with the quadriceps at the knee in this task because of its activation during hip extension. Consequently, the co-contraction level should be influenced by the desired hip extensor moment (Jacobs and van Ingen Schenau, 1992). For a leg press effort, the hip extensor moment can be enhanced by placing the pedal at a high position. Combination of knee and hip extension moments also affects synergistic co-activation of the quadriceps components. More precisely, the rectus femoris (RF) is less active in these combined efforts than the vastus medialis (VM) and vastus lateralis (VL). The purpose of this study was to evaluate the level of antagonistic co-contraction produced between hamstrings and quadriceps muscles as well as the level of synergistic co-activation of the quadriceps components during a leg press effort performed at two pedal heights.

METHODS. Nine normal subjects participated in the study. Electromyographic (EMG) signals were collected with active bipolar surface electrodes placed on the three superficial components of the quadriceps muscle as well as on the semitendinosus (ST) and long head of the biceps femoris (BF). Unilateral static leg press efforts were executed at two pedal height positions. In the low position (CKCL) the pedal was placed at hip joint level (fig. 1A) while in the high position (CKCH) the pedal was adjusted at the knee joint level (fig. 1B). In both positions, the knee joint (40”) and hip joint (90”) angles were similar. Subjects were instructed to hold these efforts at a level equal to the body weight.

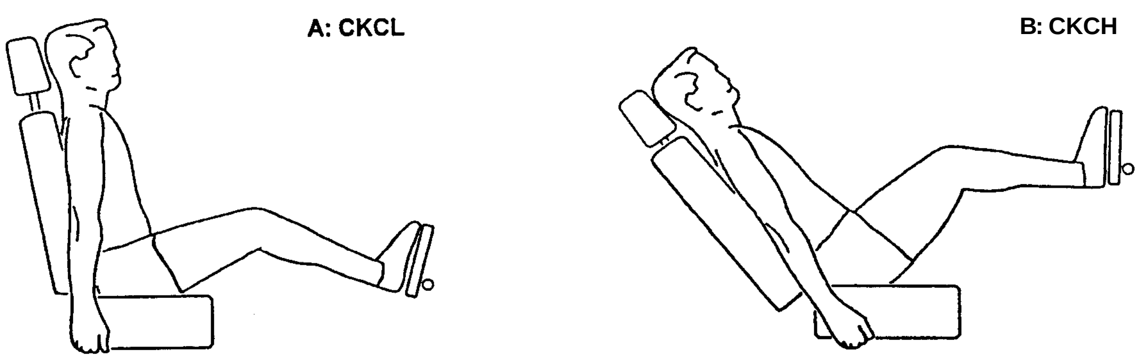

Figure 1: Low (A:CKCL) and high (B:CKCH) pedal positions evaluated in the present study.

In order to evaluate the hamstrings/quadriceps co-contraction and the quadriceps co-activation, ratios of EMG activities were calculated for pairs of muscles. In the computation of these ratios, moment equivalents of EMG signals recorded from each muscle during an isolated knee extension or flexion effort were used. This is a normalization procedure based on the moment-EMG relation. The effect of pedal height as well as muscle differences were verified using a two-way ANOVA for repeated measures with a level of significance set at 0.05.

RESULTS. Statistical analysis of results indicate that all three quadriceps components were significantly more active in CKCL than in CKCH. For both positions, BF was more active than ST. However statistical analysis failed to detect significant differences between pedal positions with regards to the co-contraction and co-activation ratios. Co-contraction ratios involving the RF (BF/RF, ST/RF) were higher than those computed for the vastii (fig. 2A). For co-activation ratios, those with RF were below l00%, thus inferior to VM/VL ratios which showed value around 100% (fig. 2B).

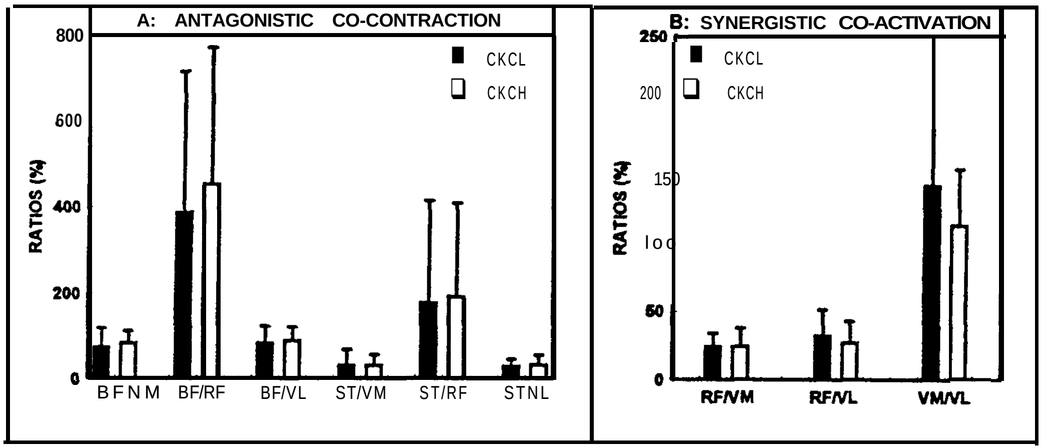

Figure 2: Antagonistic co-contraction (A) and synergistic co-activation (B) ratios calculated for the low (■) and high (Cl) pedal positions.

DISCUSSION. Pedal height affects quadriceps participation which can be explained by differences in force vector magnitude and orientation between pedal positions. In order to ascertain force vector orientation and intensity, the pedal would have to be instrumented with orthogonal force sensors. The fact that EMG values recorded from BF muscle were higher than those recorded from ST might be attributed to cross talk from the VL to BF as suggested by Koh and Grabiner (1992). Pedal height does not significantly modify the level of activity in hamstrings and the co-contraction ratios. Therefore the hypothesis suggesting a high pedal position to intensity the hip extension moment and hamstring muscle activity, is partially invalidated. For a CKC knee extension effort (leg press), co-activation ratios demonstrate that RF is less active or that there is more activity from VM and VL. Such results can be explained by the fact that RF is a bi-articular muscle. In order to develop a better biomechanical model optimizing recruitment of both quadriceps and hamstring muscles within a CKC effort, it is important to include mono- and bi-articular muscles acting on the hip joint, knee joint and even ankle joint together with mechanical measurements.

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Coactivation Differences in Men and Women During Isokinetic Knee-Extension Under Varying Speed

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INTRODUCTION

Research on Coactivation is rather limited, although a good understanding of the mechanisms would give us the opportunity to understand the ways of keeping the stiffness and the stability of the joints during contractions and to adjust our rehab protocols. In literature we find no unambiguous concerning the coactivation during isolated extension and flexion of the knee. The scientific explanation for coactivation is rather hypothetical. A review of the literature shows an enormous variation in protocols. Most of the studies are done with a very low number of participants and the positioning differs a lot, going from seated with different hipflexion angle to the prone position. The angular velocities are not higher then 240°/S except one, Ostemig et al 400°/S. To our knowledge the difference in coactivation between the sexes has never been done. In our study we wanted to know what the influence is of the angular velocity on the hamstring EMG coactivation activity and if there was a difference in the coactivation activity between men and women.

SUBJECTS

Fifteen women and sixteen men (age 2 1,3 y ± 1,5y ,) participated voluntary in this study. All of them were nonathletic university students free of any injury and of any complaint for the lower limb for at least three years.

INSTRUMENTARIUM AND PROTOCOL

A Biodex dynamometer (Promotion Zwijndrecht the Netherlands) was used for isokinetic testing. During the isokinetic measurement the EMG activity of four muscles was captured (three for the qceps and one for the hamstring). After informing the subjects there was a warming up for 10 minutes on a bicycle ergometer (90 Watt). After skinpreparation the subjectswere positioned on the chair with 90° of hipflexion. Before performing the isometric test the subject familiarised with the movement. The isometric test consisted of two repetitions (5 sec) of knee-extension and kneeflexion respectively at 30” and 60” of knee flexion. The testing protocol consisted of five maximal concentric efforts of the knee extensors and knee flexors at angular velocities of 60°/S, 180°/S and 300°/S. ROM was from 0” kneeflexion to 90” kneeflexion.

DATA ANALYSES

The Emg signals of the repetition with the highest moment was analysed. The range of motion of the knee extension was divided in three subperiods. Period 1 (60” - 90°kneeflexion), period 2 (30” - 60°), period 3 (0°- 30”) kneeflexion. Statistical analyses for the periods and the velocities was done with the One way ANOVA analyses. When there was a statistical significance both data were compared with an unpaired T-test. Difference in sexes was determent with the unpaired T-test. Significance was always set at $p < 0,005$.

RESULTS

The antagonistic coactivation exhibited a relative low level of activity compared with the high EMG discharge level in the extensors (agonists). Although we observe values going from 27% to 65% of the maximal isometric value during isometric flexion.Comparisons of the different subperiods (l- 3) for each angular velocity: At 60°/S there was no significant difference in antagonistic IEMG activity. At 180°/S, again no significant difference. Although there is a tendency of increase of the IEMG antagonistic activity in the third period. So the more the knee was extended the more the hamstrings showed activity. At 300°/S there is for the total population as for the subgroups, men and women, a significant difference between the first period (90” - 60” kneeflexion) and the third period (30°- 0”) ($p = 0.004$) and between the second (60°-30°) and the third period ($p< 0.001$). Comparison of one period (l-3) for the three angular velocities: There was no significant difference between the three velocities for the first and the second period. For the third period, there is no significant difference between 60°/S and 180°/S for the total population as for the two sexes separately. For the total population we see a significant difference between the 60°/S and 300°/S (p< O.OOl)and between 180°/S and 300°/S (P< 0.005). We also can determine a difference between the sexes. Although the EMG activity increases for both sexes for the third period, there is a significant difference between the 180°/S and the 300°/S for the men but not for the women.Comparison of the sexes: Under these conditions no significant differences were found.

DISCUSSION

During knee extension we also see an EMG activity of the hamstrings (coactivation). Although some authors mention a low antagonistic coactivity we measured values as high as 65% of the isometric values (60° kneeflexion). Of course the way of normalisation is an important factor that can affect the results. We normalised with the isometric 60” kneeflexion. The 30” periods (l-3) can be the reason why we did not find any significant difference for the three different angular velocities between the first and the second period. Many authors found a significant difference in IEMG activity between the beginning and the mid ROM , but they used smaller subperiodes ranging from 10” to 15°. In our study we found for all angular velocities that the mean IEMG activity was higher in the first period compared to the second, but not significant. Only at 3000/S the hamstrings activity increases significantly during the third period (30°- 0° kneeflexion). For the other angular velocities the increase was not significant. So we can state that there is a threshold velocity where the Emg activity reaches a significant higher value.. The higher the angular velocity the higher the Coactivation is. We think a possible explanation for the tendency of higher IEMG signals in the beginning and at the end is the fact that the tests are done in an open kinetic chain. This means we deal with inertia. Our hypothesis is that receptors, mechano, Renshaw cells and Golgi tendon organs cause a coactivational reflex which protects the knee So there is a need for Coactivation. The same explanation can be proposed for the higher IEMG activity of the antagonist at the end extension at high angular velocities. At the end we have a deceleration of the limb. This deceleration occurs due to the decrease in the activity of the agonists. But the higher the velocity the higher the inertia of the limb and the pad will be. So our hypothesis is that there are some receptors, mechano, Renshaw cells and Golgi tendon organs who activate the antagonistic muscletoprotect the subject from recurvatum. The increasing IEMG activity of the hamstrings in the third period (30” - 0” knee-flexion) with increasing angular velocity is completely opposite to the decreasing activity of the M. quadriceps . . This means that we can hardly believe that there is a common drive who excites the agonists and the antagonists at the same time during the complete ROM. From our data we found no significant differences between the sexes concerning the antagonistic EMG activity during knee extension. There was one important difference: as angular velocity increases the IEMG activity increases significantly during the end-extension for the men but not significant for the women (men from 38% to 65%, women from 39% to 54%). We believe this has something to do with the stiffness of the hamstringmuscles of the men. So while performing an extension the hamstrings of men will be sooner stretched than the ones from the women.

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Electromyographic Study of the Pectorialis Major, Serratus Anterior and External Oblique Muscles During Respiratory Activity in Humans

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INTRODUCTION

By the electromyographic activities of the pectoralis major (PM), serratus anterior (SA) and external oblique (EO) muscles it is possible, in accordance with Basmajian and De Luca (1) to get more information about their role in the movements to what they are mechanically related. For authors like Kapandji (2) Kendall at al. (3), they are, in the majority ventilatory mechanics studies , considered as accessory during the respiratory activity. The purpose of this study was to describe, in healthy humans with the use of surface EMG and by calculating the Root Mean Square (RMS) values with the use of computer, the effects of volunteers posture, respiratory movements and its different forms on PM, **SA** and EO muscles activities.

METHODS

Twenty males (mean age 22.55f6.64 yr.) and twenty females (mean age 21.25 ± 2.89 yr.) participated. With the use of a pneumotrace respiratory belt device, it was possible to control the respiratory movements by their chest wall displacements when asked to inspire (I) and expire (E) deep through the nose and maximally (forced = F), as well as during normal breathing (normal = N). The muscles signals were processed and the Root Mean Square (RMS) values calculated over a time period of 2.5 s. of activity. Bipolar surface silver/silver chloride electrodes (AS/AgCl) (Beckman Instruments, Inc.) were placed over the bellies of the PM (3 mm active diameter with a centro-centro distance of 1 .O cm), SA (3 mm active diameter with a centro-centro distance of 1.0 cm), and EO (9 mm active diameter with a distance centro-cento of 2.5 cm) muscles and parallel to its fibers and secured in place by adhesive discs. Mioelectric signals were processed by the EMG TECA TE-4 amplifier AA6 MK 11 (TECA Corporation, N. Y,), linked to a CompaqPC 486/DX4 – 100 MHz running the software AqDado and SisDin (Lynx Electronic Technology Ltd.). Two way analysis of variance (ANOVA) was used to test for differences in EMG RMS values between respiratory activities and body positions. A significance level of $p < 0.05$ was employed for all tests.

RESULTS

Descriptive data on male and female subjects in different positions, indicate that the EMG RMS values, for all muscles, increase with subjects in upright position and, for males, in both position these values are higher. Except for the EO muscle, when females volunteers inspired and expired in a quiet breathing way (N), these values were slightly low for males.

During the forced way of respiratory movement, for all muscles of subjects in both posture, the EMG RMS values were higher than in normal form of inspiration and expiration.

Except in a few case, either in N or in F forms of respiratory movements, the EMG RMS values related to inspiration are higher compared to those of expiration.

The experimental results indicated : 1) Significant differences ($p < 0.05$) in RMS between subjects positioned upright and in supine posture only for EO muscle during F movement, inspiring and expiring; 2) Significant differences ($p < 0.05$) in RMS between N and F movements for PM, SA and EO muscles during inspiration and expiration, either in supine or upright subjects position.

DISCUSSION

Healthy human subjects, here concerned, in upright posture, quietly breathing through the nose, showed, rhythmic and simultaneously, movements of the rib cage and the abdominal wall. Males and females volunteers among themselves or in comparison to each other sex, have not presented considerable difference in their thorax or abdomen movements. This is in accordance with Maccagno (4) when, in his study about types of respiration, have considered that, in fact, there is a mixed type. That is, subjects who do not present a predominance of neither abdominal nor thoracic wall expansion. We conclude that subject positions and forms of respiratory movements have a significant effect on respiratory muscles activities and that with a standardized method it is possible to evaluate their functions. Reporting normative data related to young, middle-aged and elderly persons, also provide therapists means for comparing normal scores with those of their patients. So, additional research is necessary to further check the effect of these variables on respiratory muscles activity of patients with some kind of respiratory failure.

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The Influence of Trunk Modelling in 3D Biomechanical Analysis of Simple and Complex Lifting Tasks

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INTRODUCTION

The study of complex lifting tasks involving lower limb movements require the use of a downward inverse dynamic analysis (IDA) to estimate the L5/Sl net loading. However, this modelling approach brings challenges associated with the correct estimation of segment inertial parameters (SIP) of the trunk (Kingma et al., 1996) and the consideration of the flexible properties of the trunk (Erdmann, 1997), especially during large amplitude movements performed during some lifting tasks. The purpose of this study was to evaluate different methods to estimate the SIP and to subdivide the trunk to reduce net loading errors in IDA of lifting tasks. It was hypothesized that (1) a geometric trunk model would produce smaller errors than a proportional trunk model, and (2) a 3-segments trunk model would produce smaller errors than a 2-segments trunk model.

METHODS

Twenty-one male subjects constituted three subgroups (N = 7) of subjects. Their mean age was 40 yr (range 35-45 yr), their mean height, 1.74 m (range 1.63-l .86 m), their mean mass, 72 kg (range 61-90 kg), and their mean body mass index (BMI), 23.8 (range 20-29 kg/m^2). The grouping was based on a ratio between the antero-posterior diameter at L5/Sl and height (Group A, B and C representing low (range: 0.1 IO-O. 124), medium (range: 0.127-o. 134) and high (range: 0.136-O. 158) ratios respectively).

Four lifting tasks were performed. In task Tl, the subject lowered and lifted a 12 kg box in the sagittal plane with a maximal flexion of the trunk. In task T2, a 12 kg free weight held in the right hand was lowered and lifted in the frontal plane (maximal right lateral bending). Tasks Tl and T2 were performed with the knees and elbows extended. In task T3, the 12 kg box was lifted and lowered (free lifting style) in the sagittal plane from the floor to the hip level. Task T4 was similar to T3 but the box was lifted to a shelf (height at hip level) located on their right side.

Five cameras (digitization of 27 skin markers at 30 Hz), two force platforms and a dynamometric box provided inputs to three full body (13 or 14 segments) 3D dynamic multisegment models. Two modelling parameters of the trunk modulated these models. Firstly, the SIP of the trunk were estimated with a geometric (Yeadon, 1990) or with a proportional (de Leva, 1996) anthropometric model, Data from Liu et al. (1971) were scaled and used to adjust the SIP of the three trunk segments from de Leva (1996) to get comparable segmentation planes at hip, L5/S 1, T12/Ll and C7/T 1 levels. Secondly, the trunk was subdivided in two segments (pelvis, abdomen + thorax) or in three segments (pelvis, abdomen, thorax). The SIP of the feet, shanks, thighs, arms, lower arms, hands and head were estimated with the data from de Leva (1996). The three multisegment models were constituted according to the following trunk modelling characteristics: (Ml) Geometric & 3 segments, (M.2) Proportional & 3 segments, and (M3) Geometric & 2 segments.

The behavior of these multisegment models was assessed with two error analyses (Kingma et al., 1996). In the error analysis I, the absolute difference between the L5/Sl moments computed with two linked models (lower body model vs upper body model) was computed. In the error analysis II, the absolute difference between external moments and inertial moments (rate of change of angular momentum) was computed using a procedure described in Plamondon et al. (1996).

Root mean square differences were computed on the entire task performance for each error analysis, lifting task and subject. A mean error was also computed for two postures of the trunk (upright and in maximal flexion). This error was calculated with 10 samples for each trunk posture to get more reliable results. Two two-way ANOVA with repeated measures on one factor (trunk models) was **used** to assess differences (a = 0.05) in the mean errors between groups (3 levels) and between trunk models (2 levels for each main effect: a) **Ml vs M2** and b) Ml vs M3). Differences between groups were not reported in the present paper. Post hoc test were performed with Student Newman-Keuls (a = 0.05).

RESULTS

Statistically significant differences were observed for **Tl, T3** and T4 (Table 1). The geometric trunk model led to smaller errors (up to 8.70 Nm) than the proportional trunk model. The 3-segments trunk model led to significantly smaller errors (up to 3.64 Nm) than the 2-segments trunk model especially during a flexed trunk posture. The significant differences were mainly observed for group C.

Table 1. Identification of error differences observed for the Tl,T3 and T4 lifting tasks

			Error analysis I			Error analysis II	
T	Trunk Posture	A’	Anthropometric model	Number of segments	A^2	Anthropometric model	Number of segments
Tl	Upright	T	C: Ml<M2(1.37)	/	X	C: Ml< M2 (1.68)[3]	/
Tl	Flexed	/	/	/	X	C: Ml< M2 (7.05)	/
T3	Entire	/	/	/	X	C: Ml< M2 (4.15)	/
T3	Flexed	/	/	/	X	B,C:Ml< M2(2.55;6.72)	/
T4	Entire	S	C: Ml< M2(1.00)	/	/	/	/
T4	Entire	T	C: Ml< M2(3.31)	/	X	C: Ml< M2 (3.92)	/
T4	Upright	S	/	A,C:Ml< M3(1.32;1.38)	X	C: Ml< M2 (1.12)	A,C:Ml< M3(1.33;1.28)
T-4	Flexed	L	/	C: Ml< M3 (2.00)	/	/	/
T4	Flexed	S	/	C: Ml< M3 (2.30)	Y	/	C: Ml< M3 (3.64)
T4	Flexed	T	C: Ml< M2(7.56)	/	X	B.C:Ml< M2(3.58:8.70)	/

1) L5/Sl moments were expressed about the anatomical coordinate system axes (A: axis): Longitudinal (axial rotation). Sagittal (lateral bending). Transverse (extension); 2) Moments were expressed about the laboratory coord. system axes that were oriented latero-laterally (X), antero-posteriorly (Y) and vertically (Z) relative to the subject at the initial position of each task. ; 3) Interpretation: a statistical difference was observed for the group C where the Ml mean error was 1.68 **Nm** inferior than the M2 mean error.

DISCUSSION

The use of a geometric anthropometric model and the subdivision of the trunk in three segments were two efficient trunk modelling methods to reduce errors resulting from the IDA of lifting tasks. The significant differences were sometimes small but are required to increase the sensitivity of upper body linked models applied to lifting analysis. The principal effects were mainly observed for subjects with a larger abdomen (group C) because (1) a proportional model is not suitable to account for individual anthropometric differences and (2) a 2-segments model cannot locate the center of mass of the whole trunk conveniently when the trunk is flexed, especially when large trunk segments are considered.

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The Role of the Evertors in Sudden Inversion and Gait

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INTRODUCTION

Ankle ligament injuries are very common in recreational sports and activities. Approximately one sprain occurs per 10,000 persons each day (1). In most cases, ankle sprains result from landing on an object that produces an unexpected torque about the ankle joint. A ligamentous injury to the ankle may occur when this torque is applied at a rate that exceeds the minimum time necessary for the neuromuscular system to respond. The purpose of this study was to describe the role of the peroneous longus and brevis muscles in sudden inversion and gait.

METHODS

Sixteen uninjured subjects (age = 27.7 yr, mass = 74.0 kg, height = 178.9 cm) volunteered to participate in this study. Surface electrodes were placed over the peroneous longus and peroneous brevis muscles of the right leg. Ten trials of: sudden inversion and treadmill walking at 3 mph were recorded. The EMG signals were recorded using a Noraxon Telemyo system at 1000 Hz. Sudden inversion was produced with an inversion platform that inverts the ankle 35°. Penny & Giles goniometers were used to monitor ankle inversion/eversion and the position of the inversion platform. The latency of the peroneous longus and brevis muscles was measured for each trial of sudden inversion. Footswitches were placed on the heel and toe of the shoe to identify heel strike and toe off during gait. The EMG data for the peroneous longus and brevis muscles during gait were normalized to isometric MVC and analyzed in 50 ms intervals beginning 50 ms before heel strike and concluding 500 ms after heel strike to quantify the preactivation and activation of the evertors during gait.

Table 1
Normalized Peroneal Activation Levels by Stance Time (%MVC)

Stance Time	Peroneous Longus		Peroneous Brevis	
	Mean	SD	Mean	SD
-50 ms	27.0	13.9	24.6	15.8
50 ms	23.4	11.3	23.2	11.5
100 ms	23.7	18.1	18.1	13.2
150 ms	29.5	22.9	21.5	21.4
200 ms	26.2	24.4	21.9	19.9
250 ms	26.3	21.5	24.4	21.5
300 ms	30.1	20.9	26.9	19.4
350 ms	36.4	25.1	30.4	22.6
400 ms	46.9	28.5	38.0	26.1
450 ms	45.2	24.3	37.2	22.6
500 ms	40.5	24.0	33.1	16.0

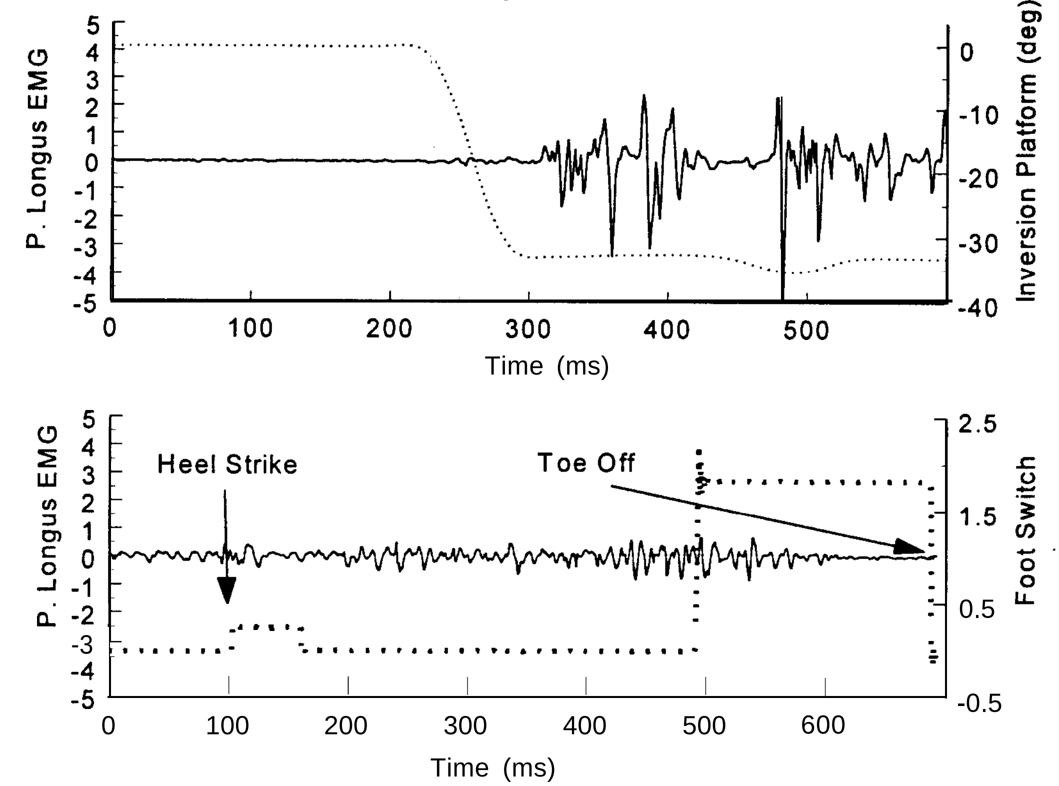
RESULTS

A typical EMG response to sudden inversion and waking is shown in Figure 1 for the peroneous longus. Mean peroneous longus latency was 76.5 ± 13.6 ms. Mean peroneous brevis latency was 78.5 ± 11.1 ms. Mean eversion response time was 114.8 ± 39.9 ms. Eversion response time was defined as the delay between the release of the platform and the first ankle eversion following platform drop. The normalized EMG for the evertors prior to and following heel strike during treadmill walking are shown in Table 1. Prior to heel strike the peroneous longus and brevis are preactivated to 27.0 and 24.6 %MVC, respectively. Maximal myoelectric activation (46.9 and 38.0 %MVC) of the evertors in gait occurred at 400 ms after heel strike for the peroneous longus and brevis, respectively.

DISCUSSION

Ankle sprains frequently occur during sport and recreational activities. The role of the peroneal muscles in protecting the ankle joint from injury is dependent upon the amount of preactivation of

Figure 1



the evertors prior to ground contact and the response time of the neuro-muscular system to a stimulus. The results of this study indicated that the peroneals were pre-activated less than 30% MVC prior to ground contact in walking. This minimal peroneal pre-activation provides very little protection against injury for the ankle joint (3). Significant changes in the muscle moments due to stretch reflexes do not occur during the first 100 ms after stretch (2,4). In the ankle joint an entirely unexpected disturbing force that catches us off guard meets with very little resistance. The minimum time for the ankle joint to respond to an inversion torque was found to be 114.8 ms. This response time is due to peroneal muscle latency and electromechanical delay. In conclusion, the ankle joint evertors offer very little protection against an inversion injury due to lack of muscular preactivation prior to ground contact and the neuromuscular response time to an injury stimulus.

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Postural Muscles Activation Associated with Isometric Ramp Efforts

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INTRODUCTlON

Maximal isometric ramp efforts have been reported to induce dynamic postural adjustments (Le Bozec et al., 1997). Indeed, it has been shown that the push force increases proportionally to the antero-posterior reaction force, which measures the global effect of muscular activity along this direction. Moreover, the push force increases also proportionally to the antero-posterior displacement of the centre of pressure. Therefore, it has been assumed that the push force is a dynamic perturbation of body balance which is counteracted by forces developped in the postural chain. To test this assumption, myoelectrical activities were considered. Focal as well as postural muscles were studied during maximal isometric ramp efforts with the aim to define the muscular sequence.

METHODS

Subjects were seated on a custom-designed device (Lino, 1995) which allowed measurement of the antero-posterior component of reaction forces (Rx) and of the displacement of the center of pressure (Xp). They were instructed to exert on a bar maximal horizontal pushes with their upper limbs. A dynamometric bar was used to obtain the horizontal force, Fx, exerted on it. Subjects were asked to perform the pushes from zero up to maximal force, as rapidly as possible, and to maintain it for 1.5 s. Measurements were made for a contact area between the seat and the thighs, including 100% of the ischio-femoral length, the foot support being bipodal. The subject's posture was controlled before each trial, the upper limbs being horizontally stretched out, with the hands gripping the bar.

Seven right-handed subjects participated in the experiments. Electromyographic activities were picked-up from the dominant side by bipolar surface electrodes. Fourteen muscles were selected for each subject in relation to the role they play in the horizontal push (see table 1) : a) the primum movens (SA), a co-agonist (DA) and a shoulder fixator (TS); b) muscles of the focal limb, which cross the elbow and wrist joints (BB, TB, ECR, FCR); c) postural muscles of the lower trunk and pelvis (TFL, GM, ES, OE and BF); d) postural muscles of the lower limbs crossing the ankle (TA and GL). All the signals were recorded with a sampling rate of 800 Hz for EMG and biomechanical data. The onset of EMGs were measured in relation to the force increase and then expressed in relation to the onset of the prime mover (SA).

RESULTS

The results showed for each subject a consistant excitation pattern. The EMGs rapidly increased, in the focal as well as in the postural muscles, which all remained active till the end of the effort. The sequence started with the muscles of the postural chain and ended with the elbow and wrist fixators (see Fig. 1). More precisely, two different sets of muscles were observed : the postural muscles, ranging from TA to GM (-100 to -71 ms before the push onset) and the focal muscles, including the shoulder and upper limbs muscles, ranging from DA to FCR (-59 to -50 ms before the push onset). Therefore, the postural muscles activity preceded the focal muscles by 42 to 14 ms (see table 1). These data suggested a distal to proximal recruitment order. The EMG onset of the last muscle (GM) of the postural sequence was significantly superior to that of the first one (DA) of the focal sequence ($t=2.78$; $p<O.01$). Therefore the postural set of muscles was compared to the focal one. The difference was also highly significant ($F(1,160) = 256$; $p< 0,OOOl$).

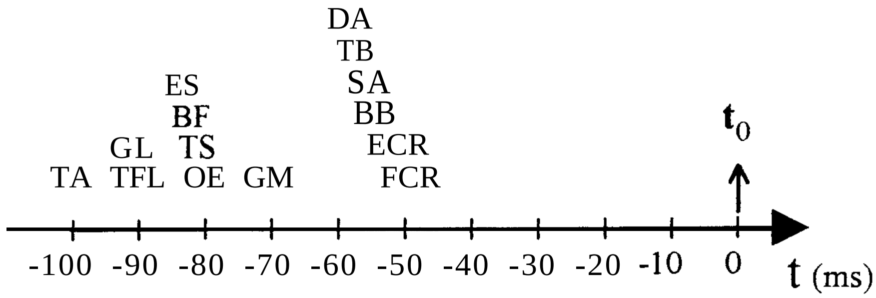
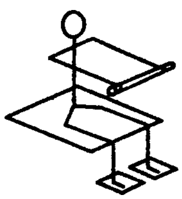


Fig. 1. Sequence of muscular activations. Latencies are referenced to the push force onset (**t_0**). Negative latencies refer to EMG onsets occuring before the push force.

	TA	GL	TFL	ES	BF	TS	OE	GM
m	**-42**	**-35**	**-33**	**-25**	**-24**	**-23**	**-21**	**-14**
s	12	13	11	7	11	8	9	9



	DA	TB	BB	ECR	FCR
m	**-2**	-1	+1	**+4**	**+7**
s	8	8	7	9	8

Table 1: EMG onset latencies. Tibialis Anterior (TA), Gastrocnemius Lateralis (GL), Tensor Fasciae Latae (TFL), Erector Spinae (ES), Biceps Femoris (BF), Trapezius Superior (TS), Obliquus Externus (OE), Gluteus Maximus (GM), Deltoideus Anterior (DA), Triceps Brachii (TB), Biceps Brachii (BB), Extensor Carpi Radialis (ECR) Flexor Carpi Radialis (FCR). Means (m) and standard deviations (s) in relation to the Serratus Anterior (SA) onset (in ms). Negative values indicate anticipatory EMGs.

DISCUSSION

Three results can be stressed : 1) there are postural phasic activations which last for the whole effort : postural forces are associated with the push force; 2) there is a consistent sequence, which originates from the support base, the foot-rest, then the seat and finally the bar: a distal to proximal progression is followed; 3) there are anticipatory EMGs with reference to the primum movens: as the postural muscular pattern precedes the focal one and the push onset, it cannot be triggered by the focal muscles and it is supposed to be centrally programmed. This suggest that the postural synergy would allow favourable postural conditions to be created in advance of the force onset.

In conclusion, transient push efforts require increasing EMG activations in postural as well as in focal muscles. The EMG sequence starts with the postural muscles. These results establish the origin of the postural forces associated with the voluntary effort. They confirm the hypothesis that the perturbing effect of the effort has to be counteracted by the postural chain in order to allow the effort to be performed efficiently. Finally they extend to isometric conditions the data obtained for upper limb movements (Bouisset and Zattara, 198 1; Lino, 1995).

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A Moving Balance Platform for Tracking of Center of Pressure

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INTRODUCTION

The center of pressure (COP) fluctuations during quiet upright standing demonstrate both stochastic and deterministic properties suggesting that both open- and closed-loop modes of control are involved in the maintenance of upright stance (Collins & De Luca 1993). The stabilogram-diffusion analysis procedure proposed by Collins & De Luca (1993) extracts parameters from the COP signals that reflect temporal as well as spatial properties of the postural control system. These parameters appear to vary across age as well as between categories of patients with different balance problems. A novel way of exploring the control properties of the postural system would be to expose it to perturbations with characteristics similar to the control system itself. The current presentation describes a PID controller implemented on a moving balance platform designed to reproduce COP fluctuations either recorded during quiet stance or simulated through mathematical modeling. The system will be used to explore the open and closed-loop behavior of quiet upright stance in healthy subjects and in patients with Vestibular disorders.

METHODS

A PID controller was implemented on a custom built balance platform BALDER (BALance DisturbER). BALDER consists of a force-plate (AMTI) built into a metal frame xy-table (2.1x2.1 m) which can be moved in the horizontal plane by two AC-servo motors controlled through two linear servo drivers. The operator controls BALDER from a PC running Windows-95. The platform has a movement range of 0.4 m in the horizontal plane and can accelerate up to 1.5g providing peak velocities of over 1 m/s. Positioning accuracy is better than 1 mm. The force plate provides information about reaction forces in all three planes and COP in the horizontal plane. In addition, the platform is equipped with a 3 dimensional accelerometer and two linear potentiometers for positional information. The PC is equipped with a 16 channel A/D and two channel D/A data acquisition board (Microstar 3200e/415). This board is instrumented with an Intel i486DX4 processor (96 MHz), which can be used for real time processing of data through a proprietary operating system (DAPL2000). In the current project ground reaction forces were collected while a subject was standing quietly on the BALDER platform for 30s. COP was calculated off-line and used as a set-point signal for the PID controller. In addition to the closed-loop control mode described here, the platform can perform in an open-loop mode control to produce discrete linear or oscillatory movements.

RESULTS AND DISCUSSION

A PID controller was designed to accurately track COP signals acquired during quiet stance. During quiet upright stance the main power of the COP signal is usually below 1 Hz. Thus, the controller was fine tuned for these frequencies. The output of the controller is described by:

$$Output = P \cdot e_m + I \cdot \int e_m dt + D \cdot \frac{de_m}{dt}$$

were e_m is the error signal defined as the difference between the current input and set-point signals (the pre-recorded COP). The parameters implemented in the final controller were P = -0.7, I = -0.001.

and D = -9, respectively. Figure 1 shows an example of the performance of the PID controller with theses parameters implemented and the set-point signal being COP fluctuations recorded during quiet stance (Fig. 1, left). The mean error during 30 s of tracking the COP acquired from quite stance was very low, 0.08±0.20 mm with a range between -0.67 and 1.14 mm. The third trace in Figure 1 shows the COP produced by the subject while being exposed to the pre-recorded COP. Notice the increase in amplitude as well as frequency content in the signal possibly indicating an increased sway and closed-loop behavior of the postural control system (Fig. 1 left). Figure 1 (right) shows the behavior of the PID controller when it is tracking COP fluctuations recorded during voluntary trunk flexion and extension movements. Notice that this signal has a larger amplitude and higher frequency content than COP from quiet stance. In addition, the error was substantially larger for the COP signal recorded during voluntary trunk oscillations (1.27±2.74 mm, range -9.1 - 9.9 mm). The current COP signal for this condition displays a more complex behavior including both in and out of phase behavior as well as larger amplitudes (Fig. 1 right).

It is concluded that the PID controller implemented on the BALDER balance platform can be used to very accurately reproduce COP fluctuations recorded during quiet stance.

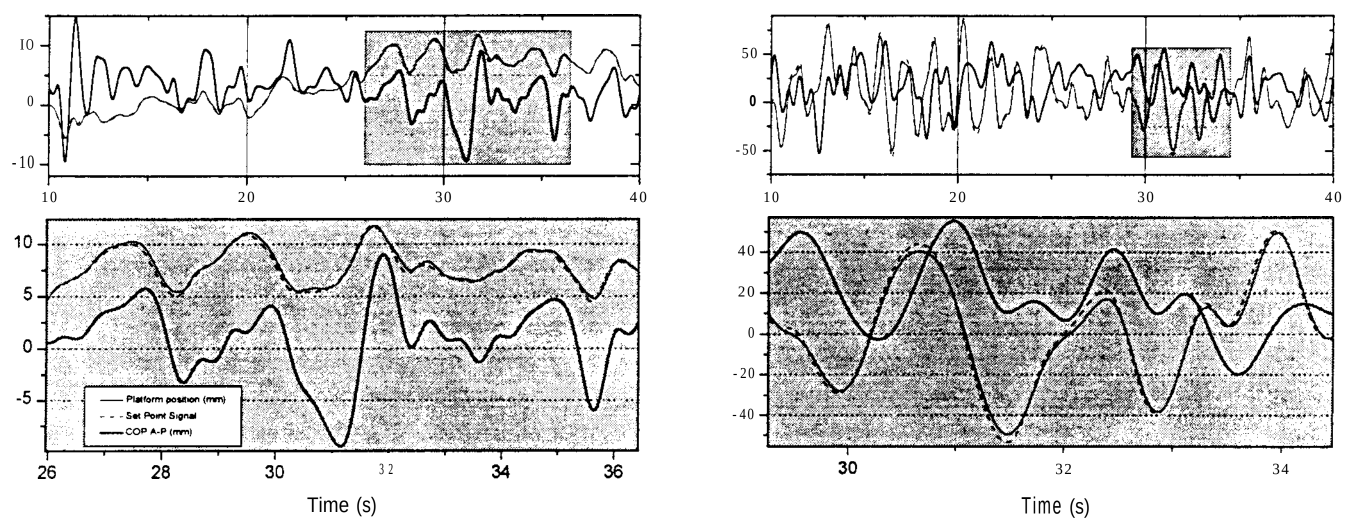

Figure 1. Position of the BALDER platform in relation to the set-point signal consisting of previously recorded COP fluctuations during quiet stance (left) and voluntary flexion-extension movements of the trunk (right, notice difference in scale). The thickest trace represents the current COP for a subject standing on the BALDER platform while being exposed to the pre-recorded COP signals.

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Hemispheric Asymmetry in the Visual Contribution to Postural Control in Healthy Adults

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This study was carried out in order to test the hypothesis of a right hemisphere dominance in the visual control of body balance (Perennou et al., 1997). In this view, we have taken advantage of the anatomical arrangement of the visual pathway, enabling a selective display of visual information to a single hemisphere by masking the ipsilateral visual hemifield in subjects performing a postural task.

Eight healthy adults (age 53 ± 12.1 years) participated in this study. All were pure right-handers, e.g. they performed virtually all skill activities with the right hand. They were asked to maintain a self-regulated, non perturbed dynamic posture while sitting on a seesaw (radius 22 cm), laterally unstable. This dynamic balance task was chosen to increase the visual contribution to body stabilization This task was also selected to be adapted to further studies in disabled people and results presented here will serve as control data for further analysis of brain-damaged patients.

Four visual conditions were analysed : open eyes with vision (OE), closed eyes in darkness (CE), left visual field-right hemisphere (LVF-RH) and right visual field-left hemisphere (RVF-LH). The two last conditions were obtained using hemifield glasses (occlusion of the same hemifield in both eyes).

The kinematics of the body movements were analysed by means of an automatic optical TV image processor (VICON system) at a sampling frequency of 50 frames/s. Two retroflective markers, inducing an horizontal link, were fixed onto the extremities of two protruding sticks fitted on the temporal area, in front of the meatus of the ear, with a view to measuring the successive orientations of the subject's head around the anteroposterior axis. Two other markers were horizontally fixed on the seesaw in the subject frontal plane, with a view to measuring lateral oscillations of the support. The *angular dispersions* (2 standard deviations) of the head ($\sigma^H a$) and support ($\sigma^S a$) in roll were calculated in each trial. An *head anchoring index AI* was used to compare the head stabilization with respect both to external space and to the support :

$$\textit{Head AI} = \frac{(\sigma^H r)^2 - (\sigma^H a)^2}{(\sigma^H r)^2 + (\sigma^H a)^2}$$

where $\sigma^H r$ is 2 standard deviations of the relative angular distribution of the head (with respect to axes linked to the rocking chair). A positive value of this index would indicate a better head stabilization in space than on the support. Three trials by subject were recorded for each visual condition, presented according to a pseudorandom plan.

The data were clear cut and confirmed the prediction of a right hemisphere dominance as regards the visual contribution to the head stabilization in space strategy. The roll *dispersions of the head* were small (about 1°), giving a first indication of an efficient stabilization of the head while sitting on the rocking platform. The best stabilization of the head was obtained in the LVF-RH condition, which differed not only from the conditions CE {F(1,37) = 45.3 ; *p* < O.OOl} and RVF-LH {F(1,37)= 9 ; *p=O.O05}* but also more surprisingly from the full vision situation {F(1,37) = 10.2 ;*p* = 0.003).

The roll *dispersions of the support* ranged roughly between 1.5 and 3.5”. The conditions CE differed from OE {F(1,37) = 37.8 ; *p < 0.00l},* RVF-LH {F(1,37) = 14.4 ; *p < 0.00l}* and LVF-RH {F(1,37) = 21.8 ; *p* < 0.001). The last three conditions (OE, RVF-LH and LVF-RH) did not differ one from the other, however.

The *head AI* was always highly positive and significant, indicating a good efftciency of the head Stabilisation in space strategy (HSSS). A quite comparable HSSS was observed in full vision, RVF-LH and CE, meaning that vision is not crucially needed to achieve this strategy. The HSSS was improved in the condition LVF-RH, however, as compared to OE {F(1,37) = 9,69, *p* = 0.004}, RVF-LH { F(1,37) = 5.8, *p* = 0.02) and CE {F(1,37) = 5.67, p = 0.02). This indicates that the right hemisphere is particularly suited to process the visual input in the view of enhancing the HSSS efficiency in roll.

The right hemisphere in right handers could thus be the dominant hemisphere for the organization of postural control based on an exocentric frame of reference. Surprisingly, however, this right hemisphere dominance seems to be limited to the HSSS, since no difference was found between the respective contributions of both hemispheres at the support level, at least in the sitting posture in healthy subjects. The fact that the improvement of the HSSS efftciency in the case of the LVF-right hemisphere condition was not accompanied by a comparable improvement of the postural performance at the support level may suggest a dominant somestetic contribution to pelvic stability.

Surprisingly, the HSSS was more efficient in the LVF-right hemisphere condition than with full vision. This may suggest that the interaction between both hemispheres in the case of full vision may increase the cost and/or the duration of information processing, and consequently diminish the right hemisphere efftciency with respect to HSSS. An alternative hypothesis may involve the « reciprocal inhibition » theory of attention proposed by Kinsbourne, who claimed that in some tasks, each hemisphere inhibits the opposite hemisphere. But this point is far from clear and will need further investigations.

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Activity of the Gastrocnemius Muscle as a Knee Flexor in Rapid Open Kinetic Chain Movement

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INTRODUCTION

The gastrocnemius (Gc) muscles are well known as a powerful plantar flexor of the ankle joint. Given their significant mass and substantial moment arm about the knee joint, however, littlc attention have been paid to their functional aspect as knee flexors. Even in studies which refer to the knee flexion moment of Gc, expected function around the knee joint is usually explained in relation to the ground contact of the foot. These situations warrant examination of the activities of Gc in the knee flexion free from ground contact.

The purpose of this study was thus to investigate the functional characteristics of Gc muscles as knee flexors in the rapid open kinetic chain extension / flexion of the knee joint.

METHODS

Ten male subjects (176.2±4.9cm, 71.2±10.0kg, 24.8 ±2.7yr) conducted whole range alternate extension / flexion of the knee joint without external load while sitting down on a chair, with the upper body and thighs fixed firmly in upright and horizontal position, respectively. Movement velocities were as follows : 1) 1.5cycles/sec, 2) 2c/s, 3) 2.5c/s and 4) maximal for each subject. One cycle means a single bout of extension / flexion of the knee joint. Surface EMGs were recorded from six leg muscles. i. e. soleus, medial gastrocnemius, tibialis anterior (Ta), vastus medialis, rectus femoris, biceps femoris long head (Bf) and medial hamstrings (Mh). Joint angles were measured with electrogoniometers for the knee and ankle joint. The timing of EMG bursts was measured with reference to the joint motion.

RESULTS

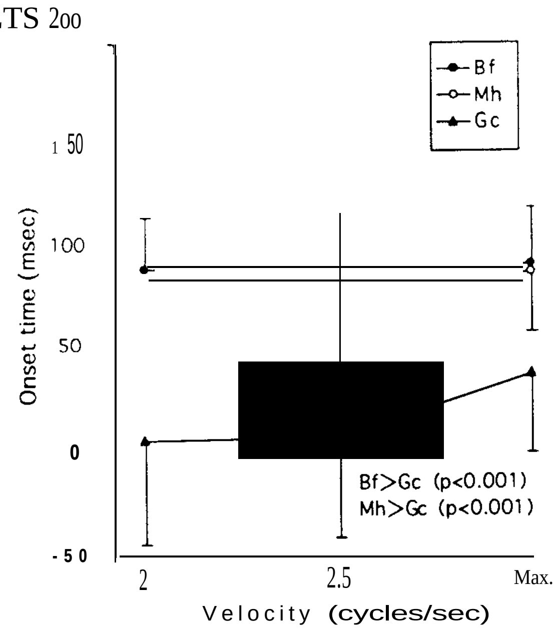
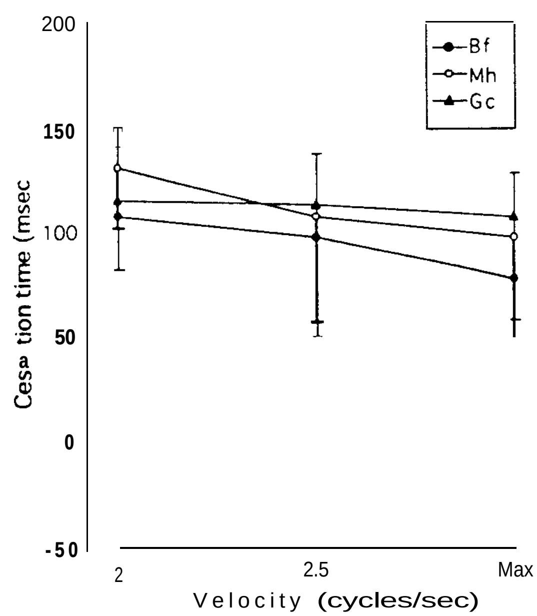


Fig. 1. Mean ard SD of the onset timing of knee flexors: biceps femoris (Bf), medial hamstrings (Mh), medial gastrocnemius (Gc). Onset time is shown as a time interval prior to the switching point from extension to flexion.



Flg.2. Mean and SD of the cessation timing of knee flexors. Cessation time is shown as a time interval following the switching point from extension to flexion.

Hamstrings (Bf, Mh) and Gc showed bursting activities corresponding to each bout of knee flexion. Hamstrings began their activity about 87 msec prior to the onset of knee flexion (Fig 1). While no significant difference was observed in the onset timing within hamstrings, the onset of Gc activities was significantly delayed for about 69 msec relative to that of hamstrings (p<O.001). The cessation timing, on the other hand, was not significantly different between the three knee flexors (Fig 2).

Gc bursts in each bout of knee flexion were accompanied by Ta activities in 9 subjects. These coactivations were more remarkable in higher velocity trials (Fig 3).

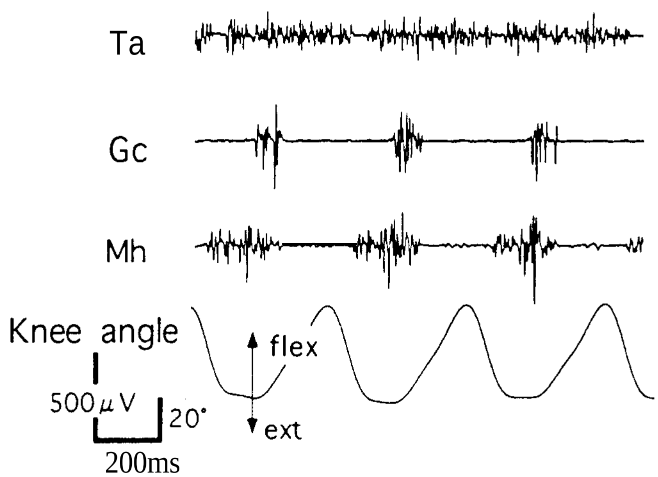


Fig.3. A typical example of EMG and knee joint excursions. See text.

DISCUSSION

The difference in onset timing between hamstrings and Gc suggests that different neural processes and / or neural inflow, such as proprioception of the muscle length, could have affected the attitude of each muscle. On the other hand, the simultaneous cessation of hamstrings and Gc activity possibly came from the reciprocal inhibition exerted by the knee extensors. If this was the case, Gc was controlled as one of knee flexors same as hamstrings, in spite of different timing of the onset of activities. It is plausible that hamstrings have generated effective tension from the end of extension phase. Considering the electromechanical delay, however, Gc is supposed to participate solely in the flexion itself, without contributing to the deceleration of the knee extension. Given the foregoing features, it is speculated that hamstrings mainly generate the impetus for knee flexion and Gc regulates the flexion accommodatingly.

Gc/Ta coactivations appear to contribute to enhance the rapid knee flexion movement. The knee flexion moment of Gc could be reinforced by Ta tension that was transmitted to Gc through the ankle dorsiflexion.

From our results, we tentatively conclude that 1) Gc functions as a knee flexor even in the open kinetic chain movement, 2) Gc has rather regulating property than hamstrings, and 3) rapid knee flexions are effectively carried out by collaboration of Gc and Ta activity as a functional complex.

The Effect of Acupuncture on Correcting Muscle Asymmetry : Dynamic Electromyographic Study

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INTRODUCTION: Although acupuncture is being increasingly utilized as a treatment modality especially for the management of pain in musculo-skeletal disorders (Culliton, 1997): very little is known about the influence of acupuncture on muscle function. The purpose of the present study was to investigate the effect of acupuncture in decreasing functional muscular distortion by evaluating the Dynamic Lumbar Paraspinal (LP) EMG activity before and after acupuncture stimulation.

METHODS:

Subiects: Thirty healthy subjects (17 male and 13 female, subjects), ranging from 18 to 60 years of age (mean age: 32.2 years old, SD: f8.93).

Procedure: Surface EMG electrodes were placed on the subjects' left and right lumbar paraspinal muscles. (Basmajian & Blumenstein, 1989). Subjects were asked to stand straight. and they were instructed to bend forward at the waist to approximately a 45 degree angle and then to slowly return to an upright position. The time duration for each complete movement was approximately 5 seconds and the movement were repeated three times consecutively. The maximum integrated EMG amplitude (absolute EMG) for both sides during the movement was recorded by the computer. The percentage difference (PD) between the left and right absolute EMG was calculated separately for the three movement repetitions. These numbers were then averaged to produce a mean PD (Donaldson. Stanger. Donaldson. Cram. & Skubick. 1993). In accordance with the criteria of the Donaldson & Donaldson (1990) study, more than a 20% difference was considered asymmetrical (asymmetrical subjects-AS), and less than 20% asymmetry was considered symmetrical (symmetrical subjects- SS). SS were asked to rest on the treatment table in prone position for two minutes. After this two minute period. the procedure was repeated in the same manner as the baseline EMG assessment. AS were asked to lie prone on the treatment table and acupuncture needles were inserted in the B23 acupuncture point (between the spinous process of the 2nd and 3rd lumbar vertebrae, approximately 2 finger-widths lateral from the midline) and in the B25 acupuncture point (approximately 2 inches inferior to the B23 point) to a depth of approximately 1 to 1.5 inches. Each point was then stimulated by gently manipulating the needles for 60 seconds,using a twitching movement of the fingers. The post-stimulation EMG assessment was conducted in the same manner as the baseline assessment.

RESULTS:

PD Change in the Symmetrical Subiects: (SS)

A Wilcoxon test could not reject that the SS group's PD had no overall change from the baseline after the subjects lay prone for 2 minutes (N=20, sum of positive ranks=130 from 11 observations, exact p=O. 164).

PD Change in the Asymmetrical Subiects (AS):

A Wilcoxon test of the hypothesis that the AS group's PD decreased was performed, and rejected the hypothesis that the PD did not decrease (N=lO, sum of positive ranks=8.5 from 1 observations, exact p=O.O49). A reduction in EMG asymmetry was observed for 9 out of the 10 asymmetrical subjects.

Absolute EMG value change in the stimulated side:

A Wilcoxon test could not reject that the absolute EMG reading for the stimulated side had no overall change from the baseline (N=lO, sum of positive ranks=25 from 5 observations, exact p=O.846).

Absolute EMG value change in the non-stimulated side:

A Wilcoxon test of whether there was any change in the absolute EMG reading from the baseline was performed for the non-stimulated side. The test could not reject the hypothesis of no change in the non-stimulated side (N=lO, sum of positive ranks=1 9, exact p=O.432).

DISCUSSION: Significant reduction in lumbar Dynamic EMG asymmetry was observed after the acupuncture stimulation. For the absolute EMG values, there was no clear pattern of response observed on the stimulated side after the stimulation. However, on the non-stimulated side, there was a trend of reduction in absolute EMG values when the baseline value for that side was high and an increase when it was low. The change in the absolute EMG values on the non-stimulated side might be either the result of a systemic muscular response through the central nervous system and/or a secondary response compensating for the local response to the acupuncture stimulus applied on the stimulation side. The exact mechanism for the decrease in EMG asymmetry after acupuncture is not certain from this study, however, it suggested that acupuncture stimulation is transmitted through not only a single pathway but through a complex interaction of the central and peripheral systems. This study showed significant decrease of Dynamic EMG activity asymmetry after the application of acupuncture and indicated that acupuncture may be a useful method for decreasing functional muscular distortion and improving synergistic muscle coordination.

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Effects of Fatigue on Movement Coordination in Lifting

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INTRODUCTION

Previous research on lifting and lowering with the load continuously in hands revealed no changes in coordination, which was attributed to the continuous availability of feed-back on the balance between the force producing capacity of the muscles involved and load mass (1). In discrete lifting this balance has to be estimated on the basis of memories from previous lifting movements. We hypothesized therefore, that back muscle fatigue in repetitive discrete lifting would lead to a change in the coordination of the legs and trunk, more specifically an increase in the phase lag between hip and trunk extension. Furthermore, we hypothesized that trunk excursions outside the sagittal plane would increase with fatigue (2), a feature that the 2-dimensional analysis employed in our previous study would have missed.

METHODS

The effects of repetition on the kinematics in discrete lifting (free technique) were studied in 10 healthy male subjects. Subjects lifted a barbell of 10% body mass at a determined speed and along a determined trajectory for 630 times during about 40 minutes. 3-D kinematics of the feet, lower and upper legs, pelvis, and trunk were recorded in the first three and the final three lifting movements of each bout of 70 lifts (3). The effect of repetition on selected kinematic variables averaged over the first 250 ms of the movement was studied by means of repeated measures ANOVA and linear regression analysis.

RESULTS

Within-subject variation in joint kinematics between the three consecutive lifts appeared very low in all recordings when compared to between-subject variance. Over time trunk extension velocity in the initial 250 ms of the lifting movement decreased (-0.5 $deg \cdot s^{-1}$ per bout, p<O.OOl), to reach negative (increasing flexion) values in most subjects (figure 1). In contrast, hip extension velocity increased (left and right hip: 1.1 and 1.0 $deg \cdot s^{-1}$ per bout, p<O.OOl, figure 1). This resulted in the expected change in relative phase angle between extension of the hips and the lumbosacral joint (both left and right: 0.01 degree per bout, p<O.OOl).

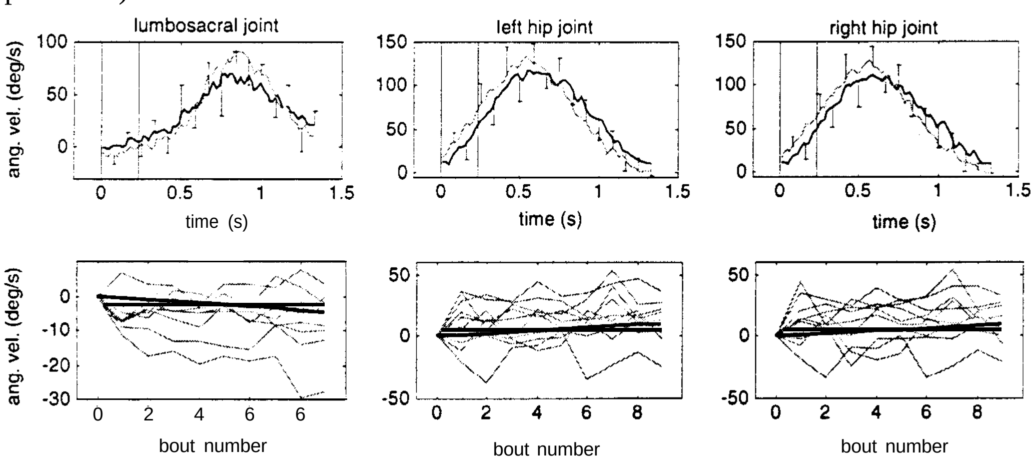

Figure 1. top: Angular velocities in the initial lifting cycles (solid line) compared to those in the final lifting bout (dotted line) averaged across subjects (n=lO). Vertical bars indicate one standard deviation. bottom: The change in mean angular velocities during the first 250 ms of the lifting movement per subject plotted against the number of the lifting bout. The thick line indicates the regression line based on the pooled data.

Also over time subjects started the lifting movement with their legs more extended (left and right ankle: 0.5 and 0.6 degrees per bout, both p<O.OOl; right knee: 1.3 degrees per bout, p<O.OOl; left and right hip: 0.6 degrees per bout, both p<O.OOl) and the trunk further flexed (-0.7 degrees per bout , p<O.OOl). Finally, the motion of the trunk around its longitudinal axis (twisting) increased (0.5 degrees per bout p<O.OOl, figure 2) .

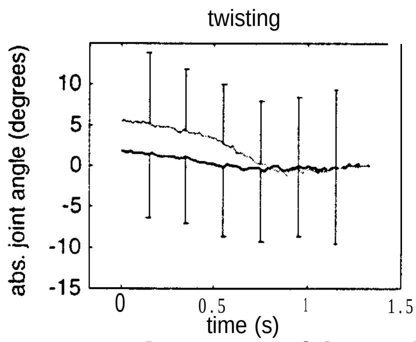

Figure 2. The motion of the trunk around its longitudinal axis averaged across subjects (n=lO) in the first (black line) and final (grey line) lifting bout. The vertical bars indicate one standard deviation.

DISCUSSION

The discrete nature of the lifting task studied requires an anticipatory scaling of back muscle activity prior to load pick-up (4). The decreasing trunk extension velocity and the resulting increase in phase lag between hip and trunk extension is, therefore, interpreted as a consequence of inadequate scaling of back .muscle activity to the load to be picked up as a result of back muscle fatigue. More specifically a decreased rate of force development of these muscles (2), not sufficiently counteracted by earlier activation explain these results. In continuous lifting this fatigue effect may be counteractedat a peripheral level by the ongoing feedback on balance between muscular capacity and load mass, for instance through type III and IV muscle afferents projecting on the gamma-system (5). The change in intial lifting posture found in the present could be an adaptation to retard further fatigue development, since it implies a shift from the leg lifting technique to the energetically more efficient back lifting technique (6).

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Influence of Fatigue on Sympatho-Vagal Balance during Skiing Exercise

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INTRODUCTION

In the appropriate physical exercise, the relationship between willpower and fatigue should be balanced to avoid overwork. The feeling of fatigue is a reaction of the central nervous system caused by a variety of fatigue related factors through metabolic and cardiorespiratory responses. On the other hand, muscular fatigue, which is local or peripheral fatigue, can be objectively evaluated by lactate accumulation, muscle oxygen consumption, and myoelectric (ME) signals. The information on the accumulation of metabolic byproducts is transmitted to the central nervous system via the cardiovascular system. That is, muscular fatigue definitely influences on the autonomic nervous activities.

The heart rate variability (HRV) is dynamically controlled under the autonomic nervous system and the frequency components are considered as the indices of the sympatho-vagal balance. The HR time-series is substantially nonstationary and is modulated by many factors with different time-varying scales. During exercise, the parasympathetic tone decreases, but the behavior of the sympathetic tone has not been completely clarified. We applied the wavelet analysis to reveal the detailed features of HRV [l] and compared the results with muscle activity during skiing exercise.

METHODS

Two male subjects participated in our experiment on separate days after they were informed of the experimental procedures and risks associated with fatiguing efforts. About thirteen trials per day were performed for around seven hours including one hour lunch break. A trail of skiing exercise consisted of about 2 minutes of skiing, 10 minutes of ski lift riding, and preparation for measurement. Two minutes of HRV was analyzed for the frequency ranged from 0.01 Hz to 1.25 Hz. The Gabor function was selected as a mother wavelet. We tracked the dominant frequency components within the range from 0.03 Hz to 0.17 Hz for sympathetic and parasympathetic activities and within the range from 0.3 Hz to 0.7 Hz for the parasympathetic activity in the time-frequency space. The time-series of a low range peak frequency and a high range peak frequency components were evaluated in each interval. On the other hand, muscular activity was estimated by the conventional amplitude and frequency indices of surface ME signals during skiing, and also evaluated by electrically evoked potentials (superimposed M waves) [2] at specific times of the day. Then we compared the HRV with muscle activities in relation to both fatigue and intensive efforts.

RESULTS

In each trial, the high and low frequency components of HRV increased during a ski lift riding phase and decreased during a skiing phase. The periodic change became prominent near the final trial of the day. Although muscular fatigue was not clearly confirmed by surface myoelectric signals, the time-varying behavior of the low frequency component seemed to correlate with muscle activities as fatigue progressed (Fig. 1). The discontinuity of frequency components was observed around the parasympathetic tone related frequency range. Even around the sympathetic tone related frequency range, we found a few discontinuity times in HRV. The muscular fatigue assessment using the superimposed M waves of 40-sec did not show the apparent difference. Only an instantaneous frequency of superimposed M waves [2] at the final trial decreased more than at the beginning. Something different might happen during skiing exercise, because the subject showed a typical change of superimposed M waves during lOO-sec 70% MVC that usually appears in the basic experiment .

DISCUSSION

It has not been clarified how much feeling of fatigue and/or muscular fatigue would affect HRV and ME signals during long-term periodical skiing exercise. In our results, the fatigue related features appeared in HRV became prominent as fatigue progressed. Among many physiological factors, the increase of metabolic byproducts probably augmented the periodic change of the frequency components in HRV. In order to confirm the relation between time-varying behavior of frequency components in HRV and muscle activities, multivariate analysis of several factors [3] would be needed.

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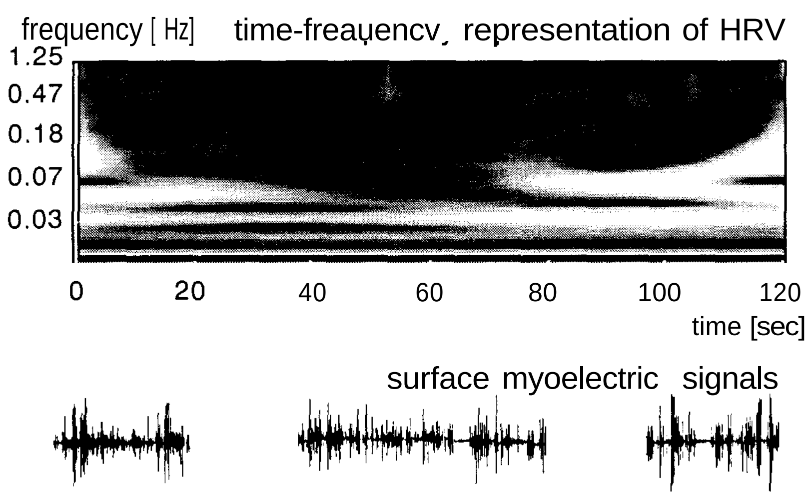

Fig. 1. Time-frequency representation of HRV using Wavelet analysis and ME signals during each segment.

Low-frequency and Central Fatigue after Eccentric and Concentric Exercise to a Standardized Torque Decrement

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INTRODUCTION: Muscle fatigue's task-dependent nature is well-documented in the literature.' For instance, repetitive eccentric volitional exercise is thought to selectively induce states of low-frequency fatigue when compared to concentric exercise.[2, 3, 4] However, in the absence of standardized parameters between modes of exercise, it is difficult to interpret whether the observed fatigue response is due to differences in velocity/load/torque-impairment level, or due to characteristics inherent to the mode of exercise. The interpretation of a fatigue response would also be enhanced by examining measures of both peripheral and central fatigue during the same experimental state. The purpose of this study was to understand if repetitive eccentric (ECC) or concentric (CON) volitional exercise selectively influences the type and magnitude of fatigue induced when speed, load, and torque-impairment parameters are standardized between exercise modes. **METHODS:** Sixteen male subjects (mean: 27.9 years) exercised a 40 degree arc of their triceps brachii (work/rest cycle: 3.1 s / 5.7s; velocity: 1 $3°sec^{-1}/7°sec^{-1}$) under eccentric and concentric modes on a computer-controlled isokinetic dynamometer. All subjects were exposed to both exercise modes on separate sessions at least one week apart, in a balanced repeated measures design. Each session's workload was set at 75% of the individual subject's respective MVC. For both protocols, exercise stopped when a 25% reduction in eccentric MVC torque was achieved. Before, immediately after, and 10 minutes after each exercise session, measurements of torque, triceps medialis surface EMG, low-frequency fatigue (LFF) and central fatigue were assessed during isometric maximal efforts. The resting peak torque ratio from high/low frequency twitches assessed LFF; increases in the post-exercise ratio indicated the presence of LFF. The measurement resolution for the entire system equated to 1.2% of the mean resting high frequency pulse twitch torque (about 0.25 Nm). Central fatigue was calculated with a voluntary activation score equation: (l- interpolated twitch torque / resting twitch force) X 100. A low- then high-frequency interpolated twitch pulse was superimposed during the same isometric MVC to assess the influence of pulse frequency on measuring central fatigue. **RESULTS:** ICCs were good for within- (>0.96) and between-day (>0.78) pre-fatigue MVC torque for all three contraction types. Significant differences in pre-fatigue MVC torque existed between all three contraction types for the chosen exercise velocity (p>.OOl). Both exercise protocols successfully achieved the target stopping criteria. Peak torque for ECC, CON, and isometric MVCs was significantly reduced compared to their respective pre-exercise value; yet, comparisons between repetitive exercise mode revealed no significant differences for either MVC, at any time. The patterns of response for mean EMG amplitude, median power frequency, and neural drive efficiency also showed no significant difference based on how fatigue was induced for any contraction type, at any time. The measures of LFF and central fatigue did not show significant differences based upon whether repetitive eccentric or concentric exercise was employed. The resting twitch torque ratio did not significantly increase to indicate the presence of LFF until after 10 minutes of recovery. The sample size provided 96.8% power in detecting differences in the selected measure of LFF. Central fatigue was also not significantly present until the recovery period, as measured by both the low and high frequency twitch interpolation techniques. At each pre, post, and recovery time, there was no significant difference between the low vs. high frequency voluntary activation scores. Only immediately after exercise was there a significant correlation between the voluntary activation score (central fatigue) and the resting twitch torque ratio (LFF) ($r > -.50$; $p<.005$). DISCUSSION: Under these conditions, LFF and central fatigue are not selective to a particular mode of exercise. This contradicts the notion that LFF is selectively induced during repetitive eccentric voluntary exercise. As a result, the level of torque-producing impairment induced during exercise should be considered in the future design/interpretation of experiments that compare the mechanisms of fatigue between difference exercise modes. In a broad sense, the study's results suggest that the concept of muscle fatigue's task-dependency may deserve further refinement to better delineate those task parameters that truly influence the response specificity observed under eccentric, concentric, and isometric repetitive contractions. The relationship between measures of LFF and central fatigue requires further investigation.

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Cocontraction of Trunk Muscles: Influence of Speed of Motion and Fatigue

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INTRODUCTION

Low back pain (LBP) has a high prevalence in the population. For instance, in the Province of Quebec, it has been experienced by 1.37% of the workers in 1981 and this resulted in an estimate of over 2 millions workdays lost(l). In spite of its high financial cost, very little is known about LBP. Diagnostic of the pathology is still difficult even if many studies have been devoted to the function of the spine in healthy subjects, In some studies, the intra-abdominal pressure is considered to be an important mechanism for stabilizing the spine during dynamic movements (2). In others, activation of trunk muscles have been analyzed during various movements and postures: for instances, Floyd and Sylver (3) found an angle at which the back muscles are not activated when performing a static flexion. Studies on trunk muscles activation have been performed mostly with the subject attached to a rigid structure which allowed only a few degrees of freedom to the movements. With such constraints, the motions performed can be quite different from those performed daily by everyone. To analyze the natural motion of the trunk, we examined the influence of fatigue and speed of movements on the cocontraction of trunk muscles during flexion/extensions (F/E) performed freely in space. Fatigue and speed of motion are also important in ergonomics.

METHODS

Ten subjects participated in the study. The EMG signals were obtained from 12 bipolar active surface electrodes. Two of the electrodes were placed on the abdomen, two others on the obliques and a fifth one over the heart to remove the electrocardiographic (ECG) artifacts from the abdominals signals. On the back, six electrodes were placed at three different levels (3 on each side) and a seventh one was placed between the upper pair to evaluate crosstalk. Kinematic data was obtained through triads of reflecting markers placed over the spinous process of vertebra T2 and over S 1. During a time span of 12 s, the subjects were asked to performed continuous F/E at periods of 3, 2.25 and 1.5 s. Each series was repeated 4 times. A fatigue task consisting of 30 F/E at a period of 1.5 s and series of F/E performed at a period which appeared natural to each subject were also realized. Each non-fatiguing series was separated by a rest of 1 min while the rest period was 5 min between each fatiguing one. Additional information on the protocol can be obtained from (4). Once the ECG contamination in the abdominals' signals was removed, all EMG were digitally filtered with a two-way 15 Hz second order lowpass filter. Then, the DC offset of this signal was removed to reduce the influence of noise. For each F/E the cocontraction between two signals (xi and yi) was given by:

$$\sum_i x_i y_i \Big/ \sqrt{\sum_i x_i x_i \sum_i y_i y_i} \qquad (1)$$

To evaluate cocontraction between muscle groups the EMG of each muscle was normalized with its average amplitude obtained in the 1.5 s task without fatigue for each F/E. Within a group, the normalized EMG were then averaged.

RESULTS

While the free period of motion was found to vary between 1.47-3.05 s (4), all others movements were made within 7% of the desired period (except for one subject who had trouble performing the fatigue task). No relationship was observed between the variation in the F/E amplitude and the motion period. When the abdominal and back muscles are considered as two groups, cocontraction ranged from 0.19 to

0.74 for all subjects. Averages obtained from individual values where not statistically correlated (a=0.05) with the period of the F/E (Fig 1). However, when data obtained at the free period were compared with those obtained at 3, 2.25, 1.5 and 1.5s with fatigue, coefficients of correlation of 0.88, 0.72, 0.39 and 0.45 where obtained. Subjects were asked, and succeeded to performed their movements in the sagittal plane without lateral bending. When the cocontraction between left and right muscles of the abdomen and of the back was evaluated, values of 0.84 and 0.92 were respectively obtained (average for the ten subjects). As shown in Fig.1, this level of cocontraction was not influenced by the speed of F/E or fatigue. When cocontraction between signals obtained at each of the three levels of back muscles was analyzed, values between 0.90 and 0.96 were obtained for the ten subjects as a group. As shown in Fig..2, these values were not affected by fatigue or movement speed. To detect if crosstalk was influencing our cocontraction measures, an electrode was placed between the upper pair of left-right electrodes in the back. With an inter-electrodes distance of 5 cm, only minimal crosstalk was detected and practically none was detected between more distant electrodes (i.e. 7.5 cm). In the fatigue task, an increase in EMG amplitude was observed in 9 of the 10 subjects and was taken as an indication of fatigue. For these subjects no effect of fatigue on the cocontraction levels was detected.

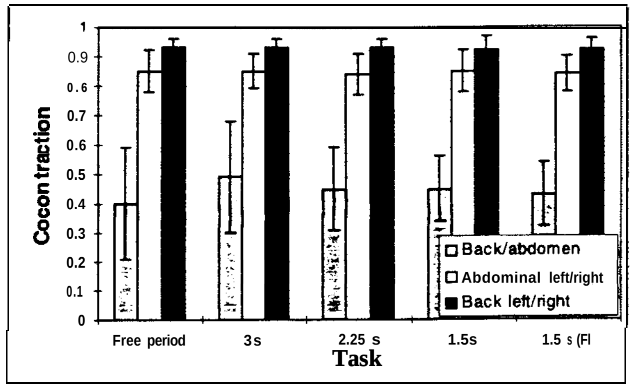
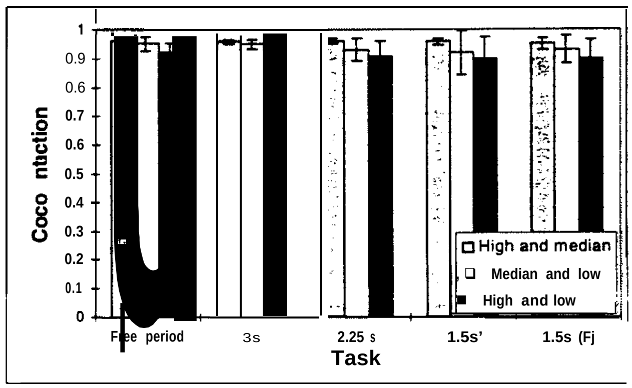


Figure 1. Cocontraction levels measured for five tasks between three different pairs of muscle groups. The error bars are standard deviatinns.



F igure 2. Cocontraction levels for the five tasks between three different levels of the back. The error bars are standard deviations.

DISCUSSION

During the free period movements, even if the period varied appreciably from subject to subject, it appears that cocontraction values obtained between the back and abdomen muscles as a group can be indicative of the level of cocontraction to be expected during movements at fixed period of 3 and 2.25 s. The high cocontraction level observed between those two antagonists muscles groups could cause an increase in the intra-abdominal pressure which would then increase the trunk's rigidity. This mechanism may be used to get a better control on the motion period: during movements at fixed periods, cocontraction was slightly higher than in the free period movements. For the back, the high cocontraction between left/right signals is a good indicator that F/E could be done in the sagittal plane without external constraints. The smaller cocontraction observed for the abdominal muscles was associated with their low signal to noise ratio. For normal subjects performing F/E, is seems that the erector Spinae muscles could be considered as a single functional unit. With fatigue, the observed lack of changes in cocontraction could come from an insufficient level of fatigue induced by our protocol.

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The Effect of Video Display Terminal (VDT) Mouse Use on Muscle Contractions in the Neck and Forearm

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In recent years, video display terminal (VDT) keyboard and mouse use have been increasingly associated with CTD. Four studies (Attwood, 1989; Franzblau, Flaschner, Albers, Blitz, Werner, & Armstrong, 1993; Hagberg, 1995; Karlqvist, Hagberg, & Selin, 1994) suggested that VDT mouse use may be a greater risk for CTD development than keyboard use. At this time, no studies have been published using electromyography (EMG) to evaluate muscle activity during VDT mouse use. This study used surface electromyography (EMG) to study, the amplitude of muscle contraction during VDT mouse use. Two questions were examined: What percent maximum voluntary contraction (%MVC) is utilized by the upper trapezius, the wrist/finger flexors and the wrist/finger extensors during mouse use? How does the change of location of the mouse from next to the computer to a lapdesk, influence muscle contractions?

Method: Thirty right handed participants, 11 males and 19 females, were recruited from a large urban private university. Participants' duration of mouse use ranged from six months to 13 years (mean 5.22 years) and average daily use ranged lrom 10 minutes to 10 hours (mean 2.33 hours). Participants were randomly assigned to sequence AB (mouse located next to the computer then on lapdesk) or BA (mouse located on lapdesk and then at the computer) Three pairs of surface Beckman miniature silver/silver chloride electrodes were attached to the skin with the centers of the electrodes 1.5 cm apart over the right: 1) descending parts of the upper trapezius muscle; 2) wrist/finger flexors; 3) wrist/finger extensors. The EMG signals were amplified using a Grass model 7D polygraph (Common mode rejection ratio at the pre-amplifier and amplifier 16OO:l; signal to noise ratio 280: 14; Input impedance 44MΩ differential; half amplitude frequency range 10 Hz to 20,000 Hz). The EMG channels were calibrated to an average of 2.36 microvolts peak to peak. Source. impedance was ≤ 1 0KΩ for all participants. Data were digitally converted using the WATSCOPE™. A pressure sensitive switch was applied to a Recoton™ mouse and connected to the WATSCOPE™ to act as a D/C event marker.

Prior to electrode application, the work station was adjusted according to the participants' reports of comfort. After a practice session of approximately five minutes, the participants moved the screen cursor to one of five 3rnrn circles in a prescribed pattern; center to upper left comer (Quadrant l), center to upper right comer (Quadrant 2), center to lower left comer (Quadrant 3), center to lower right comer (Quadrant 4). At each circle, the participants pressed the mouse button. Three successful trials at each location were recorded.

Data Analysis: EMG signals were demeaned, digitally filtered at 100 Hz. (resulting frequency range was lo- 100 Hz.), rectified, and cumulatively integrated using ANAPAC™. Results are reported as a percent of highest of three maximum voluntary contraction (%MVC) The MVC for each muscle group was calculated over a two and a half second period in each of the three five second maximally resisted efforts for that muscle. In all cases second two was the starting point.

Despite expectations, a one factor ANOVA examining quadrant indicated a significant difference among the quadrants for all three muscle groups for both locations except for the trapezius in the computer location. Since quadrants were significantly different, and as Quadrant 2 had the highest average %MVC, a two-factor (sequence by order) ANOVA with one repeated measure, order, and one between-factor, sequence, was performed on that Quadrant to identify treatment effect.

Results: Means were generally small for all three muscle groups (lapdesk: trapezius M 4.71%, wrist/finger flexors M 3.53%; wrist/linger extensors M 6.26%; computer: trapezius M 5.74%, wrist/finger flexors M 3.94%; wrist/finger extensors M 6.03%). However, individual %MVC ranged from 0.96 to 21.60% when the mouse was used next to the computer and 0.86 to 18.87% with it on the lapdesk, with the trapezius showing the greatest variation. This suggests that VDT mouse use may be a risk factor for CTD for certain individuals. Comparison of the %MVC medians for mouse use with those obtained in keyboard studies (Bendix & Jessen, 1986; Fernstrom et al., 1994; Hagberg & Sundelin, 1986) suggests that the %MVC are comparable for the trapezius, lower for the extensor muscle group, and greater for the flexor muscle group during mouse use than during keyboard use (Table 1).

Results for the trapezius muscle and flexor muscle groups for Quadrant 2 were quite similar. There were no significant sequence effects (trapezius, $F_{1,28}$= 0.28, p=.58; flexor, $F_{1,28}$=0.43, p=.52); however, there were significant order effects (trapezius, $F_{1,28}$= 6.02, p=.O2; flexor, $F_{1,28}$=8.40, p=.OO7). This result suggests participants performed significantly less efficiently during the second condition, regardless of the sequence of the conditions. There were no significant treatment effects (trapezius, $F_{1,28}$= 0.7 1, p=.40; flexor, $F_{1,28}$=0.33, p=.57) after variances associated with sequence and order were removed. The trapezius showed a strong treatment effect (r =.64) while the flexor muscle group showed a moderate effect (r = .49). Further examination of the means and standard deviations between the locations suggests that although the change in location did not cause a significant difference, there was a trend towards reduced muscle contractions for the lapdesk location.

For the extensor muscle, there were no significant sequence effects ($F_{1,28}$=2.01, p=. 17), order effects ($F_{1,28}$= 1.07, p=.3 l), or treatment effects ($F_{1,28}$=0.22, p=.64)(Table 3). Although a moderate effect size was observed (r=.42) further examination of the data did not suggest any trends.

Table 1 - Comparison of results to VDT keyboard studies

Muscle groups (range of median %MVC)	Bendix & Jessen (1986)	Fernstrom et al. (1994)	Hagberg & Sundelin (1986)	Baker et al. (1997) (computer)	Baker et al. (1997) (lapdesk)
Trapezius	4.7-10.8	4.1-5.8	3.5 (3 hrs);7.3 (5 hrs)	4.80	3.46
Flexors	__	1.3-2.1	__	3.13	2.80
Extensors	8.3-9.4	6.5-7.8	---	5.58	5.56

Notes: Each study reported a range of medians, as they tested many different pieces of equipment. Bendix & Jessen used electric typewriters with and without wrist support. Fernström et al. used several different keyboards. Hagberg & Sundelin tested keyboards over a period of three hours and five hours respectively.

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Progress of Muscle Fatigue Assessed by Using Twitch Interpolation Technique

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INTRODUCTION

Surface electromyography has been used extensively as a measure to evaluate muscle fatigue noninvasively. It has been well known that the power spectrum shifts towards lower frequencies and IEMG increases with progression of muscle fatigue”. Although the reduction in the muscle fiber conduction velocity has been reported to explain the frequency shift as a peripheral factor, involvement of central mechanism including motor unit recruitment cannot be excluded.

Recently, evaluation of muscle fatigue using voluntary and evoked contractions simultaneously i.e. twitch interpolation technique[2)] has been proposed, which may provide possibilities of discrimination between central and peripheral factors in muscle fatigue[3)4)]. The purpose of this study is to examine the mechanism of muscle fatigue progression using twitch interpolation technique, and to identify the endurance capacity in individual subjects based on the pattern of muscle fatigue progression as depending on the contraction task concerned.

METHODS

Force and EMG measurements

Force and EMG measurements were made by using an isokinetic training machine (Cybex II). Subjects exerted isometric knee extensions of 60% MVC (short duration fatigue task; SDF task) and 20% MVC (long duration fatigue task; LDF task) until the target force could no longer be maintained. Simultaneously with the voluntary contractions, the quadriceps femoris muscles were stimulated percutaneously through two 10X 1Ocm dampened electrodes bandaged to the anterolateral thigh”. Square-wave pulses of 200-300V with 0. lmsec duration were used. A train of stimuli at 1OOHz was delivered for 320rns at intervals of 5sec. The voluntary isometric forces (VF) and superimposed tetanic forces (STF) were measured at the ankle with strain gauge transducer. Using a four-bar active electrode, bipolar surface EMGs were obtained from the vastus lateralis muscle (VL) during the isometric voluntary contractions.

Data analysis

The myoelectric signals were A/D converted with 5kHz sampling frequency, and mean power frequency (MPF) and integrated EMG (IEMG) were calculated. The magnitude of the mean force generated by voluntary effort was compared with the peak force generated by electric stimulation superimposed on a voluntary contraction; the peak/mean force ratio (PMFR) indicating a reserve of voluntary activation was calculated.

RESULTS AND DISCUSSION

Task dependency in the course of muscle fatigue progression

Endurance times were 50-70 s for SDF task, and 350-450 s for LDF task. During the enduring contraction, surface EMG showed a tendency to decrease in MPF and to increase in IEMG in all subjects, indicating development of localized fatigue in the muscle. With progression of the fatigue, the PMFR showed a tendency to decrease, which was more obvious in LDF task than in SDF task. Such a difference between tasks suggest that development of fatigue in intensive contractions as SDF task are mainly dependent upon the status of muscle contractile apparatus, whereas occurrence of fatigue in moderate contractions as LDF task are dependent upon the impairment of muscle fiber and other central factors .

Individual variation in fatiguability

Individual variation in fatiguability was seen in LDF task; subjects characterized with a greater change in MPF and IEMG also showed a greater change in PMFR (Fig. lA), whereas subjects with little change in the EMG parameters also showed little change in PMFR (Fig. 1B). Besides, endurance time was longer in the former than in the latter. These differences are considered to be caused by the difference among the subjects in the relative contribution of central and peripheral factors in the time course of muscle fatigue progression. It is inferred that the different time course determines endurance time which reflect the endurance capacity in individual subjects.

Our results suggest the existence of task- and individual dependency in the course of muscle fatigue progression, and the applicability of twitch interpolation technique to the systematic evaluation of muscle fatigue.

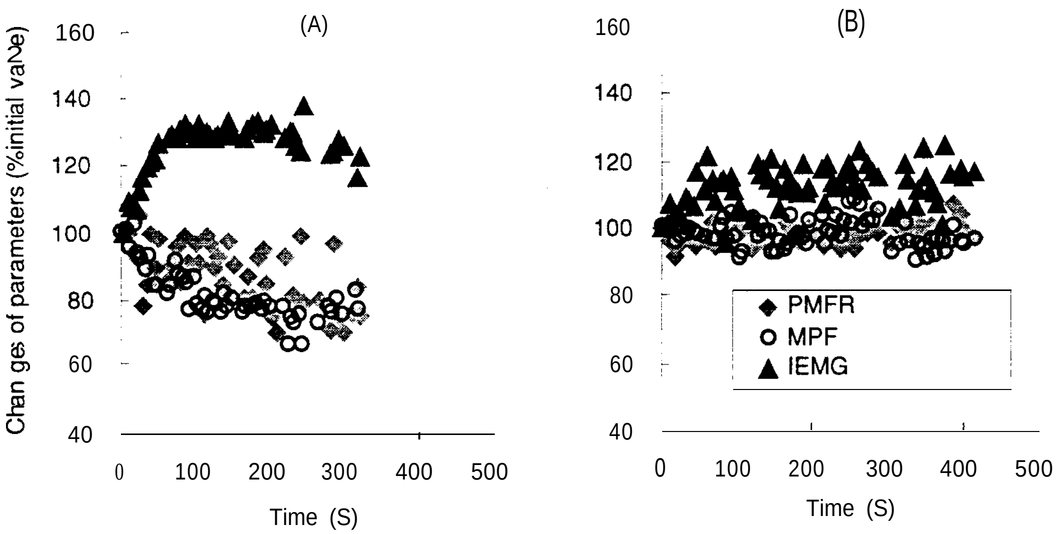

Fig. 1 Typical changes of mean power frequency, integrated EMG and peak/mean force ratio with progression of muscle fatigue in LD F task. Values are expressed as a percentage of unfatigued initial value.

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Estimation of Recruitment by Counting the Number of Active Motor Units Seen by an Intramuscular Electrode

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Introduction. Quantification of motor-unit recruitment is of interest in both kinesiological and clinical electromyography. This paper considers whether recruitment can be estimated by counting the number of active motor units (MUs) recorded by an intramuscular electrode. Two important issues are addressed. First, since the electrode may record motor unit action potentials (MUAPs) with a wide range of amplitudes, what is an appropriate criterion for deciding which MUAPs to count? Second, is such counting likely to produce a reliable estimate?

Methods. We analyzed 68 EMG signals recorded in brachial biceps and tibialis anterior muscles in three normal subjects during isometric contractions at 510% and 30% MVC. The signals were recorded using a monopolar needle electrode (TECA MF25), with a surface reference electrode placed on the skin near the insertion site. The signals were sampled at 10 kHz and digitally filtered into two frequency bands (Bl: 1.2-2.5 kHz, B2: 0.6-1.2 kHz, see Fig. 1) to convert the MUAPs into spikes. The BI signal was decomposed into its constituent MUAP trains using an interactive decomposition program developed by us. Care was taken to identify as many MUAP trains as possible, using the B2 signal to help distinguish small B 1 spikes. In all, 560 MUs were identified. For each identified MU, the MUAP and spike templates were averaged, and their -peak-to-peak amplitudes were computed. Then, the identified units were subtracted from the raw, B 1, and B2 signals to yield residual signals. The maximum peak-to-peak amplitude of the remaining MUAPs (i.e., those which had not been identified) was estimated from the peak-to-peak amplitude of the residual raw signal.

Results. Figure 1 shows a typical EMG signal, the corresponding Bl and B2 signals, and the residual signals. 13 MUs were identified, ranging in size from 340 to 1870 uV (B 1: 19 to 712, B2: 46 to 459). Additional smaller spikes could be detected in the residual Bl and B2 signals, but they were too close to the noise level to be accurately identified. The residual raw signal had a peak-to-peak amplitude of 380 uV, but no distinguishable

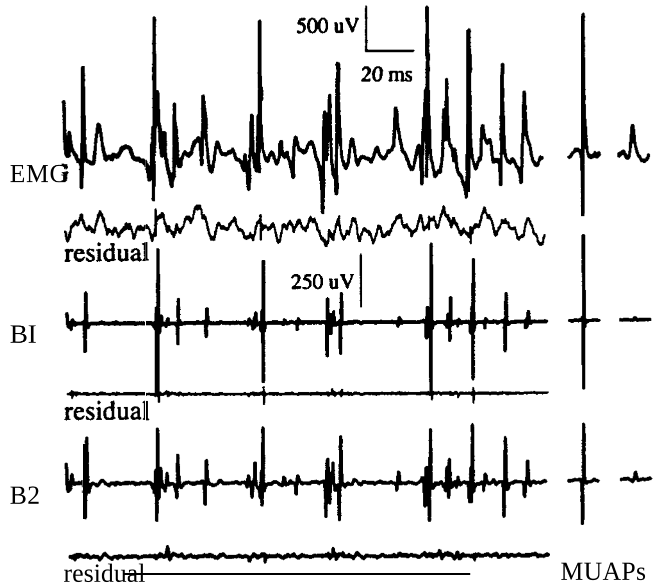

Figure 1. EMG signal, signals obtained by filtering (Bl: 1.2-2.5 kHz, B2: 0.6-1.2 kHz), and residuals obtained by subtracting out the identified units. The largest and smallest of the identified MUAPs and their Bl and B2 spikes are shown on the right. Note that the visible spikes in the Bl residual are due to imperfect subtraction because of variability in the shapes of the identified units.

MUAPs could be detected.

A scatterplot of MUAP and B 1 spike amplitudes from all three subjects is shown in Figure 2. The amplitudes of the MUAPs and spikes were strongly correlated. There were no significant differences between subjects or between muscles. However, the mean amplitudes were higher at 30% MVC than at 110% MVC (1200 vs 800 uV for MUAPs, 210 vs 140 uV for spikes), because at 30% MVC it was not possible to reliably identify as many smaller spikes. The amplitude distributions at 510% MVC are shown in Figure 3. The amplitude of the residual raw signals ranged from 1.0 to 2.8 (mean 1.6) times the amplitude of the smallest identified MUAP.

Discussion. A monopolar needle electrode sees a continuum of MUAP amplitudes, extending presumably to very small MUAPs from distant MUs. In this study, the amplitudes of the residual signals at 110% MVC ranged between 200 and 500 uV. Therefore we can be fairly sure that most of

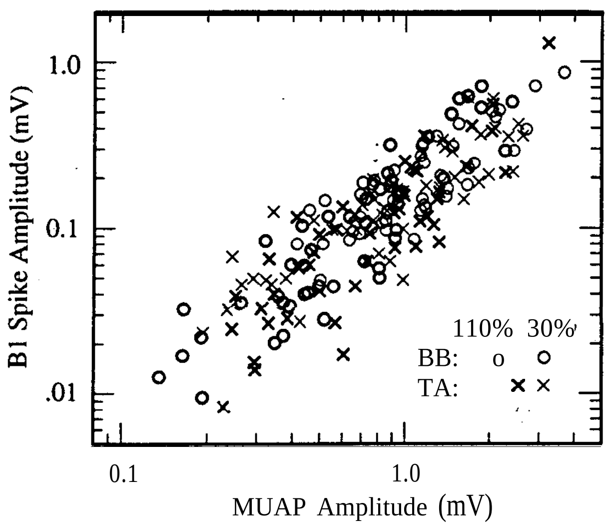

Figure 2. Scatterplot of MUAP amplitude vs Bl spike amplitude for the identified MUAPs for all subjects. For clarity, only a representative fraction of the data is shown.

the MUAPs with amplitudes greater than 500 uV were identified, but only a fraction of those between 200 and 500 uV probably were identified. These smaller MUAPs, because of their broad shapes, blend indistinguishably into, and indeed help to constitute, the general background noise upon which the MUAPs of the nearby MUs are superimposed. For this reason, it is difficult to quatitate the number of MUs that contribute to the raw EMG signal.

The spikes in the high-pass-filtered (Bl) signal stood out much more clearly than the MUAPs in the raw signal. From inspection of the residual signals, we are confident that every Bl spike at IlO%.MVC with an amplitude of at least 50 uV was identified, as were most of those with amplitudes above 25 uV. Therefore Figure 3B accurately portrays the distribution of Bl spike amplitudes seen by a monopolar needle electrode.

The spikes in high-pass filtered EMG signals are thought to result primarily from one (or at most a few) muscle fibers closest to the electrode.' Let r denote distance from the electrode. Then the number, ~1, of muscle fibers on the circumference of a circle of radius r is proportional to r, and the amplitude, *A*, of the single muscle fiber action potential is proportional to $1/r^3$.[3] For distances fairly close to the electrode, the number of MUs whose closest fiber is at distance r from the electrode will be proportional to n. Then the distribution of spike amplitudes, i.e., the likelihood that a randomly selected MUAP has spike amplitude *A*, is given by:

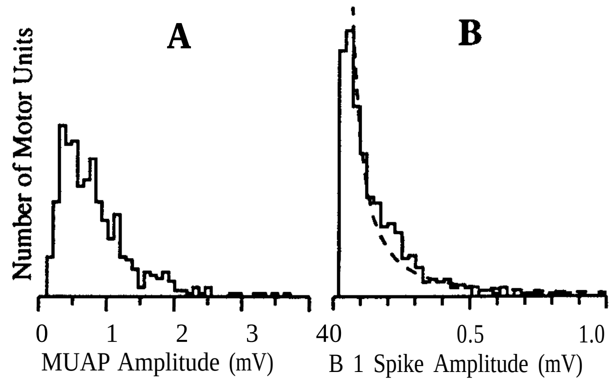

Figure 3. Distribution of MUAP amplitudes (A) and Bl spike amplitudes (B) of all the identified units from the 110% MVC contractions. The dashed line plots $cA^{-5/3}$, where c was chosen to best fit the data.,

$$p(A) \sim n(A) / (dA/dr) \sim A^{-5/3}.$$

The empirical distribution fits this function fairly well (dotted line in Figure 3B). Thus one can think of the Bl spike amplitude as being inversely related to the distance of the MU (i.e., the distance of its closest fiber) from the electrode.

These results suggest that recruitment can be estimated by high-pass filtering the EMG signal and counting how many distinct MUs have spikes which exceed- a particular amplitude threshold. This is an objective criterion, equivalent to counting the number of active MUs which have at least one fiber within a fixed radius from the electrode. In this study, the distinct spikes were enumerated using decomposition. The number of distinct spikes can also be estimated by a statistical analysis of the times of occurrence of the detected spikes.[2]

Although, in this study there were significant differences in the spike amplitude distributions from different sites in the same muscle, the overall distributions from each subject and each muscle were similar. This implies that averaging the number of active MUs seen at several sites may yield an estimate of recruitment that will allow reliable comparisons between individuals.

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Electromyographic Imaging of Motor Unit Architectural Properties

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Introduction. Muscle action potentials are shaped not only by the propagation of the excitation along the fiber, as are nerve action potentials, but also by the initiation of the excitation at the endplate and its termination at the muscle/tendon junctions.[1] As a result, the morphology of the motor-unit action potential (MUAP) is largely determined by motor-unit architecture, including endplate locations, fiber lengths, and pinnation. This dependence makes it possible to extract information about the anatomy of the motor unit (MU) from the shape of the MUAP, and information about the architectural relationships between adjacent MUs from the shapes of different MUAPs recorded during the same contraction at the same site. The objective of this work was to relate MUAP characteristics to MU anatomical features in two architecturally different muscles.

Computer Simulations. The effects of MU anatomical parameters on MUAP morphology are predicted by biophysical models.[2] Figure 1 shows computer simulated MUAPs for three MUs with the same muscle-fiber semilengths but different endplate locations. The MUAPs have one of three characteristic shapes depending on the location of the MU with respect to the electrode: biphasic (initially negative) if the electrode is at the endplate, triphasic if the electrode is between the endplate and the muscle/tendon junction, and biphasic (initially positive) if the electrode is beyond the muscle/tendon junction. Each MUAP has a distinct onset (I), marking the initiation of excitation at the endplate, and a distinct terminal wave (T) marking the termination of excitation at the muscle/tendon junction. If the electrode is between the endplate and the tendon, the latency of the negative peak is proportional to the distance between the electrode and the endplate. In all cases, the latency of the terminal-wave peak is proportional to the mean muscle-fiber semilength.

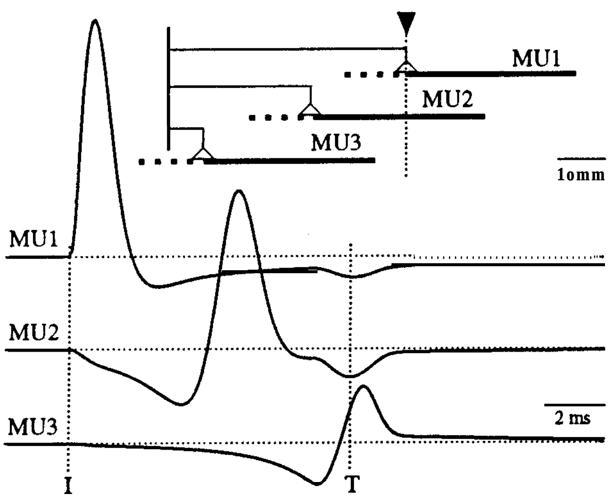

Figure 1. Simulated MUAPs. The spatial arrangements of the three simulated MUs are shown in the upper-right. The heavy solid lines indicate the mean semilengths, extending from the endplate (open triangle) to the muscle/tendon junctions. The filled triangle indicates the axial location of the electrode. The waveforms show the simulated MUAPs with muscle-fiber conduction velocity of 4 m/s. (I) marks the initiation of the excitation at the endplate, and(T) marks the termination at the muscle/tendon junction.

Experimental Methods. EMG signals were recorded using a monopolar needle electrode during voluntary isometric contractions at 10% MVC in brachial biceps (BB) and tibialis anterior (TA) muscles in four normal subjects with no history of neuromuscular disease. lo-second-long signals were filtered at 8 Hz and 8 kHz, sampled at 10 kHz, and decomposed into their constituent MUAP trains using an interactive computer program. The MUAP waveforms were then averaged using an interference-cancellation algorithm.[3] For each MUAP, the location of the endplate with respect to the electrode and the mean muscle-fiber semilength were estimated from the latencies of the negative peak and the terminal-wave peak, assuming a conduction velocity of 4 m/s.

Results. It was typically possible to identify and fully decompose 9- 12 simultaneously active MUs per recording. Full decomposition was important for obtaining MUAP averages with sufficient signal-to-noise ratios to detect the onsets and terminal waves reliably. Figures 2-4 show sets of simultaneously active MUAPs recorded from two sites in BB and one site in TA. The MUAPs from each site were divided into two or more distinct groups on the basis of their shapes. One MUAP in each group is highlighted. The estimated spatial

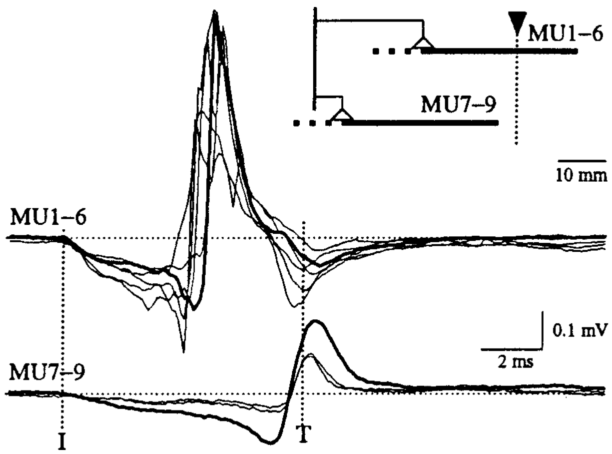

Figure 2. Nine MUAPs recorded from a single site in BB. The MUs in both groups have approximately the same mean semilengths, but for those in the second group the electrode is beyond the muscle/tendon junction.

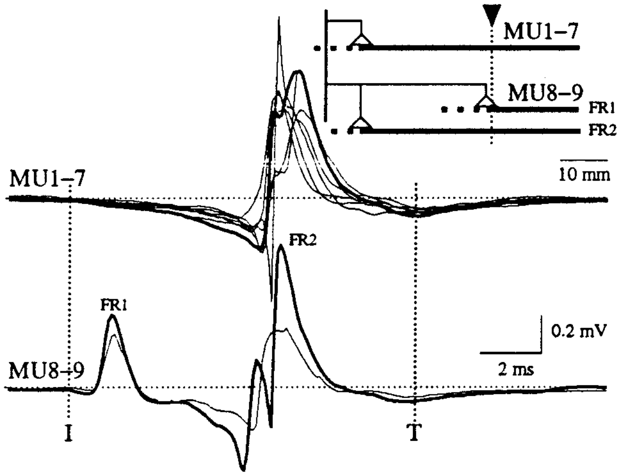

Figure 3. Nine MUAPs recorded from a single site in BB. The MUAPs in the second group have two negative phases separated by 5.5 ms, indicating that their MUs consist of two fractions with endplates separated by about 22 mm.

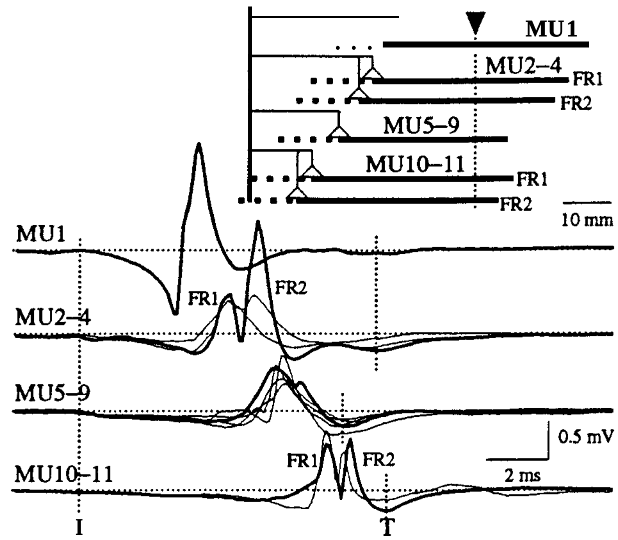

Figure 4. Eleven MUAPs recorded from a single site in TA. The MUs in each group have approximately the same semilengths, but they vary considerably in axial location. Several of the MUs contain fractions.

arrangements of the MUs in each group are shown schematically.

Discussion. MUAPs are conventionally quantitated in terms of their amplitudes and durations. However, these parameters do not have straightforward physiological interpretations. Our study shows that MUAP morphology is strongly determined by MU anatomy, and that anatomical parameters can be estimated from the latencies of MUAP waveform landmarks. An important issue that needs to be addressed is that the monopolar needle electrode 'sees' the electrical activity in the MU from its initiation at the end-plate to the termination at the muscle/tendon junction even in muscles known to have long fiber lengths, as BB. Estimation of distances from latencies requires knowledge of muscle-fiber conduction velocity. Here we assumed a fixed value of velocity for all the MUs, but in reality velocities might slightly differ (up to ±15%).[4]

Our results are consistent with known anatomical information. In particular, the wide spread in axial locations of the TA MUs is consistent with TA's bipinnate architecture and superficially located endplate zone.' The techniques presented here provide an original method for imaging the comparative architectural properties of multiple neighboring MUs.

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S-EMG Utilization in Ruling Out Muscular Symptom Magnification

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La methodologie de EMG de superficie peut etre utilisee dans le cadre Clinique de diagnostique diirentiel entre les personnes qui souffrent de symptomatologie musculaire et ceux qui simulent cette symptomatologie. Des etudes protocollaires sont demonstrees. Elles montrent les diffrences essentielles entre les courbes de l'action electrique durant l'arc du mouvement entre a) les muscles des personnes asymptomatiques, b) les muscles des personnes symptomatiques et c) les personnes qui simulent les symptomes musculaires. Les characteristiques statistiques de consistence versus l'inconsistence sont presentees pour rafforcer les impressions cliniques. L'EMG de surface est donc une methode objective qui peut differencier la simulation de symptomes musculaires de la situation des vrais symptomes.

INTRODUCTION

S-EMG is an objective electrophysiological modality. It is subject to statistical evaluation including strict statistical interpretation of consistency of data. It endeavors to measure muscular activity through rest or motion. Protocols of measurement of bilateral muscular activity through the principal joint range of motion have been devised (Sella, 1993, 1995, 1997) in addition to previous protocols (Basmajian, 1985). The principle of biological and behavioral consistency is paramount as the hypothesis that symptomatic or asymptomatic individuals will have their muscles act consistently through any range of motion and through any number of repetitions. The corollary is true in terms of symptom magnifiers or malingerers. Thus, whether at rest or during motion, Contralateral muscles will act consistently through a number of given repetitions. If a muscle is symptomatic, it will show consistently phenomena such as hypertonus, hypotonus or contracture. If a muscle simulates symptomatology, there is no consistency of showing e.g. hypertonus or hypotonus through any given number of repetitions (Hasson, 1993, Hoffmaster, 1993). This inconsistency is the basis for diagnosing symptom magnification or malingering. Protocols of testing of four Contralateral muscles simultaneously allow for a high level of confidence in ruling out symptom magnification (Sella, 1995 and 1993). Thus, aberrant curves are found on most if not all muscles "spread" through the range of motion in an inconsistent pattern.

METHODS

Well defined parameters of normal resting electrical values and abnormal resting and motion related values have been created. Symptomatic and asymptomatic patterns have been established. Coefficients of variation for the resting values as well as for 5 repetitions patterns through the range of motion have been established. Thus, it is possible to compare the normal electrical activity pattern through rest of motion with the abnormal activity. Symptomatic persons are able to reproduce consistently the abnormal patterns of activity at rest or through motion. The coefficient of variation can define this on any muscle taken by itself or by comparison with its "normal" or "abnormal" Contralateral. By comparison, symptom magnifiers may not show any consistency of curve repetition during motion or rest. The coefficient of variation is consistently >10% and close to 20% by comparison with the one found in either symptomatic or asymptomatic individuals. The statistical method of demonstrating consistency in the electrical activity curve pattern has been proven in the literature (Sella, 1997).

Examples involving motion and rest graphs of persons who are asymptomatic, symptomatic or symptom magnifiers will be presented (the graphs and the statistics would take several pages). The documentation has already been presented in several forums (Sella, 1997 Las Vegas, Sella, 1997 Nashville).

CONCLUSIONS

S-EMG is the only electrophysiological modality which is a amenable to statistical treatment and evaluation. It can demonstrate symptom magnification of muscular suffering via inconsistency of abnormalities of curves of activity through motion and rest as compared to those of similar muscles of asymptomatic or truly symptomatic persons.

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24-hour EMG Recording and Analysis

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Introduction

Information derived from 24 continuous hours of electromyographic (EMG) activity recorded during routine tasks would allow for comparison of muscle activation between different populations or between different testing conditions within the same population. However, the quantification of daily muscle activity raises many challenges mostly related to the hardware and acquisition/analysis protocols required to process such large amount of data. A novel method was designed at the University of California, Los Angeles to record and analyze 24 hours EMG during the Life and Microgravity Science (STS-78) spaceflight mission flown in July 1996. The objectives of this study were to assess the level of activation of postural and non-postural muscles during normal activity in 1G (Earth) and microgravity (spaceflight) environments, and to investigate he recruitment pattern of slow and fast muscles. It has been previously reported in animal studies that muscle unloading brought about a short term reduction in the EMG activity of the ankle extensors and an increase in activity of their antagonist muscles (1, 3), as well as an alteration in the extensor muscles recruitment pattern (4). It was thus hypothesized that daily muscle activation of the lower limb anti-gravity muscles would be markedly reduced in microgravity as compared to 1G in the predominately slow Soleus (SO) muscle while that of the faster Gastrocnemius Medialis (GM) would increase (2). It was further postulated that daily muscle activation of both the elbow flexor (BB) and extensor (TB) muscles would increase in microgravity as compared to 1G.

Method

24-hour EMG activity was collected before (L-90, L-60, L-30 and L-l 5), during (FDl , FD6 and FD13) and after (RO, R4 and R15) spaceflight using a portable system. EMG signals of the SO, MG, TA, BB, and TB muscles were acquired using disposable pre-gelled silver/silver chloride bipolar electrodes placed 2.5 cm apart (center to center) over the bellies each muscle. Two ground electrodes were positioned respectively on the tibia and on the medial epicondyle of the humerus. EMG signals were amplified (gain: 1000) and recorded on two 12 hour audiotapes by TEAC HR-40 recorder equipped with AR-40 DR Amp units (frequency range: 25-560 Hz, amplitude range: ± 1 V). The analog data was subsequently digitized at a sampling frequency of 1000 Hz using a data acquisition card mounted in a Pentium 120 MHz computer. EMG signals were rectified, filtered and smoothed (40 ms moving average), and integrated (iEMG) for every 1000 ms. For each recording session, a matrix of 3600 X 24 estimates of iEMG in Vsec was generated (Fig.1). The data was then stored on disc for further non-automated filtering before any subsequent analyses.

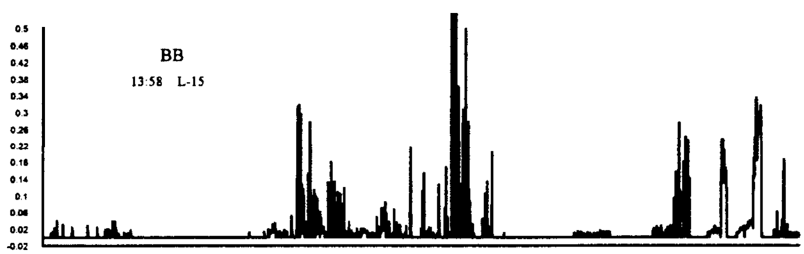

Fig. 1 One hour of Biceps Brachii iEMG (3600 data points) of a typical subject.

A pattern of the behavior of muscle activation for 24 hours was obtained by further integrating the data for every hour (Fig.2). Comparisons of hourly EMG activity was then possible across session, within muscle and within subject to highlight the effects of a testing condition. Figure 2 depicts the hourly iEMG estimates computed from day 1: 8:00 am to day 2: 9:00 am. As could be expected, there is a marked decrease in iEMG values at night. Finally, an estimate of global daily muscle activation could be computed by integrating 24 hours of iEMG values.

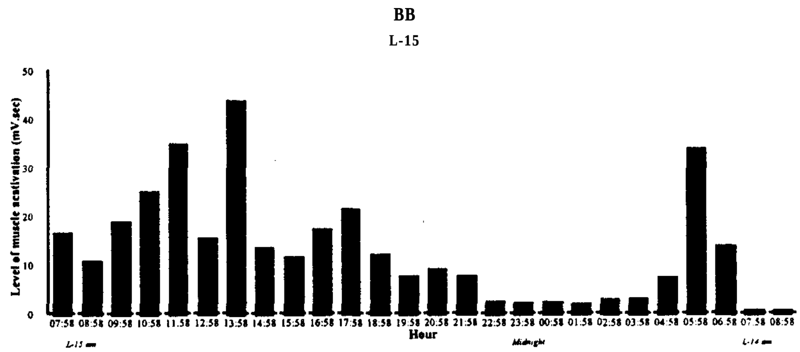

Fig. 2 24-Hour pattern of BB muscle activation obtained from the integration of corrected iEMG for every hour of 24-hour EMG recording at L- 15.

Results & Discussion

Analysis of 24 hour EMG activity recorded on eight healthy male subjects during a 17 days bedrest pilot study shows a decrease in lower limb (SO, GM and TA) muscles activation during bedrest whereas both BB and TB were similarly activated as in pre-bedrest. Furthermore, preliminary data analysis of one crewmember 24-hour EMG data shows a definite increase in muscle activation on FDl for the SO, TA, BB and TB muscles. Overall, the SO and TA daily activation increased during flight, whereas a slight decrease in the GM daily level of EMG activity is observed. The TB also appears to be activated to a greater extent throughout flight as compared to the ground based data. Besides the marked increase in BB activation on FDl, the daily muscle activation estimates computed for FD6 and FD13 are similar to the baseline and recovery values.

These results highlight the potentials of this novel method to record and analyze EMG data recorded over 24-hour periods. However, more studies using this method are needed to validate the use of iEMG estimates and to determine the sensitivity of this measurement tool. Moreover, improved automation of data filtering combined with digital recording of 24-hour EMG data either through telemetry or on portable microcomputers will greatly improve access time to the data.

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The Effects of an Elasticized Lumbosacral Corset and Posture on Myoelectric Activity of the Lumbar Spine Musculature: Repeatability and Preliminary Findings

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INTRODUCTION: Low back pain and dysfunction (LBPD) is a serious problem of large proportions for our society.[1-6] Forward-flexed postures, especially their maintenance, are considered to be a risk factor for LBPD.[6-8] Forward-flexed postures have been shown to increase myoelectric activity and force output of the lumbar spine musculature, and increase lumbar intradiscal pressure these factors have been shown to be well-correlated with each other, and are considered to be indicators of spinal compressive loading.[9-11] compressive loading of the spine is considered to be a risk factor for LBPD,[2,12-16] and a problem affecting the recovery of people who already experience LBPD.[17] It would appear useful, therefore, for both prevention and rehabilitation to identify an intervention that would decrease myoelectric activity of the lumbar spine musculature while people perform activities that require them to maintain forward-flexed postures.

Elasticized lumbosacral corsets (ELSCs) are commonly prescribed by health care professionals, and used in industry and during rehabilitation, for both prevention and treatment of LBPD. There is no evidence in the literature of their effectiveness in preventing or succesfully treating LBPD. Although several beneficial effects have been attributed to external back supports,[18-20] some researchers doubt the effectiveness of these types of supports.[21-23] There is no consensus in the literature on their effects on myoelectric activity of the lumbar musculature.[21,23] In addition, there is a paucity of data specifically on ELSCs.

PURPOSE: The purpose of this pilot study was to assess the repeatability of the surface myoelectric signal detected from the lumbar spine musculature of healthy individuals while they were standing in a forward-flexed position under different conditions of external support. An additional purpose was to perform a preliminary determination of the effect of an ELSC, as compared to no support, on the myoelectric signal of the lumbar spine musculature. Specifically, the initial value of the root-mean-square (RMS), used an indicator of the magnitude of myoelectric activity, is reported on here.

METHOD: Six healthy male volunteers, 2740 years of age, were recruited from the Politecnico di Torino community. These subjects maintained a forward-flexed posture for 10 one-minute repetitions using no external support device, and while wearing an ELSC. The order of testing was determined prior to subject selection such that each condition of support was tested first for an equal number of times, to limit any effect of test order on the results

Each subject stood with the buttocks against a wall and the lower extremities perpendicular to the floor. The feet were placed according to subject comfort approximately shoulder width apart and slightly toed-out The subject was instructed to stand in a comfortable upright posture with respect to the lumbar curve. The subject was then instructed to bend forward (while maintaining the same lumbar curve) so that the angle of the back with the floor was approximately 60" (i.e., inclined forward 30" from vertical) as measured with an inclinometer. This position of testing based in part on research by Schultz and colleagues,[9] was the same for all conditions of support, and was verified in each case by the experimenter.

Surface myoelectric signals were collected bilaterally from the longissimus thoracis at the Ll-2 spinal level, from the iliocostalis hunborum at the L2-3 level, and from the multifidus at the L5-Sl level, using active electrode. probes having two parallel detection surfaces (bars) for acquisition of one single differential signal per probe. Each parallel bar of the probe was 10mm long and lmm thick with a IO-mm separation between them.

The specifications of the signal acquisition system were as follows: (low-pass) filtering such that the overall bandwidth was lo-1,OOOHz at -3dB, with a filter slope (rolloff) of -12dB/octave; the common mode rejection ratio (CMMR) was greater than 90dB; the input equivalent noise was approximately 1microVrms; the linearity error was lower than 2%.

Signal processing occurred off-line. The signals were sampled at 1,024Hz. The samples were digitized and stored using a 486 DX microcomputer, equipped with a 12-bit A/D card (Microstar 2400).

ANALYSES: RMS values were mathematically computed for each second of each one-minute repetition using a Fast-Fourier Transform A least-square regression was performed on the 6* through the 30* second (epoch) of the one-minute time course, for each repetition, to obtain the initial value of the RMS along with its 95% confidence interval. This period was pm-selected for analysis to avoid possible artifact at the beginning of the contraction and Because the behavior of the RMS typically changes with fatigue after this period," which would invalidate the use of a linear regression for the entire period. Any epoch of this period with an RMS value that was more than two standard deviations from the value predicted by the regression line for that epoch was removed as an outlier, considered to be a manifestation of non-stationarities or instability of the motor unit pool. This was a rare occurrence. ln addition, it was found that whether the outliers were removed or not, the initial value of the RMS was not statistically different.

Descriptive statistics, including the mean and standard deviation, were performed on the initial value of the RMS for the 10 repetitions in each condition of support, for each muscle, for each subject From these data, the coefficient of variation (CV) was obtained to be used as an indicator of the measurement variability (i.e., repeatability) of the initial value of the RMS. ln addition a dependent_ two-tailed t-test was performed to determine, within each subject, the difference between the ELSC and no-support conditions.

RESULTS: The mean CVs for the initial value of the RMS for all muscles in all subjects in both conditions of support ranged from approximately .03 to .IO There was no significant difference Between the ELSC and no-support conditions for the initial value of the RMS for all muscles, for each subject CONCLUSIONS: The results for the CVs indicate that the measurements of the initial value of the RMS were highly repeatable, and that the method of data collection used for this pilot study may be appropriately applied to a larger study, using one repetition for each condition of support with a greater number of subjects. The comparison between the ELSC and no-support conditions for the initial value of the **RMS provides** a preliminary indication that an ELSC provided no reduction in magnitude of lumbar spine musculature activity. **If ELSCs** are prescribed and used to minimize indicators of lumbar spine compressive loading then they may not be performing the desired function This research is part of a larger ongoing investigation that includes other conditions of support assessment of fatigue using the slope of the mean frequency of the myoelectric signal, and planned study on injured individuals.

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Effect of Joint Angle on the Amplitude of Surface Myoelectric Signal

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INTRODUCTION

Several theoretical analyses and experimental works have shown an unequivocal relationship between the amplitude of the surface EMG signal and the force produced by the muscle under investigation. Many factors may affect the surface EMG signal recorded during a muscle contraction. However, when one limits the study to isometric activities, all these factors can be neutralized by normalizing the amplitudes of the EMG signal and the force among contractions. Linear and quadratic relationships have been shown between force and amplitude of the surface EMG signal. The nature of this relationship is determined by the muscle under study.

Unfortunately these results cannot be extended to the analysis of dynamic contractions. In fact, when the body segments modify their position, the muscle length changes and a displacement between the surface electrodes and the active muscle fibers is expected. As a result, the spatial filtering introduced by the interposed tissue changes according to the movement. Variations of the muscle length and changes of the spatial filtering introduced by the tissue cause a dependence of the surface EMG signal amplitude on the position of the relative body segment. This observation suggests that the amplitude of the surface EMG signal recorded during a dynamic contraction must be considered as a qualitative measurement of the exerted force, while a quantitative relationship can be identified only when one limits the study to isometric contractions.

This work shows experimental evidence of this statement by analyzing the surface EMG signal recorded from the quadriceps muscle during a set of isometric knee extensions at different knee joint angles and different percentage levels of the maximum voluntary contraction.

METHODS

Ten healthy volunteers (seven men and three women) with no history of knee disorders and age equal to 27 ± 3 (mean ± std) years performed a series of isometric knee extensions by using an isokinetic device (REV9000, Technogym, Italy). The lever arm of the device was set to five different positions (15, 30, 45, 60, and 75 degrees, where zero corresponds to a maximum knee extension). At each of these five positions the subjects were instructed to perform three maximum voluntary contractions. The maximum among the three was defined as the maximum voluntary contraction force (MVC). Then, a series of submaximal contractions lasting 5 s each were executed at 20, 35, 50, 65, and 80 % MVC. Three contractions were performed at each submaximal level. The order of the submaximal contractions was randomized. Subjects were instructed to rest for three minutes after each contraction.

Surface EMG signals were recorded from three muscles: rectus femoris (RF), vastus medialis (VM), and vastus lateralis (VL). Two couples of electrodes 15 mm apart were positioned on each muscle. Single differential electrodes with 15 mm interelectrode distance were utilized. Surface EMG signals were sampled at 1024 Hz by a 12 bit A/D converter. Signal analysis was performed as follows: the time-course of the torque measured by the isokinetic device was considered and the 1 s segment closest (in a least square sense) to the target level was selected. The root mean square (RMS) value was estimated for the corresponding EMG segment. Data were normalized according to the value obtained for the 80 % MVC contractions.

A stereophotogrammetric system (ELITE, BTS, Milano, Italy) was used to study the limb segment and the lever arm position. Rigid plates, mounted with four retroreflective marker balls, were attached to the distal thigh and proximal shank by wide elastic straps. Two further rigid clusters of markers were fixed to the sliding cursor and the base of the isokinetic device. In addition to relate the limb segment position to the marker arrays, the position of anatomical landmarks were recorded, using a pointer technique proposed by Cappozzo et al. The knee joint angle was calculated from the positions of the anatomical frames by using the JCS convention of Grood and Suntay.

RESULTS

The RMS value of the surface EMG signal vs. % MVC and vs. knee joint angle were considered. Data relative to the two electrode couples were compared by means of a Wilcoxon matched pair signed-rank test. The RMS value of the surface EMG signal vs. % MVC showed the expected relationship between EMG amplitude and force, i.e. an almost linear increase of the EMG amplitude when force increased. Besides, the Wilcoxon test showed that this relationship was not statistically different for the two couples of electrodes for more than’90 % of the plotted curves. At the opposite, the RMS value of the EMG vs. knee joint angle showed an unpredictable pattern and a low correlation between the results obtained by the two couples of electrodes. In fact, in almost 60 % of the tests the relationship between the RMS value of the EMG and knee joint angle was found different.

The knee joint angle estimated through the stereophotogrammetric system allowed to demonstrate a significant difference between the expected angular value and the knee joint angle. Values estimated for different subjects were found in a relatively wide range around the average. The knee joint angle measured at 80 % MVC was shown higher than the angle measured at 20 % MVC as an amount of approximately 10 % of the initial value. Besides, for low joint angles (knee almost fully extended) the average value was found higher than the expected one (i.e. the lever arm position). Oppositely, when high values were considered the average value was found lower than the expected joint angle.

DISCUSSION

The results of this work show that the amplitude of the surface EMG signal is strongly dependent on the electrode position on the muscle belly. While the relationship between the RMS value of the surface EMG signal and the force produced by the muscle is well assessed when one considers isometric contractions, this is not the case when dynamic contractions are studied. An erratic and unpredictable behavior of the RMS value of the EMG is observed when the body segment position changes. This shows that the combined effect of the variations of the muscle fiber length and the displacement between the active muscle fibers and the surface electrodes is unpredictable. Since the effect of the variations of the muscle fiber length is not supposed to be erratic, it is hypothesized that the main cause of surface EMG amplitude changes is related to the variations of the tissue spatial filtering. Therefore a function which correlates the force produced by the muscle and the amplitude of the surface EM cannot be defined when dynamic activities are studied. Besides it is shown that the knee joint angle is significantly different from the expected one and that there are variations of the joint angle which depend on the percentage MVP level of the contraction. This may be due to the compliance of the soft part of the isokinetic device seat and to a lack of constrain of the lower limb during the exercise. These factors also affect the EMG signal amplitude.

CONCLUSIONS

The results proposed in this work show that the amplitude of surface EMG signals recorded during dynamic contractions must be considered as a qualitative measurement of the force exerted by the muscle under investigation, while a quantitative relationship cannot be found.

Short Term Cross-Correlation Analysis of Biomechanical Signals

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INTRODUCTION

The cross-correlation coefficient is often used as a relative measure of the association between two signals of interest. Typical implementations of this measure involve the calculation of one cross-correlation function describing the entire data range. Such an analysis prevents the characterization of time dependent associations between the signals, masking relationships that may exist at higher frequencies. This project involved the development of a short term cross-correlation function (STCCF) algorithm which preserves higher frequency interactions. The utility of the STCCF was demonstrated in preliminary investigations of postural, kinematic, and electromyographic data.

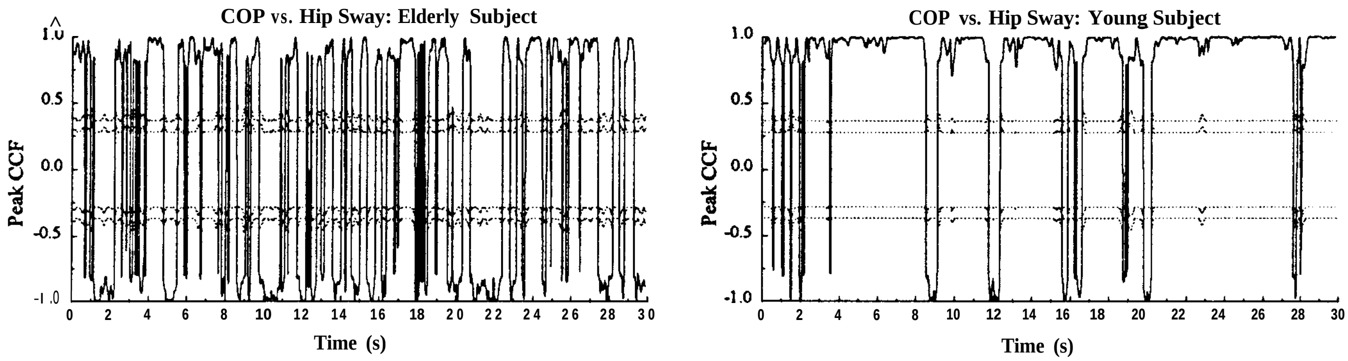
METHODS

Conventional cross-correlation analyses describe the general relationship between two signals by using the entire data range to calculate one cross-correlation function [1,2]. In contrast, STCCF analysis involves the calculation of multiple cross-correlation functions corresponding to overlapping epochs of the data range. The length of each epoch is constrained by the assumption of signal quasi-stationarity. When data is assumed to be normally distributed, Pearson's linear correlation coefficient is used to characterize the ***linear*** relationship between two time series. Statistical significance thresholds of $P<.05$ and $p<.01$ are then calculated using Student's t-test. In contrast, the need to determine statistical significance does not permit a linear correlation of data that is not normally distributed. These data sets are compared using Spearman's rank order correlation coefficient, which characterizes the ***association*** between the two data sets. Significance is again determined using Student's t-test. In either case, the resulting three-dimensional cross-correlation function describes the time-dependent relationship between two time series. The STCCF was visualized as a surface with Time, Time Lag, and Correlation Coefficient axes. This project included the development of a GUI based software package to estimate the STCCF between multiple data sets. Both linear (parametric) and rank-order (non-parametric) correlation coefficients are implemented, along with a variety of pre-processing and plotting functions.

RESULTS AND DISCUSSION

STCCF analysis was performed on anteroposterior center of pressure (COP) displacement and hip sway data acquired during quiet standing in order to detect short duration, high frequency postural corrections. If the body is considered an inverted pendulum during quiet standing, the maintenance of posture can be expected to produce high-frequency sign changes in the correlation between COP displacement and hip sway. With an initial movement of the hip in the anterior direction, contraction of the ***triceps surae*** will move the COP anteriorly in an effort to maintain balance (positive correlation). As the COP continues to move anteriorly, the hip will begin to accelerate in the posterior direction, eventually reversing direction and moving posteriorly (negative correlation). As the hip moves towards the centerline, activation of ***tibialis anterior*** will cause the COP to decelerate and begin to move in the posterior direction (positive correlation). Over-compensation for the initial hip displacement may result in several oscillations of this behavior until equilibrium is achieved. To be an effective means of postural control, this type of corrective behavior must occur at frequencies higher than the low frequencies that dominate these time series. For this reason, the COP and hip position signals were high-pass filtered (.5Hz cutoff) before the cross-correlation analysis was performed. The typical results shown in Figure (1) are consistent with the previously described chain of events and suggest that STCCF analysis may be useful for quantifying an increased occurrence of corrective postural adjustments in elderly subjects.

A second demonstration of STCCF analysis involved the study electromyographic (EMG) signals from the muscles of the lower back during sustained isometric contractions. We tested the hypothesis that a synergistic interaction between the *Longissimus* (Ll) and *iliocostalis (L2)* muscles may be manifest as an alternating muscle activation pattern designed to reduce muscle fatigue. Such a dynamic load sharing activity would cause alternating periods of positive and negative correlation between the EMG RMS of the muscles being investigated. A preliminary test of this hypothesis involved the study of one healthy male subject under normal conditions, in an asymmetrically fatigued state, and after the onset of acute asymmetric muscle soreness. The RMS amplitude of the EMG recorded from both right and left LI and L2 sites were analyzed in the various combinations by non-parametric STCCF. The representative results from this subject are shown in Figure (2). There were no sustained statistically significant negative peak correlations seen between any of the muscle sites. These results, therefore, do not support the proposed hypothesis under isometric conditions. Rather, they suggest that there is a consistent positive correlation between longissimus and iliocostalis force production during sustained isometric contractions.



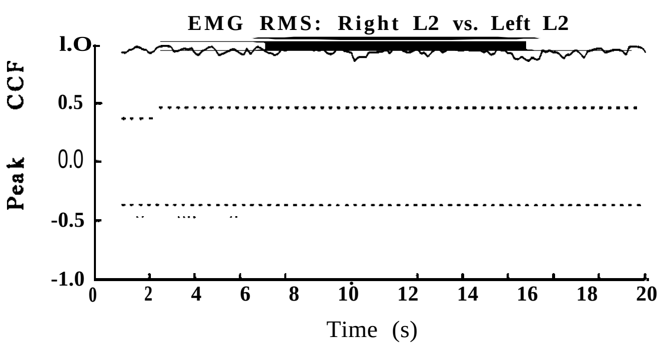


Fig. 1 (above). Comparison of non-parametric short-term cross-correlation peak (max. absolute value) between COP and hip position in elderly and young subjects during quiet standing. Dashed lines indicate statistical significance thresholds of P<.05 and p<.01.

Fig. 2 (left). Comparison of non-parametric short-term cross-correlation peak (max. absolute value) between Right L2 and Left L2 EMG RMS during sustained 60% Ideal Body Weight isometric contraction. Trial recorded during period of extreme asymmetric (right-sided) lumbar muscle soreness. Dashed lines indicate statistical significance thresholds of P<.05 and p<.01.

CONCLUSION

Short term cross-correlation analysis is an effective tool for the analysis of biomechanical signals. By preserving the time dependence of the relationship between time series, the STCCF provides insight into the complex control mechanisms of the human body. While the preliminary investigations presented here require further study before conclusive results can be obtained, they demonstrate the utility of STCCF analysis for the characterization of biomechanical signal relationships.

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User Independent Detector of Muscle Activation for Gait Analysis

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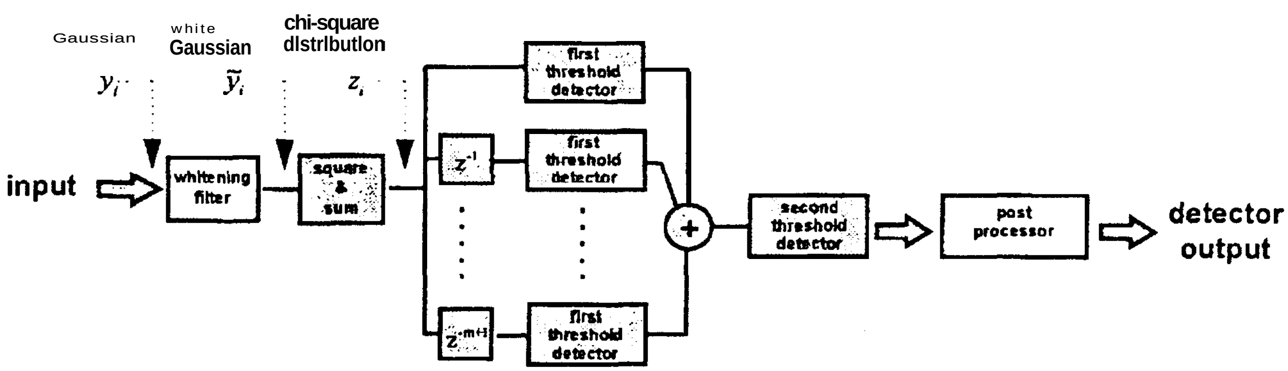
INTRODUCTION

The measurement of muscle activation intervals from surface EMG signals is relevant to the clinical study of human gait. In fact, activation intervals have been shown of paramount importance in evaluating prosthesis implants, planning the surgical treatment of cerebral palsy children, and assessing spastic paresis, just to cite a few examples. Muscle activation intervals are usually estimated by applying a single-threshold detector to the EMG signal. We demonstrated theoretically as well as experimentally that a specific family of double threshold detectors allows to obtain more favorable results than those achieved by applying the single threshold approach.

The scheme of the double-threshold detector we previously deveIoped and characterized is depicted by the figure reported below. This detector relies on the assumption that surface EMG signals may be modeled as realizations of a zero-mean gaussian stochastic process. First, the raw EMG signal is uncorrelated by using a whitening filter, then an auxiliary series is built by summing the square values of two consecutive samples. This auxiliary series has a chi-square distribution with two degrees of freedom. We may therefore calculate the probability that a specific noise sample is above a first threshold. Then, the probability that r samples out of m are above the first threshold may be derived by considering a Bernoulli distribution, and finally the false alarm and the detection probabilities can be determined. The second threshold is represented by the minimum number of samples (r_o) out of m that must be above the first threshold to assume that the EMG signal is present. A post-processor which rejects activation intervals shorter than 30 ms is finally applied to the detector output, to abate the incidence of erroneous transitions **due** to the statistical nature of the algorithm.



Unfortunately, the use of this technique assumes that a sample of the baseline noise is available to determine the value of the first threshold, and this is not always a realistic assumption. Moreover, even if the baseline noise power were known, the user should choose the statistical characteristics of the detector, namely its false alarm probability (P_{fa}) and the probability of signal detection (P_d). This choice is requires an outstanding expertise and it is crucial to obtain acceptable results. It follows that the technique discussed above gives reliable **results only** if used by a well trained operator.

In this work, we propose an improvement of the double threshold statistical detector described above. The purpose is to obtain a totally user independent algorithm that gives results comparable to those that could be achieved by an expert user. The performances of this algorithm have been carefully evaluated by analyzing computer synthesized realizations of EMG signal as well as real recordings.

METHODS

The method described below applies to the EMG signal detected during gait and specifically it was evaluated in the framework of single stride analysis. To be consistent with the experimental protocol adopted by our laboratory to analyze gait, we assumed the length of the EMG recording to be 3 s and the sampling rate equal to 1 kHz. The considered time interval corresponds to that sampled by a stereophotogrammetric system, while the subject moves within the field of view of the cameras.

To characterize the performance of the detection algorithm, we derived its receiver operating characteristic (ROC) curves corresponding to different choices of the detector parameters. Different values (ranging from 0.001 to 0.10) of the probability that a specific noise sample is higher than a certain level were considered (first threshold). The number of samples contemporarily considered by the algorithm (m) was fixed equal to 5. All the possible values of the second threshold (r_o) between 1 and 5 were tested. Signals with different SNR (ranging from 2 to 10 dB) were analyzed. A single interval of EMG activity was simulated. We considered time supports of the EMG signal equal to 20, 40, 60, and 80 % of the entire segment.

To obtain a totally user independent procedure, first we evaluate the signal quality and its relative time-support as follows. In order to estimate the noise variance, we consider 100 different EMG signaf segments obtained by randomly positioning a time-window lasting 100 ms along the time axis. The segment whose variance is the lowest is assumed as representative of the background noise, and the variance represents the noise mean power.

After estimating the noise power, the signal is classified in one out of three classes according to the estimated value of the signal to noise ratio. This is defined as low (below 5 dB), medium (from 5 dB to 10 dB) or high (over 10 dB). Then, the we estimate the support of the EMG signal normalized with respect to the stride duration and we classify the realization into one out of four classes (time support of the signal approximately equal to 20%, 40%, 60%, and 80%).

On the basis of the signal to noise ratio and of the estimated time support, we select the three strategies which, in average, give the best detection results in the specific case and we apply them to the EMG signal. Each sample is then considered as signal or noise by applying a voting technique on the results obtained on the same sample by using the three selected techniques.

RESULTS

The ROC curves demonstrate that if the SNR is as high as 10 dB, the operating point of the algorithm can easily be located in a region corresponding to a false alarm probability lower than 5 % and a probability of signal detection higher than 95 %. When the SNR is as high as 5 dB, the value of P_{fa} can still be kept lower than 10 % and P_d higher than 90 %. If the SNR is as low as 2 dB, P_d drops below 70 % if P_{fa} is to be kept lower than 10 %.

The characterization of the entire user independent procedure showed that the methods proposed to compute the noise variance and the signal percent support are sufficiently accurate to guide the selection of the detection strategies. Moreover, when processing synthesized as well as real signals the proposed user independent procedure led to results almost superimposed to those obtained by an expert user.

CONCLUSIONS

In this work we present the results obtained by modifying an algorithm we previously proposed to estimate muscle activation intervals during gait. Possible limitations of the method when either the noise characteristics are unknown or the background noise shows nonstationary variations of the amplitude are overcome. The procedure we developed is entirely user independent and allows to obtain acceptable results when processing real myoelectric signals, provided that the signal to noise ratio is over 5 dB.

Motor Units - How Many, How Large, What Kind ?

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Introduction

The modern era of motor unit physiology can be said to have begun in 1965, with the first published reports, by Henneman and colleagues, of single motor unit contractile responses in the cat (4,6). Within a few years it became possible to obtain similar data in human muscles, using various techniques for motor unit isolation. At the same time it was realized that a simple electrophysiological method would enable the number of motor units to be determined in intact muscles. The results of theses studies in human muscles form the subject of the present review.

Motor units - how many?

The number of motor units in a muscle can be estimated by comparing the amplitudes of single motor unit action potentials with the maximum response (M-wave) evoked from the whole muscle. In the original method (5), a sample of motor unit potentials was obtained by the technique of incremental stimulation of the motor nerve. Although the method continues to be of considerable value in the diagnostic clinical EMG laboratory, the motor unit number estimates (MUNEs) tend to be artificially high. This complication is due to the ability of overlapping axon thresholds to produce fictive motor unit potentials during stimulation. To overcome this 'alternation' phenomenon, other motor unit sampling methods have been developed, and at the present time at least 8 have been described. Our own approach has been to use a computer program to control muscle responses (2). Successively larger responses are stored as templates and, by subtraction, yield putative single motor unit potentials. Each putative potential is then compared with every other one in the sample and any similarities attributed to alternation are discounted from the MUNE. Another algorithm compares the configurations of the maximum M-wave and the summated sample of motor unit potentials, and in this way assesses the adequacy of the sample.

Table 1 shows the MUNEs for various human muscles and muscle groups, obtained with a number of different methods in subjects below the age of 60 years; where multiple values are available for the same muscle/muscle group, a grand mean is given.

Table 1. MUNEs derived for human muscles by different methods.

Source	Muscle(s)	MUNE (Mean ± SD, n)
Multiple studies	Thenar group (median innervated)	208
Multiple studies	Hypothenar group	263
Galea[a]	Biceps brachii	109 ± 43 (80)
Doherty et al.[1]	Biceps-brachialis	357 ± 97 (24)
Multiple studies	Extensor digitorum brevis	143
Multiple studies	Plantar group	333
Galea[a]	Tibialis anterior	256 ± 107 (22)
Galea[a]	Vastus medialis	224 ± 112 (24)

[a] Personal communication.

Motor unit numbers decline after the age of 60 to extents which depend on the individual and **on the** muscle involved. The availability of extensive control data has allowed MUNE to be employed in the diagnostic EMG laboratory. Using the automated incremental method over a two year period, we found a 27% incidence of abnormal MUNEs, with the highest detection rates being found in amyotrophic lateral sclerosis, spinal muscular atrophy, focal and generalized peripheral neuropathies, and lumbosacral nerve root lesions. It must be added that MUNEs, like other parts of the EMG examination, are often performed to exclude the possibility of denervation and that this strategy will obviously lower the detection rate of abnormalities in the clinic.
As an example of diagnostic MUNE, Figure 1 shows values for the small muscles of the hand in 63 diabetic patients referred because of possible peripheral neuropathy. Although there is a high incidence of abnormally low values, indicative of neuropathy, the sensory nerve involvement is much worse, as the right part of the figure shows.

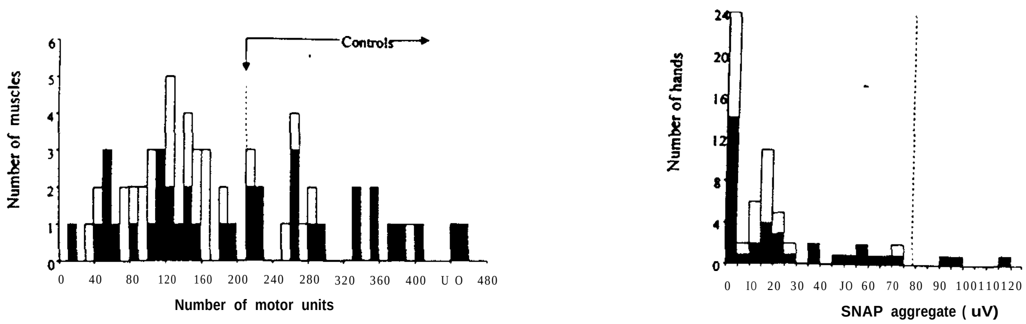

Figure 1. Left. MUNEs in thenar and hypothenar muscles of 63 subjects with diabetes mellitus. *Right.* The aggregate of the sensory nerve action potentials recorded orthodromically from each digit of the same hands. The filled and open columns show the results for subjects below and above the age of 60 years respectively. Vertical interrupted lines denote lower limits of control ranges. Note that the sensory nerve fibers are more severely affected than the motor ones.

Motor units - how large?

In human muscles it would be very difficult to determine how many muscle fibers are supplied by a single motor axon and its parent motoneuron, but it is relatively easy to assess the relative sizes of the units, by comparing their twitch responses. In keeping with animal studies, the relative sizes differ considerably, even in the same muscle, and lOO-fold ranges have been reported by several laboratories. With the advent of partial denervation, the motor unit sizes increase further, as collateral reinnervation takes place. The relationship of the mean motor unit potential amplitude to the number of surviving motor units is a hyperbolic one, indicating that successful axon sprouting can keep pace with the numbers of denervated muscle fibers present. In long standing denervation, however, it is probable that this powerful compensatory mechanism eventually breaks down, as motoneurons start to undergo metabolic failure.

Motor units - what kind?

In the first examination of single human motor unit twitches, in the extensor digitorum brevis (7), a threefold variation was found in the contraction times, and this has been a consistent feature of subsequent studies. Unlike motor units in other mammalian species, however, there is a less stringent relationship between contraction time and force output, and it would be interesting to determine to what extent either or both of these parameters are correlated with the every day usage of the units. Again, although the effects of disuse and denervation in human muscle fibers are evidence of the controlling influence of the motoneuron, in some respects this influence is less dominant than in other mammals. Thus in two studies from our laboratory we have found an inability of the twitch durations to change in the directions predicted from animal studies. In one study muscle and nerve were grafted to restore function in unilateral facial palsy (3), and in the other study twitches were compared in the normal and affected arms of children with persisting obstetric brachial palsy (Brown et al., unpublished results). Why there should be such differences between humans and other investigated species is not known, although the considerably greater ages of the former may be one factor.

Conclusions

From the number and variety of published reports, it is evident that the study of motor units in human subjects is, in many respects, as advanced as that in other species. Indeed, in some respects the human situation is superior, in that the physiological properties of the units can be matched against voluntary activation of the same units. In addition, by using an appropriate stimulation technique, it is possible to perform repeated studies on the same motor unit and thereby to examine changes brought about by such factors as aging, disuse, injury and disease.

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Are Direct Measurements of Maximum Back Extensor Strength Really Necessary ?

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INTRODUCTION

In assessing the functional capacity of low back pain patients, a knowledge of the maximum strength of the back extensors is often required, both for its intrinsic value in reflecting the size of the muscle mass and also for establishing the submaximal target force to be used in any assessment of muscle fatigability. However, back muscle strength can be a difficult attribute **to assess.** It is impossible to ascertain whether patients are performing badly due to real physical impairment (e.g. reduced muscle mass), pain, or psychological/behavioural factors such as failure to understand the degree of effort required, anxiety related to the test, depression, illness behaviour or malingering. The aim of the present study was to ascertain whether back extensor strength can be accurately predicted using either indices of body mass, or using the 'twitch interpolation technique'.

METHODS

Subjects 403 subjects (371 female, 32 male; 19-39 y) participated in the first part of the study and a further 5 (33-43 y) took part in the experiments involving twitch interpolation. None had a history of serious low back pain.

Measurement of maximum voluntary isometric contraction (MVC) of the back extensors. Subjects stood in a wooden frame with the knees slightly flexed and with the hips strapped to a crossbar, and pulled upwards with maximum effort on a handlebar attached by a metal chain to a floor-mounted load cell. The handlebar was constrained to move in the vertical direction only and the bar height was adjusted so that the subject's lumbar spine was flexed by 70% of its range between erect standing and the fully-flexed (toe-touching) position, as measured with the 3-Space Isotrak. The output from the load cell was displayed on a digital voltmeter, which the subjects could observe, and was simultaneously A-D converted at 60 Hz and input to a microcomputer. Between three and five MVC attempts were required to obtain a consistent maximum. With an estimate of the relevant lever arms, the extensor moment at MVC was calculated, to which the moment required to counterbalance the upper body weight was added, giving the total maximal extensor moment at MVC (= TotEM, Nm).

Electrical stimulation/twitch superimposition protocol. Pairs of large silicon surface electrodes, covered with conducting gel, were securely strapped to the skin overlying the paraspinal muscles, bilaterally, with the mid-point of the caudal electrodes at the level of 4th lumbar vertebra. The average twitch force at each voltage was calculated as the increment in force above the voluntary force being held. Sub-maximal contractions were held for lo-15 seconds, at 5, 10, **15** %MVC and so on up to 80 %MVC, with 2-3 min rest between each. Subjects were instructed to pull rapidly up to the designated force level and to hold this as steadily as possible whilst single, square-wave stimuli, with a duration of 50 μs, and up to 4OOV, were superimposed at a frequency of 0.5 Hz. The average twitch force (increment in force above the voluntary force being held) at a given submaximal force output was calculated as the mean of five twitches.

RESULTS

Prediction of maximum back extensor strength from total body mass and fat-free body mass. Maximum extensor moment showed a highly significant correlation both with total body mass and with fat-free body mass (Fig 1). Analysis of covariance revealed that the relationship between body size and strength was gender-dependent, in terms of both the intercept ($p<0.05$) and the slope (p<O.OOl).

Prediction of maximum back extensor strength from submaximal efforts with twitch superimposition. The twitch force decreased as the submaximal force being sustained increased towards maximal (Fig 2). Above approximately 65-70% MVC, however, the twitches became indistinguishable from the baseline noise due to their diminished size in combination with the effort-induced tremor during the contraction. On an individual basis, a curvilinear relationship between twitch force and the submaximal force being held was apparent and this was always highly significant ($p<0.001$) However, when the results were extrapolated, to give the voluntary force equivalent to zero twitch force, the predicted

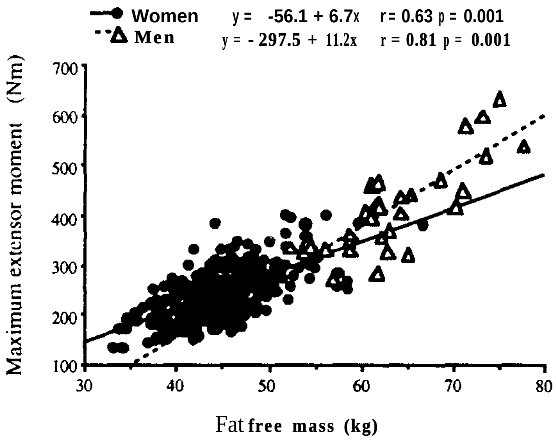

Fig 1. Relationship between maximal extensor moment generated and fat-free mass in women (circles) and men (triangles).

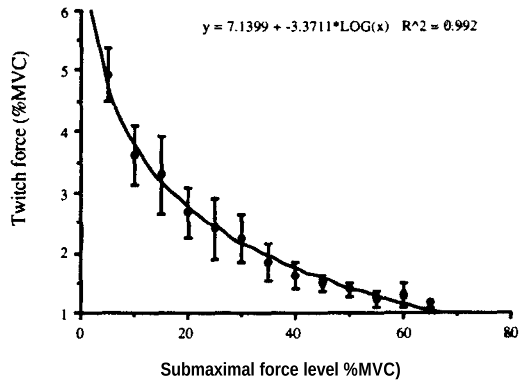

Fig 2. Decrease in twitch force with increasing submaximal force being sustained. Group mean (±SE) results.

MVC consistently underestimated the subject's previously measured MVC (by an average of 18%). Nonetheless, for the group, there was still a highly significant relationship between the predicted and the measured MVC ($p<0.01$). The values of strength predicted on the basis of body size were, in general, more accurate than those obtained using twitch interpolation, in this group of subjects.

DISCUSSION

The back extensor muscle strength testing device was designed to enable measurement in a standing, slightly flexed forward posture one which closely approximates that in which the back muscles are often utilised during activities of daily life. This was not a commercial device, however, and the absolute values of strength may only be appropriate for this particular mode of measurement. Other devices, which employ different measurement techniques and postures, would require their own set of equations.

Both body mass and fat-free mass were good predictors of maximal extensor strength. The group of individuals examined in the current study was large, and the predictive equations should therefore be valid (within the given error of measurement) for any similar, healthy population. Different equations would most likely be needed for older individuals, as the cross-sectional area (and, hence, Iikely the strength) of the paraspinal muscles declines with age.

There was a good relationship between the true maximal back muscle strength and that predicted by the twitch interpolation technique, although the true values were consistently under-predicted. This phenomenon may have been the result of unwanted compliance in the force transmission/measurement system used: the twitch force generated by the back extensors must be transmitted through a number of elastic tissues and joints before being registered by the load cell. However, the technique could still be used to establish the general presence of submaximal effort, by the existence of clear increments in force with superimposed twitch stimulation beyond 70% of a 'supposed' MVC.

The drawbacks of the twitch interpolation technique are that it involves patient effort and is somewhat less accurate than the predictions based on body mass. However, whilst the body mass equations are good for determining how impaired a patient is compared with 'normal', or for setting standardised submaximal loads — e.g. in the assessment of muscle fatigability or for rehabilitation training purposes — they would clearly be of no use in longitudinal studies designed to examine modifications in maximal strength, for example following rehabilitation. In these circumstances, the maximum value predicted using twitch interpolation may be preferable; even if this technique underpredicts, it would be expected to do so systematically over repeated trials in each individual, unless enormous gains in strength were acquired.

In conclusion, where circumstances preclude the direct measurement of MVC, it may be possible to obtain a workable estimate using either predictive equations or twitch interpolation.

ACKNOWLEDGEMENTS We are grateful to the Arthritis and Rheumatism Council (UK) the Medical Research Council (UK) and the Schulthess Klinik Research Fund (Switzerland) for financing this work. We would like to thank Chris Standell for her technical help.

Spinal Moment Contribution of Load Handling Estimated from Back Extensor EMG Applying Artificial Neural Network Technology

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INTRODUCTION

A methodology was proposed for practical ambulatory assessment of low back load exposure [11, approaching the accuracy of 'golden standard' elaborate laboratory methods. It uses 2 small inertial sensor modules assessing absolute posture and movement of trunk and pelvis, 2 or more EMG sensor modules recording trunk muscle EMG and a portable data acquisition system. Output load exposure parameters include kinematic parameters of trunk and pelvis, trunk muscle EMG patterns and spinal moment estimations around a single rotation center in the intervertebral body at level L5/S1 .

The spinal moment is estimated in 2 separate components. First there is the contribution of weight and inertia of the trunk and head estimated from trunk kinematics alone. A second component is the contribution of weight and inertia of arms and load handled or external force present. A key role is played here by a self-learning calibration system predicting this component from back extensor EMG signals and trunk kinematic data.

This paper discusses validation of the estimation method of the second component against a 'golden standard' reference system in lifting experiments

METHODS

The moment contribution of trunk and head is estimated using a simple two segment cantilever model presented and validated before [2]. Inputs for the model are: trunk absolute tilt, trunk angular velocity and acceleration, all in the saggital plane of the trunk, and trunk vertical acceleration and 4 antropometric parameters describing size and weight of the subject.

The moment contribution of the arms plus load handling is estimated using EMG signals from the back extensor muscles. Considering bad occasion to occasion reproducibility a reliable estimation of spinal moment from EMG can only be given when the EMG-moment relationship is calibrated in every recording occasion covering the relevant range of postures and movement and include a representative set of muscles. A typical traditional approach would include an elaborate set of separate recordings of moment and EMG from several muscles under many combinations of posture, movement and external load. This is not acceptable for ambulatory assessment.

An alternative, potentially more practical method for this calibration applies artificial neural network technology in creating a self-learning calibration system that only needs a small set of random postures and movements data with different external loads.

The back propagation type neural network is trained in supervised mode using as inputs Smoothed Rectified EMG signals derived from 4 positions on the erector spinae musculature. These positions are 3 cm left and right from the spine at level TlO and L2. To enable the calibration system to include modulation of muscle length and contraction velocity trunk and pelvic absolute tilt, trunk angular velocity (and trunk angular acceleration) are offered as inputs. The network only output is the back extension moment. The back propagation network has 1 hidden layer with 5 levels.

A training set was derived from 4 recordings of 20 seconds where a subject was asked to perform back flexion and extension movements with increasing or decreasing swiftness and with or without a stop in the middle of the movement. This was done with 2 different weights (0 and 15.7 kg) held in a standardized, exactly known and steady fashion. This enabled direct estimation of the moment contribution using kinematics of arms and load. These estimations were offered together with the input signals to the artificial neural network under training as desired outputs, thus enabling supervised learning.

Selection of learning moments from the data was guided by the desire for a uniform distribution of the spinal moment. The calibration was regarded valid as long as EMG electrodes and movement sensors would stay in the same place and orientation.
The method was validated by direct comparison of spinal moments against a ‘golden standard’ reference method using a ground reaction force recording from a force plate, lower extremities and hip kinematics determined using a video based movement analysis system (Vicon) and a linked segment model of the lower extremities [3]. Validation included single lifts with different lifting techniques (symmetric leg, back and free lifts plus asymmetric free lifts) at different lifting speed (slow and fast) and with different loads (6.7 kg and 15.7 kg).

RESULTS

Fig. 1 shows 2 typical results for lifting experiments from the same subject (single slow leg lift with load of 15,7 kg and free asymmetrical lift of 30 degrees out of the saggital plane with 6.7 kg). The upper trace shows the load handling contribution to the spinal moment predicted by the calibration system (‘load predicted’) and directly calculated by the algorithm used for estimation desired output values in the learning phase (‘load model’). The second plot shows total spinal moment estimated by the reference system (‘ref), the moment contribution of the trunk and head alone (‘trunk’) and the sum of the both trunk and head component and load handling component (‘trunk + load’).
Experimental conditions allowed only valid estimation of the load component after load lift off (vertical line).

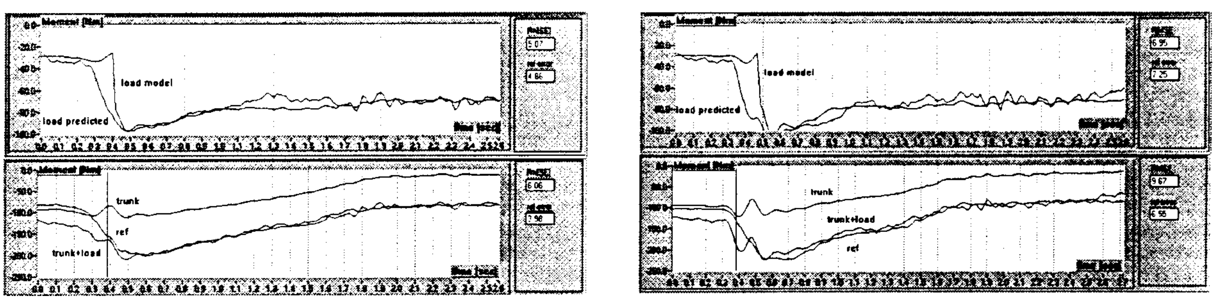

Fig I: Estimated moment curves for leg lift (left) and free asymmetric lift at 30 degree (right). Upper trace: predicted and trained data from the calibration system. Lower trace: Predicted trunk (‘trunk ') and total trunk + load hanclling components (‘trunk + load’) against reference estimation of total spinal moment (‘ ref').

The estimation of total spinal moment stayed for all lifting techniques well within 10% average relative error. Both trunk and load component contributed significantly.
By feeding only one or part of the input signals as input to the calibration system it was determined that the artificial neural network used mainly the EMG signals for estimating the moment contribution of load handling. Using any combination of the kinematic parameters gave a poor estimation of the load handling contribution over the whole lifting trial. Adding kinematic parameter inputs to the EMG inputs gave a systematic but small improvement.
Typical errors in overall moment estimation were similar over all trials with one person but differed over persons, with the largest error in a rather small subject. This indicates that this was probably caused by inaccuracy in the antropometric parameters.

EMG Assessment of Back Muscle Function During Cyclical Lifting

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INTRODUCTION: Spectral electromyographic (EMG) techniques for assessing paraspinal muscle function have been traditionally limited to isometric, constant-force contractions. During a dynamic contraction, changes in muscle force, muscle length, and electrode result in a nonstationary signal which precludes the use of traditional spectral analysis procedures. This paper utilized a time-frequency analysis procedure to estimate the frequency compression of the surface EMG signal from paraspinal muscles during a cyclical lifting task . The purpose was to identify whether EMG spectral changes among concurrently active paraspinal muscles during repetitive trunk extension produces a "fatigue pattern" of EMG spectral behavior that is indicative of normal back functioning. Such results may provide a normative database from which measurements obtained from persons with low back pain (LBP) can be compared to identify impairment.

METHODS: Four healthy subjects without LBP (26 ± 4 yr. ; 3 males, 1 female) were tested in both a Back Analysis System (Roy, S.H., et al *J Rehab R&D, 33(4): 37-47, 1997),* for sustained isometric contractions, and a LIDO-Lift Controller (Loredan Inc.) for repetitive lifting and lowering of a weighted box . The sustained isometric contraction was produced at 80% of the subjects maximum for a 25 s duration. For the lifting trials, the weight of the box was set to 10 % of the subject's static one-lift maximum at a repetition rate of 12 lifts per minute for a maximum of 3 minutes. EMG signals were recorded concurrently from six bilateral lumbar paraspinal muscles. For the isometric trials the EMG median frequency was calculated using a fast Fourier transform. For the cyclical lifting trials, the *instantaneous* median frequency, was calculated utilizing a time-frequency transform of the Cohen Class, referred to as the Choi-Williams transform. The details of this procedure have been described elsewhere (Bonato, P., et al. *IEEE Eng in Med & Biol, 15, 102-l* 11, 1996).

RESULTS AND DISCUSSION: An example of the time course of the median frequency estimates for a typical subject are presented for cyclical lifting (Fig. 1). It can be seen that the instantaneous median frequency behavior is complex, with distinct periods of decay followed by recovery. The effect of this behavior is that it complicates our ability to describe a "normal fatigue pattern" for cyclical lifting using a linear model of median frequency decay. We are, however, encouraged by the ability of the time-frequency analysis procedure to provide the resolution and apparent sensitivity needed to characterize the complex changes in spectral parameters that occur between muscle sites as well as between successive bursts of the EMG signal. It is clear from these preliminary results that previous methods used to characterize the behavior of the median frequency during sustained isometric contractions are not appropriate for dynamic activities such as cyclical lifting. This conclusion applies not only to the methods of extracting spectral estimates from nonstationary signals but also to the methods that will be needed to describe the time- and muscle-dependent changes in the instantaneous median frequency.

It may be that complex muscle interactions and fatigue-recovery cycles are the norm and that this mechanism is disturbed in the presence of LBP. Further work will be needed to support this possibility.

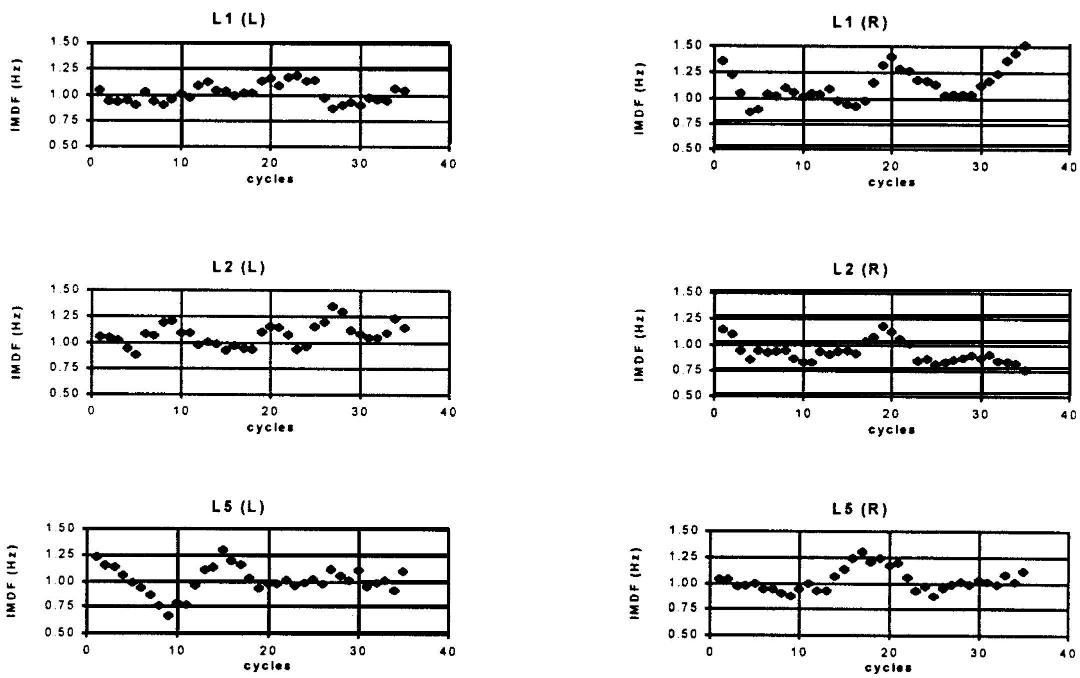

FIG 1: Concurrent changes in normalized instantaneous median frequency (IMDF) during repetitive lifting. Each data point represents the analysis of a single EMG 'burst' produced by a lift. Data are shown separately for right (R) and left (L) paraspinals at spinal lumbar levels LI, L2, and L5.

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EMG Imbalances During Experimental Low Back Pain

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INTRODUCTION

Studies of experimental muscle pain in humans commonly inject hypertonic saline into the muscle to induce pain (e.g. Graven-Nielsen et al. 1997). However, exercise induced muscle soreness triggers similar biochemical mechanisms in the muscle as muscle strain injuries, including direct tissue damage, subsequent acute inflammatory reaction and pain (Faulkner et al. 1993), suggesting that delayed muscle soreness may be a good non-invasive model to study effects of muscle pain. The induced injury is characterized by two phases, an initial mechanical injury and a secondary biochemical injury (Faulkner et al. 1993). The mechanical injury, apparent as ultra structural damage under electron microscopy, can be substantial both within the muscle fibers as well as to the muscle fiber membrane (Friden et al. 1988). The biochemical injury phase is accompanied by an increase in plasma Creatine kinase activity. The process of fiber degeneration and regeneration processes appear to follow the same pathway, regardless of how the initial injury was caused (Faulkner et al. 1993) again supporting the use of muscle soreness to study effects of muscle pain. We have recently reported that patients with low back pain demonstrate segmental imbalances in the activation of their lumbar back muscles (Oddsson et al. 1996,1997). Similar imbalances may be present in healthy subjects but they tend to cancel out across different lumbar levels (Oddsson et al. 1996). In the current study delayed muscle soreness was used to induce muscular low back pain and to study the interaction between synergistic muscles of the lumbar back in the presence of acute pain, during a standardized isometric trunk extension task. It was hypothesized that the presence of muscular pain would be associated with a redistribution of the activation of synergistic back muscles.

METHODS

A group of twelve healthy subjects were tested on four different occasions in the Back Analysis System (BAS). The BAS consists of a postural restraint apparatus which stabilizes the pelvis and lower limbs while the subject exerts a sustained isometric trunk extension force against a harness attached to two force transducers (cf. Roy et al. 1989). Subjects were tested at different levels (2080%) of their ideal body weight (IBW). Maximal voluntary contraction (MVC) was measured on each occasion. Lumbar EMG activity was recorded with six active electrodes placed bilaterally over sites at Ll, L2 and L5 levels corresponding to the longissimus thoraces, iliocostales lumborum and multifidus muscles. EMG signals were processed to assess spectral information (median frequency - MF, and amplitude - RMS). After the initial test in the BAS, delayed muscle soreness was induced by having the subject perform 10x25 rapid unilateral trunk bendings with a 10 kg load held across the shoulders. This exercise will specifically load the contra lateral lumbar back muscles with little load placed on the ipsilateral muscles. Subjects rested for 1 min between each set. Subjects were re-tested in the BAS directly following the fatiguing exercise. Tests were repeated three and seven days after the initial test. Perceived exertion was assessed during each contraction with a Borg Scale and subjective sensation of pain was assess with a visual analog scale and a pain drawing.

RESULTS AND DISCUSSION

An example of the effects seen on the mean RMS and MF parameters for a 80% IBW contraction is shown in Figure 1 for one subject. There was a clear redistribution of activation levels between the different muscle sites immediately after the fatiguing task. The fatigued side (right Ll, L2 & L5) showed an increase whereas the non-fatigued side showed a decrease in the RMS of the EMG signal. The MF of the fatigued side showed a decrease before and after the exercise whereas there was no change for the non-fatigued side. Subjects also displayed a 15-20% decrease in MVC both immediately after the fatigue exercise which persisted on day three after the exercise. Previous studies have found that this force loss is closely related to the degree of injury present in the muscle (e.g. Faulkner et al. 1993). On day three after the exercise most subjects displayed highly localized pain on the right L2 to L5 level with reddening and edema indicating an inflammatory reaction in the muscle. There appeared to be a gradual increase in the MF of the non-fatigued side on the day three and seven. It was concluded that induced muscle soreness may be a good model to study effects of muscular pain on the interaction between synergistic muscles of the lumbar back.

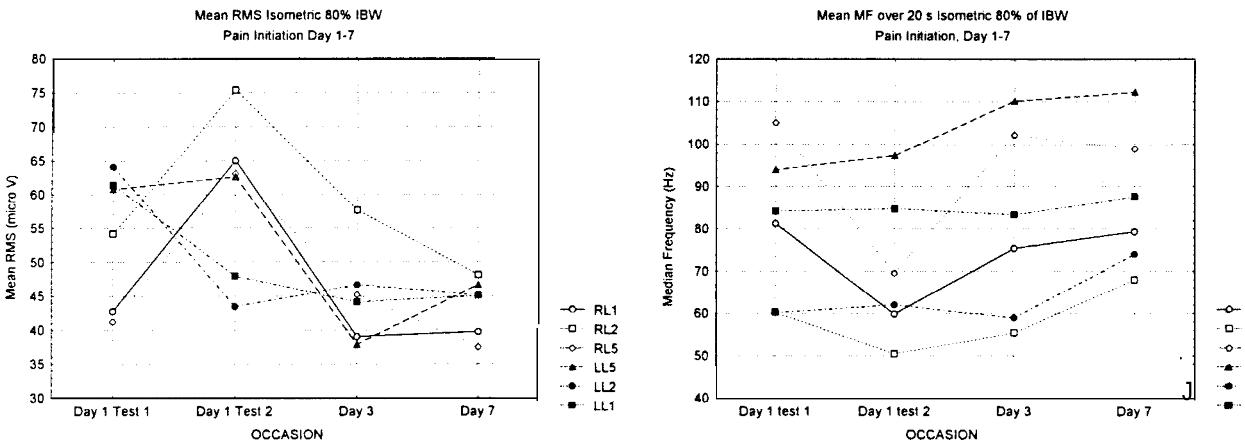

Figure 1. Mean RMS and MF for one subject over the four different test occasions.

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SMU Directed Neuromuscular Retraining of Chronic Low Back Pain

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Introduction

Through biofeedback procedures muscle activity may be either increased or decreased depending upon the desired goal. The rationale for the use of these procedures came from research studies involving the Single Motor Unit (SMU) with the pioneering works by Basmajian (1963, 1967, 1972, 1979) and others (Carlsoo & Edfeldt, 1963; Harrison & Mortensen, 1962; Scully & Basmajian, 1969; Simard & Basmajian, 1966) confirming the efficacy of SMU training in motor control. Johnson (1976) examined several aspects of this training and concluded that: a) on-off control, b) rhythm control and c) SMU isolation control (maintenance of the picket fence pattern) were the best predictors of success. Despite this success there has been no application of these training protocols to muscle interactions or muscular based chronic low back pain.

Price et al (1948) suggested that chronic low back pain was due to biomechanical factors operating upon the vertebral bodies. Several authors (i.e. Janda, 1978; Wolf & Basmajian, 1978) have agreed with Price suggesting that an imbalance of muscle activity during standing and when moving produces a biomechanical imbalance of the vertebral bodies leading to activation of the pain receptors. Surface electromyography (sEMG) techniques allow for the examination of the electrical activity of the muscles. Donaldson & Donaldson (1990) suggested, that within certain limits, muscle activity on one side of the body may be compared to that of the other side establishing which muscle is increased or conversely decreased in activity while performing movements, thus allowing for the assessment of biomechanical imbalances. The concept of imbalances has been replicated by numerous authors including Sella (1993a, 1993b) and Cassisi (1993).

Psychology has traditionally focused on the higher side reading using relaxation training techniques to reduce this activity. However longitudinal studies have indicated that these techniques lose their efficacy over time (usually due to diminished practice) (Nigel & Fischer-Williams, 1980).

The question posed is “can **muscle imbalances be corrected by uptraining those muscles which show decreased activity using SMU techniques, and if they can what is the effect upon chronic low back pain?“.**

Methods

Thirty six medically screened subjects (average age of 38 years) suffering from chronic low back (T12 - Sl) pain (average duration of pain 7.8 years) were randomly assigned to 1 of 3 treatment programs; a) biofeedback, b) relaxation training, c) education program. The biofeedback program consisted of increasing the electrical activity of the lower side using SMU protocols (holding an isometric contraction for 10 seconds, relaxing the muscle for 50 seconds - producing a picket fence pattern). The relaxation group received a form of Jacobson’s relaxation training, while the education group received education on balancing muscle activity in the back. All subjects were treated 10 times. Blinded assessments were conducted pre and post treatment and 90 days post treatment. Two measures of pain (McGill Pain Questionnaire and a VAS), 2 psychological tests (MMPI, behavior checklist), a postural evaluation, and SEMG readings in 4 positions (lying, sitting, standing and movement) were completed at each assessment. Daily recordings of pain and change in levels of pain were conducted using VAS techniques. Telephone contact was established with the subjects 4 years after the study to determine the long terms effects.

Results

The results showed the muscle imbalance was quickly corrected (within 3 sessions) for the biofeedback group. This corresponded to a significant decrease in pain for this group which was maintained in the 4 year follow up. The relaxation group did not show significant changes, while the education group showed changes after 90 days which were partially maintained 4 years later.

Discussion

The rapidity of the improvement and changes in the muscle imbalance suggest a neural involvement in the changes although alterations of joint positions could not be ruled out. Possible neural mechanisms are proposed. The use of SMU training principles appears to be a useful method for directing neuromuscular retraining and correcting muscle imbalances associated with chronic low back pain. Limitations of the study conclude the paper.

Development of EMG Parameters for Identifying Temporal Excitation Patterns of the Trunk Muscles in Low-Back Pain Patients

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INTRODUCTION

EMG signal has been used to represent a different activation pattern of injured muscle during ballistic or controlled trunk movement (Ahern et al, 1989; Pacquet et al, 1994; Roy et al, 1995). Studies have shown that EMG signal from the back of LBP patients with neurologic problems resulted in a phasic abnormality (Knutsson and Richards, 1979; Woltering et al, 1979). On the other hand, the EMG pattern between flexors and extensors of the back and hip muscle in healthy subjects showed a good reciprocal excitation pattern (Oddson and Thorstensson, 1987). These studies collectively demonstrated that there could be a different temporal pattern of EMG signals between LBP patients and healthy subjects. Yet, no quantitative technique has been developed for measuring such temporal excitation patterns of the trunk muscles.

METHOD

Subjects: Ten healthy subjects (five male, five female) with no history of low-back disorders and ten patients (six male, four female) with on-going LBP were recruited. In order to minimize the variability due to age, 18 to 50 years old male and 18 to 40 years old female were recruited (Marras et al, 1995). The average weight and height are 67.4 kg and 169 cm for healthy subjects, and 79.1 kg and 171 cm for patients respectively.
Apparatus: Lumbar motion monitor (LMM) was used to record the flexion/extension motion of the trunk and hip. Noraxon 2000 EMG system with surface electrode was used to collect the EMG signals from eight different muscles: erector Spinae (ES) at L5 level, rectus abdominis (RA), external oblique (EO), internal oblique (IO), rectus femoris (quadriceps:QUD), biceps femoris (hamstringsHAM), tibialis anterior (TA), and gastrocnemius (GAS) on the right side only.
Experirnental design: A 'group' including healthy subjects and LBP patients was used as an independent variable. The temporal differences of peak EMG between flexors and extensors at the back (RA-ES), hip (QUD-HAM), and the knee (TA-GAS) were used as dependent variables. Also, the duration of coexcitation between flexors and extensors such as RA-ES, QUD-HAM, TA-GAS, EO-IO was used as dependent variables. Graphical definition of dependent variables are shown in Figure1.
Procedure: EMG surface electrodes were applied to the eight muscle sites. LMM was fitted with subjects and a brief instruction was given to perform a free dynamicflexion/extension motion for ten seconds.
Normalization: A normalization of EMG was performed based upon a maximum filtered IEMG value in each muscle within individual subject. Thus, no additional MVC value was measured. This normalization technique was able to reduce inter-subject variability of the EMG signal.

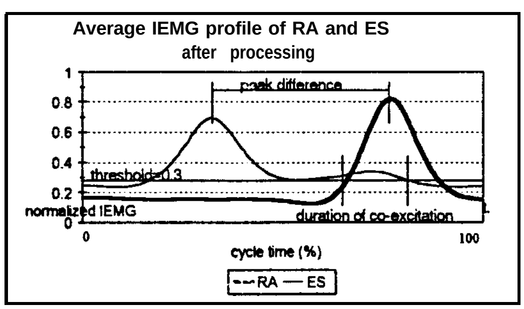

Figure 1. Average EMG profile of flexor and extensor muscles during a single cycle

Definition of active muscle: Active stage of muscle was defined as the duration when the amplitude of IEMG was 30 percent and above in a normalized scale. In fact, the separation between healthy subjects and LBP patients could be observed between 10 and 40 percent of threshold value in this study.

RESULTS

Table 1. ANOVA table

dependent variables	source	DF	F	Pr>F
peak diff. RA-ES	group	1	4.54	0.4720
peak diff. QUD-HAM	**group**	**1**	**18.76**	**0.0004*** **
peak diff. TA-GAS	group	1	0.01	0.9162
coexcitation RA-ES	**group**	**1**	**4.60**	**0.0458** **
coexcitation QUD-HAM	group	1	0.64	0.4358
coexcitation TA-GAS	group	1	0.29	0.5989
coexcitation EO-IO	**group**	**1**	**3.61**	**0.0734***

*significant at $p<0.1$, **significant at $p<0.05$ ***significant at $p<0.01$

Peak timing differences of the hip flexor-extensor pairs of muscle and the duration of coexcitation of trunk muscle pairs showed a significant difference between healthy subjects and LBP patients. This showed that the parameterized EMG variables in this study were very effective in identifying LBP patients from healthy subjects.

DISCUSSION

The peak timing difference of hip flexor/extensor muscle was found to be the most effective parameter in identifying LBP patients in this study. It was interesting to see the significant change of hip muscle activity pattern compared to the back muscle activity pattern. It could be said that LBP and the hip dynamics have a notable causal relationship even though the exact neural mechanism was not yet identified. Also, LBP patients showed a longer coexcitation pattern of the back muscle during the bending movement. This could be related to hesitation or slowing down of bending motion, that resulted in a simultaneous activation of flexor and extensor muscles. This phenomenon also explained the absence of relaxation pattern during trunk flexion motion observed by Pacquet et al (1994) and Ahern et al (1990). Regarding inter-subject variability, it was the main obstacle in increasing the accuracy of diagnosis in this study. Therefore, additional efforts should be made in advancing the test protocol or data processing technique to minimize the variability in the future.

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Effects of 4-aminopyridine on Motor Evoked Potentials and EMG Interference Patterns in Spinal Cord Injured Patients

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INTRODUCTION

4-aminopyridine (4-AP) is a K^+ channel blocking drug that has been shown capable of reversing conduction block due to demyelination in axons of the central nervous system. 4-AP is now undergoing clinical trials for its therapeutic efficacy in multiple sclerosis and spinal cord injury (SCI).[1,2,3,4]

The present open label study tested the hypothesis that 4-AP enhances maximal voluntary EMG interference patterns in paretic muscles of patients with chronic incomplete SCI. Central motor conduction deficits were also examined and those results have been reported elsewhere.[3]

METHODS

Subjects

Twenty (n=20) patients with stable SCI participated as experimental subjects and 13 healthy able-bodied adults served as controls for the determination of normal height-adjusted cen tral conduction time values.

Table 1. Clinical Characteristics of SCI Subjects

Sex			**Lesion Level**						**Frankel**				
M	F	Age(yrs)	C5	C6	C7	T7	T8	T12	A	B	C	D	Years since iniurv
14	6	37 ± 10.9	5	7	4	1	1	2	4	1	6	9	65(1-28yrs)

Protocol

SCI subjects were administered 10 mg oral 4-AP (gelatin capsule with lactose). Their maximal voluntary EMG interference patterns were assessed prior to the drug administration and at 2 hours and 4 hours post drug. Some subjects were further examined at 24 hours.

Maximal Voluntarv EMG

Maximal voluntary EMG interference patterns were recorded bilaterally from Tibialis Anterior (TA), Lateral Gastrocnemius (LG) and Extensor Digitorum Brevis (EDB) muscles. Contractions were initiated following the appearance of a green light, sustained for 3 secs where possible, and terminated after a red light. The myoelectric signals were full wave rectified and the average EMG (AEMG = $\int |\text{EMG}|\, dt/T$) calculated using 2 different methods (a) visual detection of onset and offset (b) integration over a fixed time interval.

RESULTS

Ten mg 4-AP caused significant ($p < .05$) increases in the amplitude of lower limb MEPs, and significant decreases in CMCT, MEP_{lat}, cortical stimulation threshold.[3]

Voluntary EMG patterns were recordable from 12/20 subjects with incomplete SCI. The effects of 4-AP on LG AEMG are illustrated in Fig. 1 and the results obtained from all muscles are summarized in Table 2. All 12 subjects exhibited increased AEMG (Signtest:$p < .05$) and this was evident in the mean values of all 6 muscles examined. The same results were obtained independent of the method of analysis.

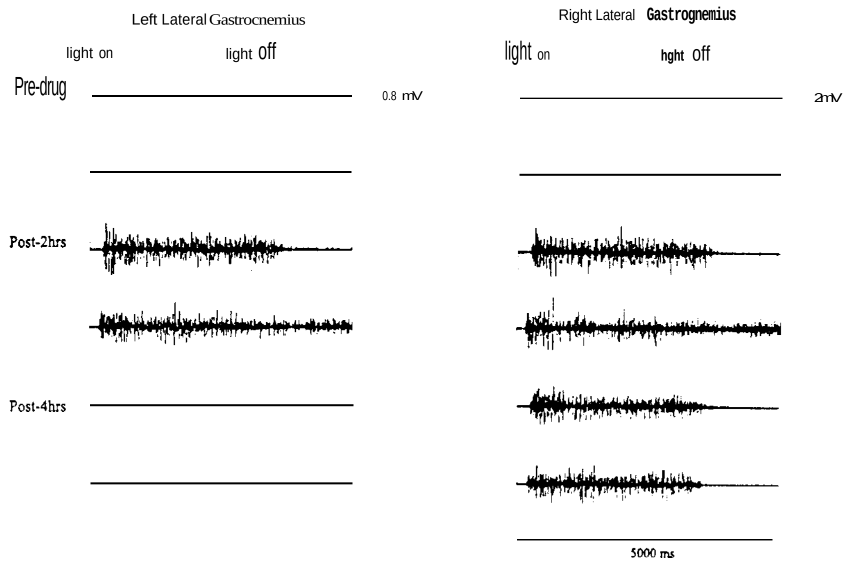

Fig. **1** Effects of 4-AP on **maximal voluntary EMG**

Table 2. Effects of 4-AP on MVC Average EMG Interference Patterns

Muscles	n	Pre-drug	Post-2 **hrs**	**Post-4 hrs**
Left Tibialis Anterior	8	0.11 [O.ll]	0.14 [O. 161	0.13 [O. 15]
Right Tibialis Anterior	9	**0.12 [0.07]**	**0.14 [0.08]**	**0.12 [0.07]**
Left Lateral Gastrocnemius	8	**0.06 [0.03]**	**0.07 [0.03]**	**0.07 [0.04]**
Right Lateral Gastrocnemius	7	**0.07 [0.04]**	**0.10 [0.02]**	**0.09 [0.02]**

[ ] Standard Deviation units = mV

CONCLUSIONS

This study demonstrated that 4-AP enhances central motor conduction in SCI patients and this is associated with increased voluntary EMG activity. These observations assist in understanding the physiological basis for the reported therapeutic effects of 4-AP. The K^+ channel blocking action of the drug most likely leads to enhanced conduction through demyelinated internodes, enhanced neural transmission, and may cause an increase in extracellular K^+ in the motor cortex leading to heightened excitability of pyramidal tract neurons.

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Surface Electromyography: An Early Indicator of Recovery After Facial Reanimation Surgery

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INTRODUCTION: Facial nerve grafting and free muscle transplantation (reanimation surgeries) are acceptable methods for dealing with the often devastating condition of facial paralysis. Recovery after reanimation surgery is a prolonged process. Patients are always looking for some sign of recovery and are at times frustrated by the long waiting period. Several studies have reported the outcomes of facial nerve grafting.[1-3] The outcome measures have typically been designated as poor, fair, good or excellent and in a few cases a categorical scale, the House-Brackmann rating of facial nerve disorders, has been used.[1-2] Less common are studies tracing the recovery process using needle EMG.[3] Surface electromyography (sEMG) is a noninvasive technique which can be used to monitor changes in muscle activity. The process of recovery can be described in terms of changes in facial impairment, but also in terms of the patients experience (facial disability). We define the recovery after facial reanimation surgery, using specific clinical measures of facial impairment and an index of psychosocial disability related to facial nerve disorders. The purpose of this study is to describe the natural process of recovery after facial reanimation surgery using an observer based rating scale of facial movement and posture, surface electromyography (sEMG) recordings of facial muscle activity and self report of psychosocial disability.

METHODS: Eight adult outpatients who underwent a facial reanimation surgery to address unilateral facial paralysis were studied (mean age = 50.6 SD=15.4). The most common cause of facial paralysis was tumors of the cervicofacial region (n= 4) followed by acoustic neuroma surgery (n=3) and Bell's palsy (n=l). The duration of paralysis prior to reanimation surgery ranged from 0 to 96 months (mean duration = 16.4 months SD=32.7). The facial reanimation surgeries included: primary facial nerve grafting (n=4), hypoglossal nerve to facial nerve anastomosis (XII-VII anastomosis) (n=l), and cross facial nerve grafting and XII-VII anastomosis (n=3). One of the patients with cross facial nerve grafting and XII-VII anastomosis also had a free muscle transfer. All subjects were referred for physical therapy evaluation at the Facial Nerve Center after their surgical procedure. Patients were evaluated using the Facial Grading System (FGS), an observer based scale of resting posture, voluntary movement and synkinesis, SEMG bipolar recordings of facial muscle activity and self-report of psychosocial function, the Facial Disability Index social well-being subscale (FDI social). Surface EMG recordings were primarily collected from the zygomaticus muscle group during smile. Surface EMG quantification of muscle activity using the NeuroEducator II (Therapeutic Alliances, Inc, Fairborn, OH) involves full wave rectification, real time processing and a 100 msec time constant of integration. Pregelled Ag-AgCl electrodes, 1 cm in diameter, were used to record from the muscles. Standardized electrode placement was used for all of the measurements. Evaluations were conducted at regular follow-up visits over at least a 12 month time period. Descriptive statistics were calculated to describe the interval of time from surgery to change in selected outcome measures, FGS rating of movement in smile, SEMG muscle activity of the zygomaticus muscle group and the FDI social. Rank order of the initial recovery of the outcome measures was determined for each patient.

RESULTS: The first indicator of recovery of facial muscle function was the SEMG activity recorded from the zygomaticus muscle group during smile (n=5, Table). Improvement in SEMG of zygomaticus muscle activity occurred on average at 5.2 months (SD=4.2). Changes in muscle activity proceeded the changes in observable voluntary smile as determined by the FGS movement ratings, by an average of two months. With a few exceptions, resting facial posture as measured by the FGS resting posture subscale, was the last measure to demonstrate change. The data for the FDI social was limited, but for subjects who reported social disability the score was one of the earlier measures to change (Table). The group mean for time until change (months) for each of the measures indicated the following order of recovery: zygomaticus muscle activity with smile, observer based rating of smile, social disability and resting posture (Table).

Table Rank Order of the Onset of Recovery in Outcome Measures

Subject	Zygomaticus Muscle Activity, SEMG; n=8	FGS Movement Rating for Smile; n=8	FGS Rating of Resting Posture; n=8	FDI Social; n=4
1	1	2	3	
2	1	2	3	1
3	1	2	2	2
4	1	3	3	2
5	1	2	1	
6	2	1	3	
7	1	1	1	2
8	2*	2	1	
Mean (SD)	5.2 (4.20)	7.4 (4.10)	9.0 (3.82)	3 (7.41)

* Change in muscle activity for levator labii muscle group proceeded all measures.

DISCUSSION: For the patients studied, SEMG recordings of facial muscle activity were an early indicator of recovery following facial reanimation surgery. The time frame of change in muscle activity as recorded by SEMG appears similar to that identified by needle EMG.[3] Changes in the zygomaticus muscle group activity usually occurred prior to observable facial movement in smile. The changes in muscle activity provided the patient with an early sign of their facial nerve and muscle function well before observable signs of facial motion. The early information may be an encouraging factor for the patient during what is often, long and psychologically demanding recovery process. Interestingly, for the patients in which we had completed self-report measures of social disability, changes in social-well being either occurred with the changes in muscle activity or as the next indicator of change.

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The Dynamic Response to Unloading in the Hemiplegic Arm is Related to Functional Impairment

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INTRODUCTION

Deficits in the control or coordination of movement following central nervous system injury or disease may result from altered descending commands and segmental reflex mechanisms. At the segmental level, magnitudes of H- and stretch reflex responses are significantly increased and vibratory inhibition of the H reflex is depressed in spasticity (Ashby and Verrier 1976; Levin and Hui-Chan 1992) While muscle stretch allows us to study the effects of excitation of muscle afferents, unloading of a tonically active muscle (the unloading reflex), examines the response to deactivation of these afferents. When a steadily activated muscle is suddenly unloaded, the arm moves to a new position at which the residual load is balanced. The normal phasic reaction is a silent period, followed by a large burst or rebound of activity in the same muscle (Angel 1973). The phasic responses are followed by a decreased level of tonic EMG related to the magnitude of the residual load (Feldman 1986). At the same time, due to the lengthening of the antagonist muscles during unloading, a stretch reflex response is evoked in these muscles. It has been hypothesized that the silent period results from the cessation of the excitatory drive from Ia spindle afferents resulting in a reduction of muscle action potentials when the muscle is suddenly shortened (Struppler et al 1973; Burke et al. 1978). Mechanisms to explain the termination of the silent period are less clear. In spastic hemiplegic subjects, the latency of the unloading response is reportedly normal but the response is prolonged and the rebound or after-volley is markedly reduced in amplitude compared to normal (Angel 1973). The goal of this study was to investigate the response to sudden partial and complete unloading in the arms of spastic hemiparetic subjects in order to gain further insight into how segmental mechanisms may contribute to motor deficits in these subjects.

METHODS

Unloading reflexes in the elbow flexor muscles were evoked by the sudden removal of a constant load from the pre-activated flexors in 6 healthy and 10 hemiparetic subjects. The initial load (20-30% MVC) was decreased in a step-like manner resulting in residual loads varying from trial to trial (10 trials for each of S-6 levels of the residual load).The procedure was repeated from two starting positions of the elbow (130’ and 100'; i.e. when initial muscle length was changed). EMG activity from two elbow flexors (biceps and brachioradialis) and two elbow extensors (long head of triceps brachii and anconeus), as well as torque, velocity and joint position were recorded. In each session, the level of clinical spasticity and residual motor function in the affected limb of the hemiparetic subjects was also evaluated. The latency of the silent period in biceps or brachioradialis EMG activity was calculated as the time between the beginning of the unloading (as seen from the torque trace) to the beginning of the EMG decline. The offset of the silent period was defined as the time at which the EMG signal surpassed and was sustained at more than two standard deviations (SD) of the baseline activity.

RESULTS

The latency of the SP did not vary with initial muscle length and was not significantly diierent in the two subject groups (27.8 ± 5.2 ms in healthy compared to 32.8 ± 5.3 ms in stroke subjects for unloading performed from the initial arm position of 100°). However, the SP was significantly prolonged and the after-volley was decreased in magnitude in stroke compared to healthy subjects. For all levels of unloading, the overall SP duration was 148.3 ± 75.6 rns in stroke compared to 57.5 ± 16.4 ms in healthy subjects. Neither the initial muscle length nor the amount of partial unloading had an effect on the duration of the SP in healthy subjects. However, SP duration increased from 68.5 ± 23.4 ms to 212.5 ± 57.8 rns with increasing amounts of unloading (from 100") in stroke subjects. This effect was most apparent in patients with moderate to severe motor symptoms. There were no significant differences between the SP durations measured from the two initial muscle lengths whereas the correlation between the amount of unloading and duration of the SP was present at both initial muscle lengths. The peak velocities of unloading varied with the amount of unloading in both subject groups. It ranged from 43.4 to 259.8 deg/s in the healthy subjects and from 32.5 to 153.8 deg/s in the hemiparetic subjects for decreases in load ranging from 2 to 12 Nm. The differences in velocity of unloading were significant between the subject groups only for decreases in load of 2 and 6 Nm. A subgroup of 4 stroke subjects was identified in whom SP durations, although prolonged, did not vary with the amount of unloading. These subjects had almost full functional recovery of the upper limb as measured by the Fugl-Meyer Upper Limb Motor Assessment Scale (62-66 / 66). In the remaining subjects, there was a significant correlation between the duration but not the onset of the silent period and level of clinical spasticity.

DISCUSSION

Silent periods were prolonged in stroke compared to healthy subjects despite the lower peak velocities of unloading in stroke subjects. In hemiparetic but not healthy subjects, the duration of the SP varied with the velocity of shortening. In hemiparetic subjects, higher velocities and magnitudes of unloading resulted in longer silent periods. One explanation for the abnormal prolongation of the SP may be an increased dynamic sensitivity of the stretch reflex. Our data also suggests that the process of motor recovery following stroke may be associated with a decrease in dynamic stretch reflex hyperexcitability.

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Factors Affecting the Contribution From the Lower Limbs During Seated Reaching Tasks in Healthy and Stroke Subjects

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INTRODUCTION The purpose of this series of studies was to develop and test the efficacy of a task-related training program aimed at reducing the disability associated with poor sitting balance following stroke. Three studies were undertaken, the first two sought to determine the biomechanical characteristics of performance of seated reaching tasks in healthy subjects. The findings from these studies were integrated with existing knowledge to develop a 2 week intensive training program designed to improve sitting balance and maximum distance reached following stroke. Efficacy of this program was tested in the third study using a randomised placebo-controlled design.

METHODS In the first two studies, hand movement time, vertical ground reaction force (VGRF) through the feet and activity in leg muscles were recorded as healthy subjects performed self-paced reaching tasks under varied conditions. In the first study, three reaching distances (60%, lOO%, 140% arm's length) and two tasks (reaching to grasp a glass, and reaching to press a switch) were examined. In the second study, three reach directions (ipsilateral: 45 deg to the right, forward and across: 45 deg to the left) and three thigh support conditions (25 %, 55 %, 85 % thigh length) were examined as subjects reached over a standardised distance (140% arm's length). Planned comparison ANOVAs were used to analyse force and hand movement time data.

In the third study, twenty subjects at least one year after stroke were randomly assigned to an experimental or control group. Both groups participated in standardised training programs for two weeks. The experimental group had training designed to improve their sitting balance which involved emphasis on loading the affected leg while practising reaching beyond arm's length. The control group had sham training. In pre and post-tests, sitting balance was measured by examining the hand movement time, VGRF through each foot and EMG activity in VL, SOL,TA bilaterally as subjects reached with the unaffected hand to pick up and drink water from a glass under the same three reach direction conditions used in Study 2. In addition, the maximum distance reached in each direction was measured. Change scores from pre to post test in peak vertical force through each foot, hand movement time and reaching distance were analysed using T-tests. EMG data in all studies were analysed descriptively.

RESULTS It was found that reaching beyond arm's length required not only coordinated motion of the trunk and arm segments but also active contribution of lower limbs to support, balance and propel the body mass. Peak vertical ground reaction forces through the feet increased as reaching distance increased ($p<O.Ol$). There was no significant differences due to task. Peak VGRF through each foot was significantly affected by reach

direction ($p<0.01$), with the maximum loading on the right foot arising from reaches to the right, and the maximum loading on the left foot arising from right hand reaches across the body. Peak vertical GRF was also significantly affected by the extent of thigh support ($p<0.01$), with peak GRF increasing as the extent of thigh support decreased. In the first study, at the short distance (60% arm length) none of the leg muscles monitored were activated while at the long distance (140% arm length) at least one leg muscle was activated in all trials. The muscles most commonly activated were tibialis anterior (TA) and soleus (SOL). In the second study, all subjects activated TA, SOL and vastus lateralis (VL) bilaterally in all trials in all conditions. In the third study, following training, experimental subjects were able to reach faster and further, increase the load through the affected foot and increase the number of trials in which affected leg muscles were activated compared with the control group ($p<0.01$).

CONCLUSIONS The results provide strong evidence of the efficacy of task-related training in improving the ability to balance in sitting after stroke. The challenge for stroke rehabilitation is to take advantage of the potential for neural plasticity by providing appropriate intervention to drive and shape the reorganisation in order to maximise recovery. This research highlights the value of developing and testing scientifically based rehabilitation procedures.

Evaluation of Postural Muscle Responses in Spinal Cord Injured People Using Different Chair Configurations

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INTRODUCTION

A controllable sitting balance is very important for the optimal performance of activities of daily living. This importance is reflected in the amount of therapy spent on training sitting balance in an early stage of the rehabilitation process of people with a spinal cord injury (SCI). In SCI people impairments of the sensorimotor system not only lead to a diminished balance control but also to a poor task performance. Adaptations in seating may add to sitting stability and may thus improve task performance in these people. In able-bodied subjects the effect of chair configuration on sitting comfort, myoelectric back muscle activity, lumbar disc pressure and task performance has been studied extensively, e.g. [1,4]. Little is known, however, about the relationship between sitting balance and the ability to perform ADL or labour in people with a disability [2,3]. This is especially true in people with a SCI who are confined to their wheelchair most of the day. Aim of this research is to study balance control and task performance in different chair configurations in both SCI and non SCI subjects.

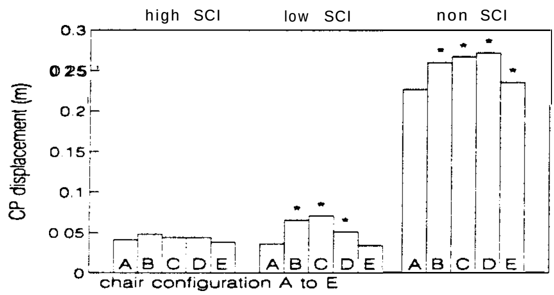
METHODS

The study included ten people with a complete high thoracic SCI (T2-T8) and ten people with a complete low thoracic SCI (T9-T12). All subjects had finished their active rehabilitation process at least six months before and had no secondary pathology. The SCI people were matched with a group of ten able-bodied controls. The three groups were matched for sex, age, height, weight and educational level. Sitting balance was systematically perturbed using a bimanual reaching task in anterior direction, presented as a visual reaction time task. The forward displacement of the upper limbs constituted a change in sitting balance which had to be accompanied by postural adjustments affecting the rest of the body. Muscle activity was recorded bilaterally from the erector spinae (ES) at level L3, T9 and T3, latissimus dorsi (LD), ascending part of the trapezius muscle (TPA), Serratus anterior (SA), sternocostal head of the pectoralis major (PM) and the oblique abdominal muscles (OA) by means of surface electromyography. Ag-AgCl electrodes with a contact surface of 1 cm^2 were used. Preamplifiers were directly mounted on top of the disposable electrodes (CMRR > 110dB; input impedance > $500 M\Omega$; signal amplification 100; noise referred to input < 2 μV_{rms}). Inter-electrode distance was 2.3 cm. An amplifier with a 3rd order Butterworth high pass filter (cut-off frequency 20 Hz) was used. A sample rate of 500 Hz was used as spinal and other trunk muscles have frequency components up to approximately 250 Hz. The skin was cleaned with alcohol and excess hair was removed if applicable. The electrode arrangement was in the direction of the muscle fibres. A ground electrode was placed in the anterior region of the lower leg. Sitting balance was monitored by measuring the changes in the location of the centre of pressure (CP) using a AMTI force platform mounted underneath a multi-adaptable chair. Five different chair configurations were used in this study, viz.: A) 0” tilting and 10” reclining (standard chair), B) 7” tilting and17° reclining, C) 12” tilting and 22” reclining, D) 0” tilting and 22” reclining and E) a configuration similar to A but fitted with a ROHO cushion instead of a standard foam cushion. Maximum CP displacements (CPmax) and mean rectified EMG values were calculated per group and per reaching distance in a period of submaximal balance perturbation.

RESULTS

CPmax values for all chair configurations for ail groups are depicted in figure 1. The range in which SCI people could move their CP during task execution was significantly smaller compared to able-bodied subjects in all five chair configurations. The standard chair (A) was the only chair configuration in which no significant difference in CP displacement was present between the high and low SCI subjects. Those chair configurations that led to a significant difference in CPmax compared to chair A, within one group, are marked with an *. EMG data revealed significant differences in only a few cases when comparing the standard chair with the other four chair configurations, within one group. High SCI subjects showed less activity of the LD in chair B and less activity of the LD, ESL3 and OA in chair C and D. Manipulation of the seat cushion (chair E) did not lead to a difference in muscle activity in this group. The low SCI subjects showed a difference in muscle activity in chair D and E. In chair D they revealed less activity of the OA whereas in chair E they revealed more ESL3 and EST9 actvity. In the group of able-bodied subjects significant less activity of the ESL3 was found in chair C but more activity of the SA. Chair E led to more activity of the OA, SA and LD.

Figure 1. CPmax values of the 90% reaching condition, for all chair configurations, for all groups.



DISCUSSION

Although low SCI subjects showed a significant increase in CPmax when the chair was inclined backwards 7” or 12” or when the backrest of the chair was reclined 22°, this did not lead to notable differences in muscle activity, in the 90% reaching condition. In other words the extent to which sitting balance could stably be maintained increased without a significant increase in muscle activity. Similar conclusions can be drawn for the able-bodied subjects, except for the standard chair fitted with the ROHO cushion. In the latter chair able-bodied subjects revealed more activity of the OA, SA and LD. Most differences in muscle activity due to different chair configurations were present in the high SCI group. Inclining the chair backwards 12” degrees or reclining the backrest **of** the chair led to significant less activity of the ESL3, OA and LD. However, the range in which the CP could activily be moved did not differ between the different chair configurations in this group.

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Neuromuscular Modifications Induced by Strength Training

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INTRODUCTION

It is generally accepted that during the first 6 weeks of strength training there is no significant change in muscle cross-sectional area although force increments are evident [3,4]. Therefore, increments in muscle strength must be mainly explained on the basis of a functional modification in the motoneuronal activity. The purpose of the present research was to study this neuromuscular adaptation. In particular, we wanted to evaluate the possible modifications of the motor unit (MU) recruitment strategy induced by the strength training. To accomplish this purpose 6 sedentary subjects performed a six week strength training protocol, working out 3 days per week. The subjects underwent evaluation tests, before, during and after the training period. During each test electromyogram (EMG), ability in increasing force from 0 to 100% maximal voluntary contraction (MVC), and both isotonic and isometric strength improvement were recorded. The median frequency (MF) of the power density spectrum (PDS) of the EMG was used as a tracking parameter to describe MU recruitment [1,2]. In fact, it has been demonstrated that MF increases linearly with orderly recruitment of the MUs and that the maximum MF is obtained when full recruitment is achieved [5]. Conversely, the MF is not significantly affected by increments in the MU firing rate, even if additional force is obtained.

METHODS

Six healthy sedentary subjects (2 male and 4 female) with an average age of 28.5 $\pm$ 5 years volunteered to participate in this study. They performed five linearly increasing isometric contractions (in a ramp fashion) from 0 to 100% maximal voluntary contraction (MVC) with the dominant arm. The ramps had to last 3 seconds and were performed following a track displayed on a computer screen. In order to quantify the subjects' ability to track the ramp, the linearity was estimated using the standard error of estimate (SEE) of the linear regression curve of the actual ramp and the slope was estimated using the regression coefficients of the linear regression curve of the current ramp compared to the ideal ramp. Surface myoelectric signals from the medial head of the biceps brachii and the lateral head of the triceps brachii were detected with two sets of bipolar silver/silver chloride electrodes. The MF, the frequency value that divides the PDS of the EMG into two regions of equal power, was calculated by means of a standard Fast Fourier Transformation. At the beginning of each test, the frequency content of the EMG signal was calculated averaging the three MF values collected during the first 3 seconds of a preliminary trial of the MVC. For each ramp, the force value at the highest MF (%MVC) was considered as full MU recruitment for the biceps and as recruitment of all the needed MUs for the triceps. Each subjects' typical full recruitment was obtained averaging the result of the five ramps. After baseline measurements, subjects performed a six week isotonic training protocol on a camm arm-curl machine (Technogym, Top-XT). Subjects trained 3 days per week starting from an intensity of 70% of the maximal load that they could lift only once (lRM), with a 5% weekly increment. Every two weeks subjects were asked to repeat their 1RM in order to adjust their training workload. Every two weeks, at the end of the second, fourth and sixth week of training, surface myolelectric signals were collected as previously. Of the 4 tests performed, we considered the baseline test and the test during which we measured the highest force increment in accordance with the MVC values obtained during each test. All data were analyzed through paired student t tests ($p<0.05$).

RESULTS

After the training period. we noted that in 5 subjects the major isometric force increment was achieved at the fourth week of training, and in the remaining subject the same increment was achieved at the sixth week. On average, compared to the initial isometric force value, this increment was of 16% with a standard deviation of 5%. The isotonic force based on the measurement of the IRM improved on average by 26.5% **(±14%)**. Four subjects obtained the highest improvement at the fourth week of training and two subjects at the sixth week.

The analysis of the EMGs and of the ramp performance at the test before training (BTT) and at the test with the highest strength improvement (HSIT) gave the following results. In the HSIT, fill MU recruitment of the biceps was found at **64±9%** MVC. This percentage was significantly lower than the value measured at the BTT **(77±6%** MVC). In the triceps the full recruitment of the needed MUs occurred on average at **67±8%** MVC in the BTT and at **75±15%** MVC in the HSIT. This difference was not significant. Training induced a statistically significant shit? towards higher frequencies in the frequency domain representation of the EMG signal of the MVC measurement. The MF changed from **81±12** Hz at the BTT to **99±13Hz** at the HSIT. The SEE and the slope of the linear regression curves of the actual ramps of the ramp showed that there was no statistical significant difference in tracking ability with training, that is there was no learning effect in the way the subjects performed the ramps.

DISCUSSION

The results of the present study demonstrate that the strength training performed was effective in improving the isometric and isotonic force of each subject. The force increments observed after training may be explained by a functional modification in the motoneuronal activity. In particular, training would result in a facilitation of the recruitment of the large motor units. The earlier recruitment of fast-twitch muscle fibers and therefore the earlier full MU recruitment would allow the MUs to reach higher values of firing rate and to synchronize. Secondly, the frequency content of the EMG also increased in four subjects. this could be explained either as the training induced ability to recruit MUs, that were not previously recruited, or as a change in the biochemical characteristics of the muscle fibers.

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Recovery of Shoulder-neck Muscles After Sustained Submaximal Effort

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INTRODUCTION

A muscle subjected to sustained exercise will be influenced by fatigue progressively until the limit of endurance. The recovery of muscular effort after an endurance task depends strongly on the preceding force production (amount of force, type of exercise: continuous or intermittent, length of the cycle time, duty-cycle, etc . ..). In previous experiments, the recovery of fairly high submaximal loads was investigated (e.g. Barnes and Williams 1987: 60% MVC, Le Bozec and Rogier 1991: 50% MVC). However, many of todays occupational tasks are characterised with very low efforts of neck and shoulder muscles. Therefore, the purpose of the present study was to analyse muscular fatigue and recovery of neck and shoulder muscles after submaximal static endurance tasks. Parameters extracted from the electromyographic signal (EMG) represent objective information concerning the amplitude or frequency content of the signal.

METHODS

Ten female subjects performed a sustained forward flexion of the right arm against a handle. The task was performed at 20% MVC till endurance. After a 10 min 'silent' recovery period (subjects remained seated, arms on the laps, no activity performance), the endurance task was repeated. Maximal voluntary contractions were asked before and after the two endurance tasks. EMG signals were obtained from the upper trapezius and anterior deltoid, RMS (root-mean-square) values and MPF (mean power frequency) values were considered as amplitude and frequency parameter respectively. The activity parameter (ACT, Spaepen et al. 1986) and the ratio RMS/ACT were also calculated with the purpose of differentiating force from fatigue changes (Hermans 1996).

RESULTS

Mean torque values of the four MVC's (MVC1, MVC2, MVC3 and MVC4) were significant different from each other ($p < 0.01$), except between MVC2 and MVC3. Regarding the EMG parameters, most significant differences were found for the deltoid between MVC1 and MVC4.

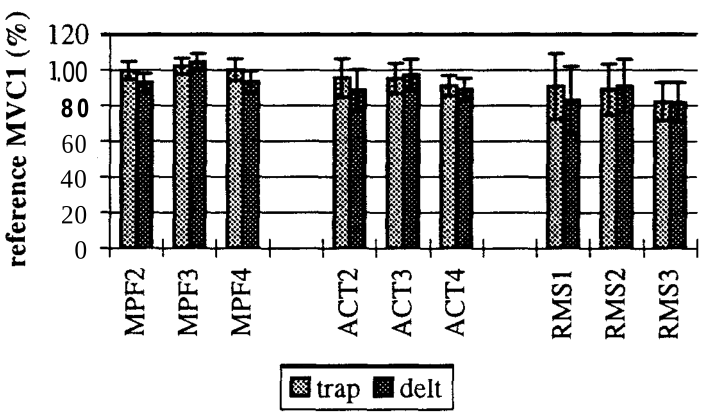

Figure 1: Changes in EMG parameters (MPF, ACT and RMS) of MVC2, MVC3 and MVC4, relative to the values of MVC1 for the trapezius (trap) and deltoideus (delt) muscle.

In both endurance tasks, the increases in RMS and ACT and the decrease in MPF were significant for both muscles ($p > 0.01$). Regression coefficients were not different between the two tasks, except a larger trapezius ACT increase ($p < 0.05$) and larger MPF decrease (not significant) in the second test.

		MPF	ACT	RMS
test1	delt	-0.059** (0.005)	-0.017** (0.006)	0.068** (0.014)
	trap	-0.019** (0.006)	0.056** (0.005)	0.073** (0.018)
test2	delt	-0.061** (0.004)	-0.014** (0.005)	0.059** (0.012)
	trap	-0.023** (0.005)	0.073** (0.006)	0.052** (0.014)

Table 1. Mean slope constant. expressed in relative change per minute, for mean power frequency (MPF), activity (ACT) and root mean square (RMS), after regression analysis of individual normalised values over the endurance time. The results for deltoid (delt) and trapezius (trap) are given for both tests (testl, test2) and probability levels are presented by **:P<0.01.

DISCUSSION

The difference in EMG recovery of trapezius and deltoid is due to the different muscle function. Because of the relatively low stabilising effort of the trapezius, it is not expected to be affected to the same extent to fatigue as the deltoid, the prime mover of a forward flexion. However, to delay the fatigue feeling, the deltoid transfers torque to other muscles: an increased appeal on the trapezius is made. The small ACT decrease for the deltoid supports this: subjects have difficulties in maintaining the initial force due to an increase of physiological tremor (Garnet and Maton 1989). Because a 10 min recovery seems not to be enough to completely relax the muscle (the initial values were not reachted), an increased appeal on the trapezius is made during the second endurance task.

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Muscle Activity Patterns During Computer Mouse Work

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INTRODUCTION. High prevalences of musculoskeletal symptoms among computer workers may be associated with specific work tasks or ergonomic and psychosocial factors at the work place. In a questionnaire survey involving 149 computer-aided design (CAD) operators musculoskeletal symptoms in the shoulder and upper limb in the side operating the computer mouse were reported more frequently compared to the Contralateral side (1). Surface electromyographic (EMG) recordings on the upper trapezius muscles from a subsample of these CAD operators showed that the levels of activity on both shoulder muscles were low (mean median levels: 4.6% vs. 2.7% EMG,, (1)) and it seemed unlikely that the significant, but only slightly higher activity level of the upper trapezius on the mouse side was an important risk indicator for the development of muscular symptoms. Therefore, it was investigated whether larger differences in the activity patterns were present expressed as the numbers of “silent” periods (EMG gaps) and as the patterns of repetitiveness. Such analyses were also carried out on recordings from the extensor digitorum communis muscle on the mouse side to investigate whether the activity patterns of shoulder and forearm muscles on the mouse side were similar.

METHODS. Bipolar surface electrodes (Ag-AgCl) were used for EMG recordings for 20-25 min of the m. trapezius descendens on the mouse side (n=14) and on the non-mouse side (n=7), and of m. extensor digitorum communis (n=20) on the mouse side. Interelectrode distance was 20 mm. The EMG signal was amplified, low-pass filtered (cut-off 400 Hz), sampled on a computer at a frequency of 1024 Hz and high pass filtered (cut-off 10 Hz). The signals were full-wave rectified and RMS converted within windows of 100 ms duration. The resting signal level was quadratically subtracted from the EMG signal. The RMS amplitude was normalised in relation to the RMS amplitude recorded during maximal isometric contractions (EMG,,). The RMS-converted EMG values were analysed according to the gap analysis which quantifies the number and duration of “silent” periods (level below 0.5% EMG,, for at least 0.2 s, (2)) and according to the exposure variation analysis (EVA) which quantifies the EMG signal both in the level and time domain (3). The Mann-Whitney test was used to test for differences at a significance level of $p<0.05$.

RESULTS. The number of EMG gaps and the total gap time of the upper trapezius on the mouse side were significantly lower than the values for the other upper trapezius, but not significantly different from the values for the extensor digitorum communis (table 1).

Table 1. Median values of no. of EMG gaps and total gap time (25-75 nercentiles).

	gaps (min^{-1})	gan time (% time)
Upper trap, mouse side	1.6 (0.0-6.1)	2.9 (0.0-5.3)
Upper trap, other side	22.1 (2.4-43.1)	17.3 (6.2-32.8)
Ext digitorum communis	0.8 (O.O- 1 .0)	0.7 (O.O- 1.8)

The EVA analyses showed that the recorded activity typically stayed within the same level interval for 1-3 s (fig. 1). However, the peak for the l-3 s period length was significantly higher for the upper trapezius muscle on the mouse side (31% of the time) than for the other side (24% of the time). The tendency of activity period lengths of l-3 s was even more prononunced for the extensor digitorum communis (36% of the time).

DISCUSSION. The gap analyses indicated that more continous muscle activity was present in the upper trapezius muscle on the mouse side than on the other side. The differences in the number and duration of periods with no or very little muscle activity seemed much larger than the previously reported differences in activity <u>levels</u>. The relevance of registering the duration of periods with no muscular activity relies on the hypothesis that continous activity of even a small number of motor units may contribute to the development of pain (4). Furthermore, the EVA analyses showed that the activity pattern of the upper trapezius on the mouse side was less variable, i.e. the activity typically stayed within given level intervals for l-3 s and less often for shorter or longer period lengths. This repetitiveness in period length was also present in the muscle activity pattern of the forearm muscle during computer mouse work.

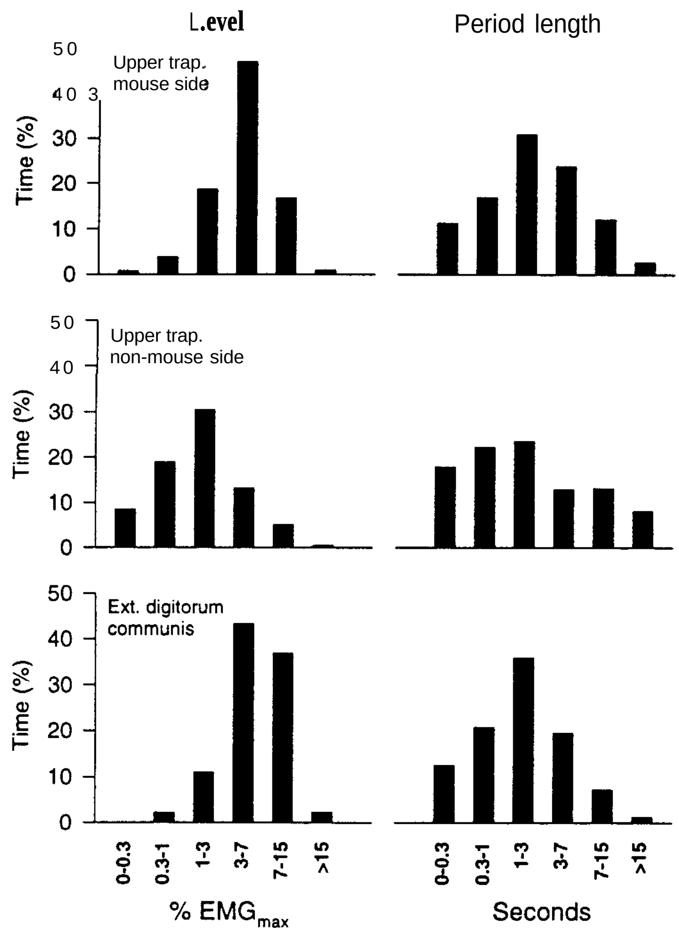

Figure 1. EVA analyses of the RMS values. Median values of duration within specific level and period length intervals are shown.

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Fatigue-Induced Changes in Biomechanics and Muscular Activity during Manual Wheelchair Propulsion

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INTRODUCTION

Overuse injuries often limit the independence of long-term manual wheelchair users. Factors predisposing musculoskeletal structures to these injuries include the imbalance of strength and flexibility that can result from adaptation of muscles to constant use, and fatigue from the need to use arms for locomotion in addition to other activities of daily living. The purpose of this study was to identify changes in muscle activity, joint kinematics and kinetics and wheelchair handrim kinetics that result from wheelchair propulsion to fatigue as a means of understanding risk factors for overuse injuries among manual wheelchair users.

METHODS

After medical screening, 20 long-term manual wheelchair users without upper extremity involvement participated in the study. Length of time in a wheelchair ranged from 3 to 38 years (16.4 $\pm$ 10.3), and age ranged from 21 to 68 years (43.7$\pm$ 11.1 yrs). Wheelchair propulsion mechanics were measured during a submaximal exercise test to fatigue which was terminated when velocity could not be maintained at 3 km/hr. Load for the fatigue test was that which corresponded to 75% of the peak V02 that occurred during a graded maximal exercise test on the wheelchair ergometer.

The wheelchair measurement system included a prototype wheelchair ergometer, three Peak 3D CCD cameras (Peak Performance Technologies, Colorado Springs, CO), a VCR/monitor assembly, an image processing unit, a 3D force/torque transducer, a potentiometer, an amplifier, an analog-to-digital unit, and a PC. The wheelchair ergometer was instrumented with a PY6-4 six-component force/torque transducer (Bertec Corp., Worthington, OH) in its wheel hub which uses bonded strain gauges to measure three dimensional (3-D) handrim forces and moments. A potentiometer monitored the angular position of the wheel. For each subject, reflective markers were placed at the fifth metacarpal head, styloid process, lateral epicondyle of the humerus, acromion, and greater trochanter. Data collected during the fatigue test included 3-D motion analysis data (sampled at 60 Hz) and 3-D handrim contact forces and moments (sampled at 360 Hz). Joint kinetics were calculated using a three-dimensional inverse dynamic model of wheelchair propulsion that employed the Newton-Euler method based on body coordinate systems. This model assumed the arm to be three rigid segments (hand, forearm and upper arm) connected by the wrist, elbow and shoulder joints.

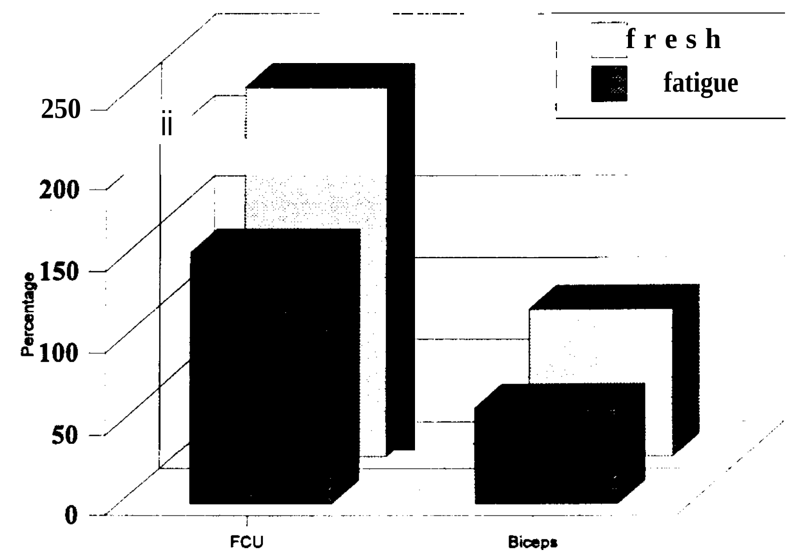
Electromyographic (EMG) data was collected from an eight-channel telemetry system at 960 Hz (Noraxon USA Inc, Scottsdale, AZ). Pairs of surface electrodes were placed on the subject's right side over motor points of the pectoralis major, anterior and posterior deltoid, middle trapezius, biceps and triceps brachii, flexor carpi ulnaris and extensor carpi radialis muscles. EMG signals were processed using the RMS method then normalized to maximal isometric contraction activity. Peak magnitude was determined from the average of three propulsion cycles for each trial. Timing of muscle activity and integrated EMG were determined from a representative cycle for each trial and related to propulsion and recovery phases.

Kinetic and kinematic data were averaged over three cycles (contact to contact) for each condition (fresh and fatigued). A one way repeated measures ANOVA was used to evaluate effects of fatigue state (fresh/fatigued) with a significance level of 0.05.

RESULTS

Significant changes resulting from fatigue were observed in most measurement categories. Temporally, stroke frequency increased 6.5% with fatigue. Joint kinematic changes included an increase in trunk flexion during propulsion (3.9%) and at release (3.4%), increased shoulder flexion at contact (4.6%) and during propulsion (13.1%), and increased elbow flexion at contact (3.6%) and during propulsion (1.7%). Elbow extension moment decreased with fatigue (8.9%) while shoulder compression force (Fy) increased (45.6%). Fatigue induced EMG changes included a decrease in the peak magnitude of flexor car-pi ulnaris and biceps muscle activity (Figure 1) during propulsion, a later onset of extensor carpi radialis muscle activity prior to contact (10.3% of cycle) and a decrease in the area of biceps activity during an entire cycle (35.3%). There were no significant changes in handrim forces or moments related to fatigue.

Figure 1. EMG changes with fatigue



DISCUSSION

When fatigued, subjects maintained propulsion speed by increasing stroke frequency as opposed to increasing propulsive moment. The latter would have been more difficult for fatigued muscle. Shoulder and elbow kinematic changes can be explained by the increase in trunk flexion. Leaning forward during the propulsion cycle may be advantageous since the shift in body weight can be translated through the arms and applied to the wheel to gain propulsion moment when fatigued. This kinematic adaptation to fatigue may explain the associated increase in shoulder compressive force. Fatigue induced changes in the FCU and ECR activity may predispose the wrist to injury. Similarly, the decrease in elbow extension moment and biceps activity with fatigue may be precursors of injury. Since the peak magnitude of biceps activity occurs during propulsion at about 15% of the cycle, it may act at its insertion with sufficient magnitude to become deleterious with fatigue. These findings elucidate fatigue mediated alterations in propulsion mechanics in wheelchair users. These changes may be related to chronic fatigue injuries and decreased independence. Analysis and interpretation of the effects of fatigue and exercise on wheelchair propulsion is continuing.

The Variation in Electromyographic Measures Obtained from the Erector Spinae Muscles During Simulations of Workplace Tasks

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INTRODUCTION: The reproducibility'** of the EMG signal is of great importance. Normalization[3,4,5] of signals is used to reduce variation between individuals, introduced by the measurement system, and to put various exposure measures on comparable scales.

The purpose of this study was to determine the reproducibility in several EMG measures of simulated workplace tasks when measured at the same or different time slots in one day, and on different days. It was also questioned if normalization of erector spinae (ES) EMG to spine compression force reduces between participant variation introduced by the measurement system.

METHODS: Eleven males participated in this study. Three data collection sessions were done; one in the morning and afternoon of the same day and one a week later. In every session a simulation of three workplace tasks was performed, each task contained 20 cycles of 1 minute cycle time resulting in a task duration of 20 minutes. Each task was divided into 4 blocks of 5 cycles.

Three tasks were performed; sagittal lifting of a 7.4 kg box from the floor to a table at waist height and back; lifting of two boxes (4.0 kg and 7.4 kg) and carrying to a table 1.6 m behind the participant; a right handed metal scrap pick up task in which eight 1.0 kg weights were lifted from the floor to a bin at knee height. Participants were free to choose a technique to perform a task. Calibration tests were performed to obtain a scale factor (spine compression force/EMG amplitude) for the normalization procedure[3] which is a linear calibration between the EMG amplitude and spine compression force. The EMG amplitudes obtained during the simulations of workplace tasks were multiplied by the scale factor to obtain spine compression forces.

EMG was recorded from the ES at the right side of the body at thoracic 9. After the first session the electrodes were left on for several hours and re-used in the second session. For the third session electrodes were reapplied (electrode spacing of 2.5 cm). Muscle activity was recorded with a portable A/D convertor (Mega ME3000P). EMG signals were pre-amplified (CMRR>l30dB; Bandwidth 20-500Hz). EMG signals of the tasks and calibration tests were full wave rectified and then averaged over 0.1s successive intervals. EMG was sampled at 1000 Hz. The average EMG (AEMG) and the 90th percentile of the amplitude probability distribution function(APDF) of the tasks were expressed in micro volts and Newtons spine compression force.

The variation in EMG measures introduced by differences between blocks of cycles, tasks, sessions, and participants is expressed as a percentage of the total variation. This percentage of total variation was obtained by dividing the mean square of each source by the total mean square and by multiplying this ratio by 100 percent. F-tests were done and a level of significance of 0.05 was used.

RESULTS: For 4 blocks of 5 cycles, the variation introduced within a task is very small and no significant differences between the 4 blocks of 5 cycles were found. This indicates that these EMG variables were very consistent during the simulations of all three workplace tasks (table 1). For tasks, more variation was introduced for the 90th percentile of the APDF compared to the AEMG indicating that the three tasks are similar in their AEMG estimates but have different "peak" characteristics. However, no significant differences between the three tasks were found. For sessions, an expected increase in variation was introduced but again no significant differences were found between the three sessions. And for participants, most of the variation could be explained by differences between participants which were significant. Variation between participants introduced by the measurement system was reduced using the normalization procedure.

Table 1. The percentages of total variation of the AEMG and 90th percentile muscle activity (unnormalized) and spine compression force (normalized) which are introduced by differences between 4 blocks of 5 cycles within tasks, between three tasks, between 3 sessions and between participants.

	Unnormalized data		Normalized data	
Source	Average (%)	90th percentile (%)	Average (%)	90th percentile (%)
4*5 cycles	<1	<1		
Task	4	15		
Sessions	9	12		
Participant	57	47	22	19

The amount of variation introduced by differences within a day (between sessions 1 and 2) and by differences between days (between sessions 1 and 3) was separated and is shown in table 2.

Table 2. An increase in the amount of variation from within day (between sessions l-2) to between days (between sessions l-3) is shown for unnormalized data.

	Average (%)	90th percentile (%)
Within a day	<1	4
Between days	18	26

DISCUSSION: AEMG and the 90th percentile of the APDF are reproducible for simulations of workplace tasks with a duration of up to 20 minutes. The reproducibility of EMG measures decreased for measurements done at two different times during a day without removal of the electrodes. When recording on different days, which requires reapplication of electrodes, the measures were not reproducible for the simulated workplace tasks studied. The EMG technique is able to distinguish between some of the actual differences between simulations of workplace tasks but improvements are needed by reducing the variation around the means so that the technique is more sensitive to actual differences. The normalization procedure reduced between participant differences which were introduced by the measurement system.

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The authors acknowledge the contributions of P. Grills and M. Fraser to the data collection.

Influence of Bed-height Adjustment on Muscle Activity During a Nursing Task, Using Kinesiological Surface EMG and the Rating of Perceived Exertion

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INTRODUCTION

Kinesiologal EMG is used as a variable in several but specific ergonomical situations, in recent studies: e.g. a study of static and dynamic lifing tasks (Yates and Karkowski, 1992), fatigue during repetitive light work (Nakata et al., 1992), excessive drafts on shoulder muscles (Sundelin and Hagberg. 1992). during typewriting and keybord use (Fernstrom et al., 1994), effects of precision and force demands in manual work (Milerad and Ericson, 1994), in symmetric and asymmetric lifting tasks in restricted postures (Gallagher et al., 1994), verifying spinal and abdominal muscle activity during garden raking (Kumar, 1995), and during repetitive lifting tasks (Kim and Chung, 1995), assessing load on upper Trapezius in jet pilots (Harms-Ringdahl et al., 1996), monitoring the influence of the operating technique on muscular strain (Luttmann et al., 1996a) and on muscular fatigue of surgeons in urology (Luttmann et al., 1996b). The measure of perceived exertion using the Borg scale is commonly used in sports, training, exercices and occupational circumstances, and supplementary to physiological variables it is considered to be a standardised tool for describing the perception of physical efforts (Noble and Robertson, 1996).

PURPOSE

The aims are to study the influence of individually chosen bed-height adjustments upon muscle activity using kinesiological surface EMG and to study the rating of perceived exertion, performing a patient-handling task at a standard and at an adjusted bed-height. This study is based on the effects of individually chosen bed-height adjustments on peak values and time integrals of spinal compression and shear forces in the low back of nurses using a dynamic biomechanical model (de Looze et al.. 1994).

METHODS

The performed nursing task is lifting a hemiplegic patient from sitting position on the edge of the bed. to erect position (Caboor et al., 1997). Twenty-two nurses volunteered for the study, including 14 female (mean age 31±7.57 year, mean height 166.41±6.33 cm, mean body mass 61.95±5.28 kg) and 8 male nurses (mean age 32f7.48 year, mean height 175.18±5.60 cm, mean body mass 75.79±7.11 kg). The active bipolar Ag-Cl electrodes were attached after standard skin preparation and were positioned in a longitudinal direction above the bulkiest part of the contracted muscle belly of the M. obliquus externus abdominis, the M. erector spinae and the M. biceps femoris both at the left and the right side of the body. The raw EMG-data were recorded using a Teat-MR30, and AD converted with a sample frequency of 1000 Hz. The data were full wave rectified and the linear envelope was normalized to the highest peak. The integrated values (iEMG) of the muscle activity during the task were compared for standard bedheight (51.5 cm) and adjusted bedheight. At the end of each session they were asked to rate the perceived exertion on the 15-graded Borg scale.

RESULTS

The nurses could adjust the bedheight as in lowering as in raising it. The female nurses raised the bed up to 54.67 cm (n=9), lowerd it down to 43.88 cm (n=4), or did not change it at all (n=l). The male nurses raised the bed up to 53.13 cm (n=4) or lowered it down to 44.88 cm (n=4). Both for women as men, the analysis of muscle activity shows an overall tendency of lower iEMG values in lifting the patient out of the adjusted bed, but no significant differences are found ($p<0.05$). The measures on the 15-graded Borg scale were 12.85f1.95 for the standard bedheight and 12.93 ± 2.13 for the adjusted bedheight, so the perceived exertion shows no significant differences ($p<0.05$).

DISCUSSION AND CONCLUSION

The influence of bed-height adjustment on muscle activity depends on the capability of the nurses to reduce this activity by bed height adjustment. According to de Looze et al. (1994) we can conclude that the major factor interfering with possible favourable effects, is the restricted capacity of the nurses to select an optimal bed position with respect to the specificity of the task. It is assumed that the standard bed-height is already close to optimal for the majority of the subjects. These data suggest that the adjustments of the bed-height are decided for other reasons than minimizing low back stress, or that changing the bed-height created a psychosomatic placebo effect.

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Ergonomical Comparison Between Left and Right Back and Leg Muscle Activity, Using Kinesiological Surface EMG, During a Typical Nursing Task

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INTRODUCTION:
The present ergonomical study is a part of a European Biomed Program. According to the study of De Looze et all (1994) it is proven that adjustable bedheights have a preventive value in low back pain.

PURPOSE:
This study tries to find out whether there are changes in muscular activity (iEMG) during the performance of a typical nursing task. The left side of the body (M.Erector Spinae, M.Obliquus Ext., M.Biceps Femoris) is compared with the right side. Adjustable bedheigts are introduced during the taskperformance.

METHODOLOGY:
A group of nursing personnel (Female: n=14, Male: n=12 all but two were righthanded/left-footed) were trained to perform several typical nursing tasks while EMG was registrated continuously. One of the tasks was the positioning of a right-sided hemiplaegic patient from the upright sitting position to the standing up position. This task was carried out several times before the actual recording in order to decrease the intra-variability of the EMG signals. Both standard bedheight conditions (5 1,5 cm) and adjusted bedheight conditions were evaluated. Individually adjusting a bedheight could result in both lowering and raising the bed. Bipolar active surface electrodes were used and located upon the right and left sided M.Erector Spinae, M. Obliquues Externus and the M. Biceps Femoris. The sample frequency was 1000 Hz.

RESULTS:
For woman, the muscular activity in standard bedheight conditions show a higher integrated EMG-value (except for the M.Biceps Femoris) for the left side of the registrated muscles when compared to the right side. In the adjusted bedheight positions, data show the exact opposite, namely, left muscle activity is lower than the opposite side of the body. These results however do not reveal a significant difference.
For men as opposed to women, right muscular activity shows a higher integrated EMG-value (except for the M.Biceps Femoris) when compared with the left side of the body, during standard bedheight conditions. In adjusted bedheight conditions the left sided muscles produce a higher iEMG value, except for the M.Erector Spinae. The only significant difference ($p < 0.05$) for men in iEMG is found in the M.Biceps femoris during standard bedheight conditions.

DISCUSSION & CONCLUSION:

During the standard bedheight conditions men seem to start (in this typical nursingtask) in opposite contractions as opposed to women, by producing higher iEMG values on the right side of the body as opposed to woman who produce higher iEMG values on the left side of the body, both in back- and legmuscles (except for the M.Biceps Femoris). When the bed is adjusted both men and women adapt by inverting the left-to-right contractions, however not significantly (except for the M.Biceps Femoris). The analysis supports the the need for adequate training of nursing personel as the results show an overall decrease of the muscular activity. Left to right comparison does not only show a non-significant difference in iEMG values but also a difference in co-activation muscle patterns.

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Referent Body Configuration for Multi-Muscle Control in Complex Movements

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It has been suggested (Feldman & levin 1995) that the coordination of the activity of multiple muscles results from the comparison of the actual configuration of the body with a referent configuration specified by the nervous system so that the recruitment and gradation of the activity of each skeletal muscle depends on the difference between these two configurations. Active movements may be produced by the modification of the referent configuration. The hypothesis predicts the existence of a global minimum in electromyographic (EMG) activity of multiple muscles during movements involving reversals in direction. This prediction was tested in 5 subjects by analysing movements resembling the act of reaching for an object placed beyond one's reach from a sitting position. In such movements, initially sitting subjects raise their body to a semi-standing position and then return to sitting. Consistent with the hypothesis is the observation of a global minimum in the surface EMG activity of 16 muscles of the arm, trunk and leg at a specific phase of the movement. When the minimum occurred, EMG activity of each muscle did not exceed 2-7% of its maximal activity during the movement. As predicted, global EMG minima occurred at the phase corresponding to the reversal in movement direction, that is, during the transition from raising to lowering of the body. The global EMG minimum may represent the point at which temporal matching occurs between the actual and the referent body configurations. This study implies a specific link between motor behavior and the geometric shape of the body modified by the brain according to the desired action.

Neuromuscular Effects of Exhausting Rowing Ergometry

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INTRODUCTlON. Rowing ergometry is routinely used in performance diagnostics of elite athletes. Cardiorespiratory and metabolic parameters serve as standard measures for endurance capacity (1). The recent study focussed on the effects of a 6-minutes maximum power test on intermuscular coordination between m.tibialis anterior, m.gastrocnemius medialis, m.biceps femoris and m.rectus femoris. Timing of EMG-onsets as related to normalised stroke duration did not change over time. For the rectus femoris typical fatigue effects such as increasing AEMG amplitudes and decreasing MedianPF content of the myoelectrical signal could be identified and attributed to the work load distributed to the knee extensors during rowing.

METHODS AND PROCEDURES. Each of twenty male top level rowers (age: 17,8 +/- 3,3yrs, height: 185 +/- 7cm, weight: 74,7 +/- 4,4kg) performed a six minutes maximum power test on a Gjessing oar rowing ergometer. Eight minutes of rowing below the anaerobic threshold were used as a uniform preloading procedure in order to ensure adequate warmup. All subjects have been well accustomed to the ergometer before. Stroke force was measured between the oar and the steel rope connecting to the ergometer with a onedirectional force transducer. Simultaneously surface EMGs have been recorded for m.tibialis anterior, m.gastrocnemius medialis, m.biceps femoris and m.rectus f'emoris. The interelectrode distance ranged from 1,8cm to 2,Ocm; the skin below the selfadhesive electrodes (Medicotest type N-00-S) was shaved before electrode placement, which reduced skin resistance below $20k\Omega$. To minimize movement artifacts the electrodes together with the preamplifiers have been clued to the athletes skin with selfadhesive tape (Leukofleece). EMG output signals have been preamplified directly behind the electrodes avoiding the amplification of artefacts due to cable movements. EMG and force data have been gathered at the start and after every consecutive 50sec after the start for always 10s at a sampling rate of 1OOOHz. All data have been stored online to a PC. Before the maximum power test, directly after and 5 times within five minutes after the test blood samples were taken from ear in order to measure blood lactate concentration. EMG data have been rectified, integrated and time normalised according to the duration of the respective stroke cycle (one cycle =lOO%) as identified from the stroke-force time history. Data for 5 cycles out of each measurement period for each and all of the athletes were averaged. Signal amplitudes and median power spectra have been calculated for every measurement period and muscle. EMG-onsets were related to the respective onsets of the stroke force in order to identify intermuscular coordination patterns.

RESULTS. The external power output as given by the ergometer averaged 276,21 Watts; mean lactate concentration directly after the test was l4,4mmol/1. The subjects reached a mean maximum heart rate of 198bps. After the start the stroke force decreased for every subject throughout the test with only small increases during the final spurt (see fig.1). As external mechanical power output was nearly constant over time with a small increase towards the end of the test the athletes must consequently have increased their stroke frequencies with time. EMG patterns were not found to have changed significantly over time eccept for the m.rectus femoris ($p < 0.05$). Mean AEMG amplitude increased and median power of the m.rectus femoris shifted towards lower frequencies. These effects are generally attributed to muscular fatigue (2). They additionally identify the knee extensors as the leg muscles being predominantly involved in the stroke action. This can be supported by relating the timing of muscle actions to the stroke-force time history. M.tibialis anterior is primarily active during the forward roiling action, when it dorsiflexes the foot to give mechanical support for the biceps femoris flexing the kneejoint. The m.gastrocnemius is only active during the end of the stroke time piantarflexing the foot and thus contributing to leg extension. The m.rectus femoris shows a two phase action starting shortly before stroke force onset when knee extension is initialised. Towards the end of the test this preactivation period is prolonged. While the force is still rising its electrical activity decreases and increases again when stroke force already decreases (fig. 1). The m.rectus femoris now acts as a hip flexor errecting the trunk and bringing it back to the power position. The m.biceps femoris cocontracts with the m.rectus femoris in the first activation phase and the m.gastrocnemius until full extension of the leg. With the rowers' feet fixed to the foot stretcher the net effect of the biceps action is a knee and hip extension as previously described for sprinting (3).

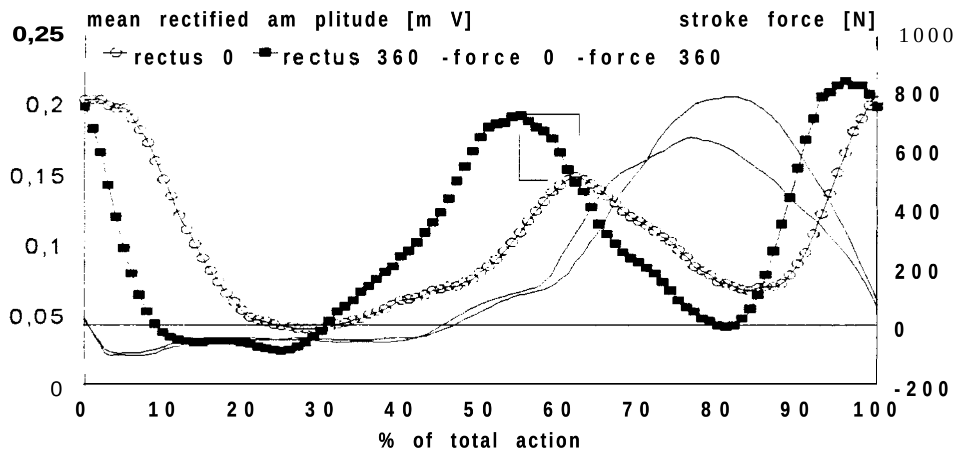

Figure 1: Mean rectified m.rectus femoris AP amplitude [mV] and stroke force [N] to action duration (%of stroke time) history for the very first (rectus 0, force 0) and the last measurement period (rectus 360, force 360)

Comparing the different muscles activation times reveals that the m.rectus femoris has considerably longer been activated than any other of the muscles under investigation. As fatigue effects have only been identified for this specific muscle we might speculate that also its work load was higher than for the other muscles. For ail that the m.rectus femoris became fatigued the muscle coordination as given by the order of the AP onsets did not change with time. The exhausting character of the six minutes maximum power test as it is also reflected in the blood lactate concentrations suggests that local fatigue effects might limit the performance more than it can be assumed from physiological measures alone. Further investigations are necessary to additionally study the contribution of the other knee extensors and hip flexors. To fully describe the accelerating rowing action the trunk muscles need to be included as well. To relate our findings to the actual rowing performance the study needs to be repeated on the open water.

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Muscle Fatigue in Male and Female Rowers

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INTRODUCTION

Little is known about possible differences in muscle fatiguability between gender. Older women ($\geq$60y) have been reported to fatigue less rapidly than older men of the same age (2,4) whereas in younger age groups, studies have reported either no gender differences (3,4) or greater endurance in women (5). Part of the difficulty in comparing gender differences in muscle endurance and fatiguability lies in the difficulty of pairing individuals based on their level of physical fitness. To circumvent this problem, we investigated male and female elite and novice rowers.

METHODS

A total of 24 rowers volunteered to participate in the study. During baseline evaluations contractile properties of the quadriceps muscle were obtained to determine the unpotentiated peak twitch (Pt), time to peak tension (TPT), half-relaxation time (1/2 RT), compound muscle action potential (Mwave) and integrated electromyography (IEMG) of the non-fatigued muscle. The Interpolated Twitch Technique (ITT)(1) was administered to determine the ability of each subject to maximally activate the quadriceps muscle during a maximal voluntary contraction (MVC). Rowers then performed submaximal isometric contractions at 50 % of their estimated MVC until exhaustion. Each cycle consisted of a 3s ramp-up, a 10s hold, a 3s ramp down and a 4s rest period. The muscle was considered exhausted once the force could not be maintained at 50 % of the MVC for 10s. The number of successive submaxima! contractions was recorded as a measure of muscle endurance. Immediately following exhaustion, the quadriceps muscle was reevaluated along with subsequent evaluations at 1, 2 , 5 and 10 min of recovery in order to monitor the recovery pattern of all parameters tested.

RESULTS

Women were able to complete significantly more contractions than men during the fatigue protocol (15.6 f 9.3 versus 9.8 ± 5.3, respe[illegible]
post-fatigue was less in females (29.9 ± 4.0 % than males (38.1 ± .7 %)($p<0.0008$). imilarly, Pt decreased less in the females O. 02). The effect of fatigue on all other parameters investigated was similar across gender (table 1).

During the recovery period, the MVC recovered to the same extent in men and women. Activation, as measured by the ITT, was significantly less during the recovery period in females than males $p<0.05$). Twitch tension remained significantly depressed by the end of the recovery period. The pattern of recovery of TPT ($p<0.05$) and ½ RT($p<0.01$), however, differed between gender. In males, TPT decreased with fatigue and increased past baseline during the recovery period. In females, TPT also decreased with fatigue but remained unchanged throughout the recovery period. The ½ RT was not influenced by fatigue in either gender; in the recovery period, however, it decreased in males and was unchanged in females. The changes in quadriceps IEMG and Mwave were similar between gender.

Table 1. Fatigue (time 0) and 10 min recovery (Time 10) values for male and female rowers, expressed as a percentage of pre-fatigue values.

	Men		Women	
	Time 0	Time 10	Time 0	Time 10
MVC	61.9±5.4	80.4±6.9	70.1±14.0	85.1±14.0
ITT	94.4±5.8	98.4±2.5**	94.5±7.0	92.8±10.1**
Pt	34.5±19.6	46.4zk26.3	40.0±15.3	44.5±18.5
TPT	89.4±13.6	109.4±27.9**	92.0±15.0	93.6±16.5**
½ RT	102.8±74.2	76.4±54.3**	112.0±79.6	111.2±79.0**
Mwave	114.0±80.8	106.4±69.4	109.1±48.5	85.8±15.8
IEMG	63.0±24.0	67.7±29.3	76.6dz14.9	70.6±19.3

The underlined c ata represent significant gender differences in the extent of decrease from pre-fatigue evels. The aster sks represent significant gender differences in recovery profile.

DISCUSSION

Women were able to complete more contractions while sustaining a smaller decrease in MVC than men. Fatigue had similar effects on the ITT, Mwave and IEMG indicating that differences in the level of central fatigue or excitation were not factors contributing to the women's superior endurance. In contrast, Pt decreased more in males than females. This would seem to indicate that the greater fatiguability in the men was due to greater impairment in excitation-contraction coupling (ECC). During the 10 min recovery period, MVC recovered to a greater extent in males but activation levels were lower in females. If activated to the same extent as the males, recovery of MVC would have been similar. Similarly, there was a tendency for a greater recovery of Pt ($p<0.09$) in men. The fact that TPT remained faster in women may help explain why their Pt recovered to a lesser extent. In conclusion, the greater endurance in women could be due to less impairment in ECC.

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Metabolic Effects of Exhaustive Rowing Ergometry

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INTRODUCTION

Rowing performance is to a high degree influenced by the power which can be produced during each stroke (S) of the average 230 S per race. Assuming a competition time of 6 to 7 min, 75% of the energy are recruited aerobic, 15% are generated by anaerobic lactic, and 10% result from the anaerobic alactic source. - To explain fatigue during exhaustive rowing it is not clear, whether primarily neuromuscular effects during rowing (paper Knicker et al.: Neuromuscular effects of exhausting rowing ergometry) or metabolic effects are mainly responsible. – Therefore the aim of this study is to explain the energy delivery during simulated rowing using a computer simulation model, which allows the calculation of the dynamic of the metabolic activity of glycolysis and oxidative phosphorylation as well as the phosphorylation state of the ATP/CRPH-system as a function of power output.

METHODS AND PROCEDURES

During a 6 min maximal tests (6MMT) on a Gjessing rowing ergometer the average mechanical power output of 20 male young talented rowers (18±3 years, 185±7 cm, 75±4 kg) was about 276 watt, heart rate (HR) was 191 beats/min and maximal oxygen uptake (VO2max) could be calculated with 4400 ml/min (relative VO2max 60 ml/kg body weight (BW)), maximal post exercise blood lactate (PEBLA) 17,5 mmol/l. - Tests were conducted with 2.75 kg brake load, stroke rate ranged from 30-35 S/min. The ergometer was equipped with a strain gauge (Hottinger Baldwin Messtechnik, Darmstadt, Germany, type U9, 5000 N), positioned in the pulling rope. An ultrasound echo system (self-construction) was used for velocity measurement and was installed at the pulling grip. For further technical information see (1).

RESULTS

After the starting phase (12 s) a steady decrease of peak force (PF), peak velocity (PV), and peak power (PP) was obvious. - Related to the PF of the 1st max S (1 OO%), PF was approximately 64% after the 1 0th S, 63% at the end of the 2nd min, and 57% during the finishing phase of the 6MMT. - PV was lowest during the 1st S (60% of max) and increased exponentially to the 5th S (100%). After 2 min PV was approximately 85%, and at the finish of the test it corresponded to 86%. - PP was 73% in the beginning and was 100% in the 5th **S.** During the 2nd min it was 77%, during the finish 74%.

THEORETICAL BASE OF MODELLING

According to those results a typical load situation was developed on the basis of the results during routine laboratory testing: 230 rowing strokes; average 33 S/min, duration 7 min, every 1,85 s an active muscle action performing a S. Load is characterised by an intermittent time course; occurring power is around 1700 w in the starting phase and 1400 w on average after 20 strokes at the beginning. - The model itself is based on the assumption that the cytosolic phosphorylation state regulates the rate of oxidative phosphorylation as well as the glycolysis. A decrease of CRPWATP-content by contraction increases first at a low level of power output the rate of oxidative phosphorylation. At about 60% decrease of CRPH-content which corresponds to about 65% of VO2max, glycolysis increases steeply leading to a net lactate formation and lactate accumulation. - The change of the cytosolic phosphorylation state results from the rate of ATP consumption and the total resynthesis. - Including a time delay of 02 transport of first order this can be calculated with a system of two nonlinear differential equations (3). - The real power output as given was used as input in the simulation model. – In the literature extensive explanations and clarification of the purely theoretical foundations (3) and about the configuration of an „virtual athlete“ (2) are given.

SIMULATION OF MAXIMAL LOAD DURING ROWING COMPETITION

The post-exercise simulation of the maximum load in rowing shows significant parallels to the parameters measured in the practical experiments: Assuming an absolute O2-uptake of about 4400ml/min during a 336 watt maximal performance (276 watt plus 60 watt body shift performance) of young talented rowers lead to a gross VO2max of 129ml/kg for actively used muscle mass (MM). This corresponds to a net VO2 of 115ml/kg per kg active MM and a mean performance of 11.5w/kg. - Because of the high initial performance (see above), the CRPH content of 21 mmol/kg decreases by 13 mmol/kg or 62% within 25 s. During this period the calculated decrease of ATP is about 0.5 mmol/kg (17%). PeakVLass reaches about 35%VLamax (= 0.5 mmol/s*kg), after about 25 s it decreases by about 25% and later again to 20%; from then it hardly does not change until the end of the test. - After an initial short-term delay of VO2 there is a steep rise which comes to a maximum after about 90 s. Because of the initially short-term activation (60s), the blood LA rises to about 4 mmol/kg, which corresponds to about 28% of the final concentration. The maximal blood LA concentration occurs with delay after 8 min in the resting phase.

Even if the complete removal of CRPH at the end of the load of the 6MMT were assumed, a 83% utilisation of VO2max at approximately 18 mmol/l PEBLA could be calculated. The utilisation of the theoretically available VO2max could be increased through the reduction of VLAmax (glycolytic performance). Because of the high start power output, a CRPH-concentration close to the level of ATP was calculated already after 30 s. A steeper decrease would, however, make the continuation of exercise impossible. – The behaviour of the individual metabolic parameters during the phase immediately following the exercise will be presented.

CONCLUSION

From the metabolic performance available, no immediate inferences to competition velocity are possible, not even if a corresponding motivation during competition is taken into account. Athletes can only to some extent compensate for a low metabolic performance by using an excellent rowing technique. Although a high metabolic performance is necessary, it is not a sufficient prerequisite for competition success. - It becomes clear that the exercise simulation of metabolic conditions allow deeper insights and further-reaching interpretation possibilities. - Beside the neuromuscular effects to explain the fatigue during exhaustive exercise there are also massive influences by the given metabolic conditions.

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To Ergo or not to Ergo ? The Biomechanics of Ergonometric Rowing

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INTRODUCTION **Since** its conception, the sport of rowing has been viewed as an excellent athletic discipline for increasing overall strength, endurance and power. In recent years, rowing simulators have formed an integral part of on-land training programs and become an indispensable device for evaluation and research purposes.[3,4] Due to the limitations of its design, however, athletes may sacrifice an effective on-water rowing technique and still attain a competitive performance. With experience, rowers develop alternative postural mechanisms which compromise technique while attempting tomaximize output scores on the ergometer. These poor postural habits may adversely affect on-water performance and potentially lead to injury. The incidence of lower back and knee injuries has grown very rapidly as the number of participants in the sport tends to increase. To help reduce the risk of injury, physiotherapy, comprising stretching and strengthening exercises, has been recommended. [7,8] The pelvis and the spine should be stabilized to support the forces being generated during the rowing motion to further minimize the risk of injury while maintaining an effective stroke.[1] Precise measurement of effective rowing technique may greatly contribute to injury prevention for athletes rowing an ergometer. In some studies, muscle activation and fatiguability characteristics have been found to differ significantly amongst novice and elite rowers. Kinetic analyses have revealed more consistent peak force applications in elite than in novice rowers.[2,9] Elite rowers have also shown a distinct pattern of activation in muscles that are more resistant to fatigue.[9] At the moment when the oar is perpendicular to the boat, the hip and knee joint angular velocity of elite rowers is greater and more consistent than for novice rowers, resulting in a more efficient stroke.[5] Novice rowers tend to flex their knees and hips before the completion of the drive phase creating a sudden instability of the lower back and a decrease in power output.[9] A more in-depth understanding of the biomechanical factors which govern an effective rowing technique, both on and off the water, may reduce injury-related stress.[10] The objective of this study was to identify the biomechanical factors influencing effective and safe ergonometric rowing technique. It has been hypothesized that providing biomechanic feedback to the athlete may enhance rowing performance.@

METHODS A novice (NV) and an elite (EL) rower were evaluated during a single 2500 m race on an instrumented Concept II model-C rowing ergometer (wind damper set at level 4). Performance data are presented in Table 1. Kinetic data was recorded using a miniature load cell which measured the force exerted on the chain. Kinematic data was recorded utilizing a motion analysis system Reflective markers were placed on the left side of the ankle, knee, hip and shoulder joints, the mid-shank, mid-thigh, and the ergometer handle. Ten consecutive strokes were recorded at the start, at each 500 m interval and during the last 200 m of the race. Subjects rowed at a self-selected stroke rate.

RESULTS Kinetic data revealed a distinct force profile for each subject. The EL rower consistently produced higher peak force, impulse per stroke and impulse per minute. This subject was quicker to begin force generation, while the NV generated force further into the stroke. Kinematic data revealed similar

Table 1 **Novice (NV)** and elite (EL) subject's data

	Age (years)	Weight (± 0.4 N)	**Height** (cm)	**Experience**	Test-time (min:s)	Strokes per 25OO m test	**Stroke rate** (strokes/min)	**Split-time** (min:s)	**Impulse/min** (± 0.1 kNs/min)
NV	**22**	**758.7**	**178**	**8 months**	**8:21**	**241**	**28.7**	**1:39**	**10.8**
EL	**39**	**829.8**	**191**	**8 years**	**8:03**	**215**	**26.6**	**1:35**	**11.5**

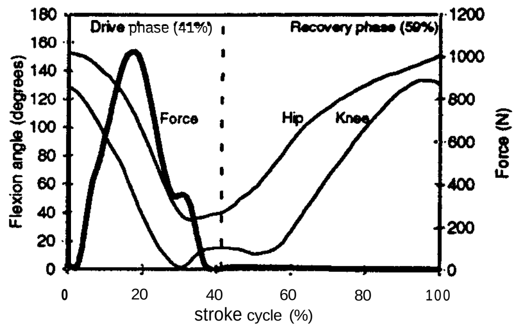

Figure 1 Hip and knee flexion angle and force profile of the elite rower at interval 4 as a function of the stroke cycle.

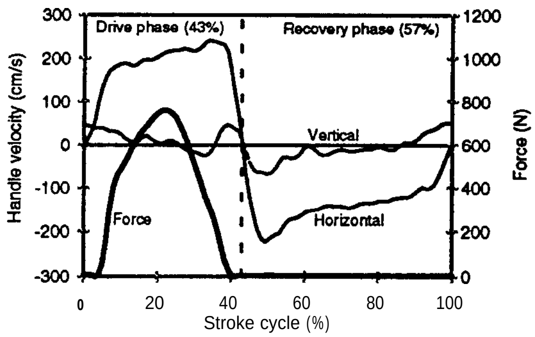

Figure 2 Force and handle velocity profiles of the novice rower at the interval 4 as a function of the stroke cycle.

knee and hip joint angles at the catch and finish of the stroke for both subjects. At the end of the drive phase, a knee joint bounce and initiation of hip flexion were observed. This was more evident in EL, indicating an out-of-phase knee and hip joint reversal (Fig. 1). Near the onset of the catch, EL began to extend the knee while the hip continued flexing. Similar horizontal handle velocity curves were obtained from both subjects. Vertical handle velocity curves, however, showed a bi-directional oscillation, with greater amplitude in the EL rower. The largest oscillation for EL was observed just before the finish of the drive phase concurrently with **a** "step" in the descending force curve (Fig. 2). At the catch, the NV produced an upward displacement of the handle whereas the EL had no movement.

DISCUSSION **The analysis** of rowing strategies at the elite and novice level in this study has provided further insight into the biomechanical factors influencing ergonometric rowing. Both subjects had a similar knee flexion angle at the moment of peak force occurrence; this may correlate with an optimal muscle length-tension relationship. The EL rower produced a consistently higher peak force, impulse per stroke and impulse per minute. The greater peak force generated by EL likely resulted in high articular contact pressure. It would be recommended to reach a lower peak force over a longer duration in the drive phase to minimize mechanical stress at the joints. Both subjects expended unnecessary energy as demonstrated by sudden vertical handle displacements. If made on-water, these movements may have compromised the stability **of the** boat and stroke efficiency. The pronounced bounce of the knee joint, and the subsequent out-of-phase knee and hip joint reversal may also have compromised the stability of the pelvis and spine. This may place the rower at risk for injury during the lag phase. Although EL demonstrated a superior performance on the ergometer, the NV exhibited a safer and more effective on-water technique. Possibly with experience rowing an ergometer, EL has adapted a different postural mechanism to maximize stroke output. If this poor postural habit had been used on-water, however, it would have radically diminished performance. To prevent a carry-over effect of poor ergometer training habits to on-water rowing in the early stages of learning, coaches must supervise and emphasize the use of on-water rowing technique during simulated rowing. Further research will identify the crucial technical characteristics that are essential to enhance athletic performance. With access to the instrumentation used in this study, coaches and rowers can obtain quantitative biomechanical feedback on the technique of their rowers.

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Posturo-Kinetic Control During Forward Oriented Stepping Initiation on Level Surface and During Ascent

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INTRODUCTION

Forward oriented stepping (FOS) initiation is associated with important functional goal as far as it **is** essential element of the gait initiation. FOS serves to orient and move the body to a given goal. The position or alteration of the center of foot pressure (CP) beneath each leg, is a parameter of interest because it is related to the trajectory of Center of gravity (CG) the main variable to be regulated during posturo-kinetic coordination. Thus, better to gain inside in the mechanisms underlying bilateral organization, it is essential to obtain a position of CP alterations during FOS initiation. In a recent studies by means of two force platforms (Gantchev at al, 1996; 1997) bilateral leg organization during stepping initiation on level surface was studied by CP profiles. To our knowledge there are no data concerning the CP alterations beneath each legs during FOS initiation during ascent. If we assume that the motor program for the FOS control on level surface is likely to be the same as for the FOS initiation during ascent, however differences in the shape and the time course of the CP curves beneath both legs could not be neglected. If they exist they might be associated with the specificity of the motor tasks especially related to the influence of the gravitational forces. Thus the present study was designed to compare CP alterations beneath both legs during FOS on level surface and during ascent, to test the hypothesis that FOS is govern by one and the same motor program regardless of the mode of initiation (Crenna & Frigo 1991) but specifically modulated (Gantchev et al 1996). The specific aim was to analyze the CP beneath each leg, to gain inside of the bilateral leg organization of FOS initiation during maintaining the body balance.

METHODS

Six male subjects with no history of neuromuscular disorders aged between 25 and 35 years old were asked from initial quite stance on two force platforms to initiate a single step. The FOS started with the right leg (leading leg) as soon as the tone appeared and was terminated with the left (trailing) leg placed in parallel. Two mode of FOS were executed: a/ FOS on level surface; b/ FOS during ascent. With each mode of FOS initiation eight consecutive steps were collected. The CP alterations beneath both legs were recorded and analyzed. The onset, pure phase of initiation and phase of transfer (Gantchev et al 1997) were compared among the two modes of FOS. The significance of mean values was estimated by Student t-test.

RESULTS

With each mode of FOS initiation, based on CP alterations in saggital plane two phases mutually connected were considered. Phase of pure initiation (PPI) starting with the onset of CP beneath each leg and finishing with the maximal backward shift beneath the leading leg. During both modes of FOS initiation the onset of CP beneath each legs was practically simultaneous. In general the shape of CP alterations was similar among both modes of FOS. However, the duration of PPI of the leading leg was significantly longer during FOS on level surface than FOS during ascent. The initial CP shift, always directed backward, was significantly higher during FOS on level surface for the leading leg than during FOS in ascent. Transfer phase immediately follows the PPI and ends when leading leg leaves the platform. Again there was similarity in the shape of the CP alterations. Interesting is to note that the duration of this phase is the same for the two modes of FOS initiation. The maximum backward shift beneath the trailing leg tend to be smaller than FOS during ascent. Total duration of postural phase for the FOS up on level surface was significantly shorter as well. The forward CP shift maximum following the postural phase in the FOS during ascent series was significantly larger.

DISCUSSION

The effective FOS initiation, which is a complex motor task, represented by transition from a quiet standing position to a given position depends on precise coordination between the whole body posture and subsequent movement. While during FOS the CG is displaced forward, during stepping up (climbing a stair) the CG is displaced forward and upward against the gravitational forces. Thus, the control of body posture and balance is supposed to be different, evaluated by PPI and TP in the present study. Higher and lengthened EMG activity of Triceps surae and Quadriceps muscles responsible for the body elevation was reported McFadyen & Winter (1988). It could be argue that the role of the propulsive forces in saggital plane during FOS at level surface are higher compared to the that during stepping up. That might be a reason PPI duration to be significantly longer during FOS on level surface compared with the FOS during stepping up. Analyzing the CP alteration in saggital plane beneath both legs during different modes of FOS, two main conclusions can be drawn. The observed increase of PPI lengthening together with the higher initial CP shift directed backward represents higher mechanical effectiveness needed for the subsequent forward propulsion necessary for the FOS initiation on level surface. As for the smooth transfer of the body weight laterally, represented by TP before unloading and advancing, it seems that the control is the same during both modes of FOS initiation. However, the tendency for higher maximum shift for CP alteration beneath trailing leg and the significantly larger forward CP shift maximum following the postural phase at FOS during stepping up represented the need for adapted bilateral coordination associated with the increase effect of the gravity force. The data lead to conclusion that the program for these motor activities is the same but specifically modulated by the biomechanical constraints. Thus, it is more reasonable to accept the hypothesis that one and the same motor program surveys to promote precise coordination between initial posture and subsequent FOS initiation despite the specificity of the mode of initiation already suggested during climbing the stairs (McFayden & Winter 1988). The approach used in the study is adding additional information for the bilateral organization during different modes of FOS initiation.

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Determination and Sensitivity of Lower-Limb Muscle Lengths for use in Clinical Gait Analysis

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INTRODUCTION

Gait analysis is now a frequently used diagnostic tool for the analysis of pathological gait, especially for operative planning of individuals with cerebral palsy. In conventional gait analysis, angle-based kinematics and kinetic data are typically presented. Data in this form present a simplified picture of the gait pathology, where other derived information such as muscle lengths and velocities may be of clinical use. Aside from the obvious clinical utility in knowing the changing muscle kinematic patterns during the gait cycle, these values may also have utility in the ultimate estimation of muscle force. Results from a sensitivity analysis and a case-study are presented.

METHODS

To provide an estimation of muscle lengths, the following components are necessary: a means of recording three-dimensional movement of the lower-limb during gait, and a scaleable database of muscle origins and insertions. A VICON V370 (Oxford Metrics Inc.) system utilizing four-60Hz CCD cameras (system accuracy ±6mm) was used to record the unilateral movement of the lower-limb via two wand and seven surface reflective markers (see figure 1)[1].

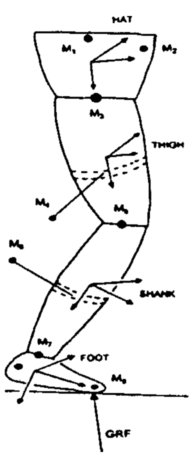

Figure 1: Marker configuration

Local (segment-based) co-ordinate systems were determined using equation (1) to identify the segment orientation vectors (e.g. Pelvis) and a 4x4 trans.-formation matrix (2) using the Denavit-Hartenburg notation[2]:

$$X_{pel} = \|M_3 - M_1\|$$
$$Y_{pel} = \|(M_2 - M_1) \times (M_3 - M_1)\| \qquad (1)$$
$$Z_{pel} = X_{pel} \times Y_{pel}$$

$$R^{PEL}_{LAB} = \begin{bmatrix} X_1 & Y_1 & Z_1 & t_x \\ X_2 & Y_2 & Z_2 & t_y \\ X_3 & Y_3 & Z_3 & t_z \\ 0 & 0 & 0 & 1 \end{bmatrix} \qquad (2)$$

Accurate skeletal dimensions and landmarks are essential in determining musculotendon lengths. A recent survey of skeletal landmarks by the NIH [3] was employed. This database has as at least 50 samples in total of each cadaver bone taken from different gender and race. The database was used to map the origins and insertions of some 41 muscles per side referenced to the pelvis, femur, tibia-fibula, and foot. The NIH database's bone co-ordinate systems were based on the work of White et a1.[4], and were transformed to align with our marker-based co-ordinate system.

Common anthropometric measurements taken in the laboratory were used to enable linear-scaling of bone dimensions along three independent local (bone-based) co-ordinate system axes. This scaling was necessary to estimate subject-specific landmark (attachment) sites.

A software program was written in the MATLABTM (The Mathworks Inc.) interpretive language' using the aforementioned Denavit-Hartenburg notation to simultaneously transform position and orientation vectors and calculate muscle lengths via global co-ordinates of muscle origin, pathway and insertion for each time step.

RESULTS

To ascertain the sensitivity of estimated muscle origin and insertion sites, a limited sensitivity analysis was performed on ten muscles of relative importance to clinical gait analysis. Ten unilateral trials of the right-side were taken from an able-bodied male (29 yrs, 81kg, 1.82m). Angle and muscle-length kinematics were calculated for each trial and ensemble averaged. The estimated origin sites for each muscle were then moved lcm proximally and distally along the long axis of the bone, and the errors in muscle lengths were calculated. The amount of error derived from this change in muscle length was dependent on the muscle involved and the bones it crossed. Of the ten muscles surveyed, the following changes in muscle length resulted: semimembranosus (± 1.7%), adductor brevis (±3.5%), rectus femoris (± 1.8%), gluteus medius (±4.8%), vastus lateralis (±3.7%), medial gastrocnemius (±2.0%), lateral gastrocnemius (±3.0%), tibialis anterior (±1.6%), tibialis posterior (±1.9%),and soleus (±2.5%).

To demonstrate the utility of muscle-length estimates for clinical analysis, clinical data were analyzed from a single subject with spastic diplegic cerebral palsy prior-to and following selective dorsal rhizotomy [5]. Normal and pre-operative muscle lengths were normalized to the post-operative levels by multiplying by a factor to account for height differences.

Figure 2 depicts the pre-operative shortened state of most muscles except for the longer-than- normal excursions of the adductor brevis gluteus medius. The post-operative analysis indicates more normal data for all of the muscles.

DISCUSSION

The documented use of muscle-length estimation for clinical gait analysis is fairly limited [6][7][8][9]. The difficulty in providing accurate estimates of muscle length is because skeletal databases have limited subject-size and demo-graphies, difficulty in accurately monitoring underlying skeletal movement, and the gross estimation of muscle origin, insertion and pathway. Straight-line approximations of muscle length result in in underestimation of absolute length Our laboratory is currently improving the level of accuracy in estimating muscle pathways through developing a graphical user interface to further enhance the NIH database [3], and volumetric reconstruction of MRI images.

The sensitivity analysis indicated that a small error in origin and insertion estimates can result in as much as 5.0% absolute error, depending on the muscle involved. The absolute change in muscle length varied between 60% and 100% of the introduced error. Minimization of estimate-error is important as indicated by this case-study, where absolute muscle length differences between the subject and normal profiles were as little as 6%, and changes between pre- and post-operative lengths were as little as 12%.

The application of this technique would provide more information than current pre- and post-operative studies in clinical gait analysis and, if used routinely, has the potential to quantify the change in muscle length for a range physiotherapy treatments, and orthopaedic procedures. This was demonstrated using results from a single-subject case-study of a child with spastic diplegic cerebral palsy, which indicated changes in muscle excursion as a result of a surgical procedure.

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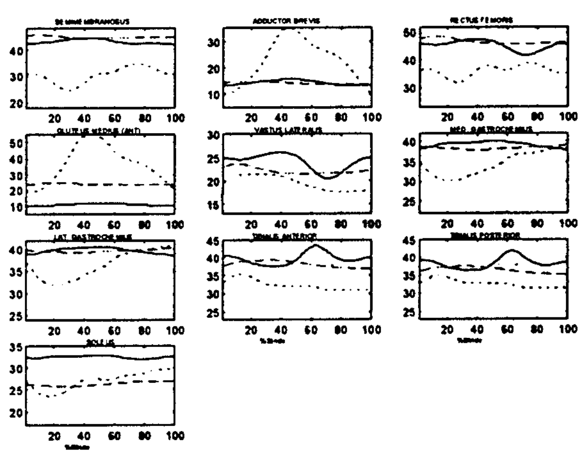

Figure 2: Case-Study - Muscle length [cm] vs. % stride (stance/ swing) for a child with spastic cerebral palsy pre- and post-dorsal rhizotomy surgery and a normalized able-body profile.

___ normal
..... pre-op
- - - post-op

Muscle Power Relationship in Able-bodied Gait

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INTRODUCTION: Although gait asymmetry has been well documented (Sadeghi et al. 1997) little is known about propulsion and control tasks performed by each limb and how these are managed between the lower limbs. Our understanding of able-bodied locomotion improves by studying the interaction among the many biomechanical gait descriptors (Lasko et al. 1990., Olney et al. 1994., Andrews et al 1996). We postulate that propulsion and control functions during able-bodied walking can be explained in part by the relationship between the gait parameters developed within a limb and that these tasks are responsible for the observed gait asymmetry.

PURPOSE: The purpose of this study is to demonstrate that limb propulsion is mainly associated to the interaction between specific muscle power generation bursts developed throughout the stance phase while control is mainly achieved by the Contralateral limb through power absorption bursts. Furthermore, these muscle power generation and absorption activities are not independent but inter-limb related.

METHOD: For two consecutive gait cycles, the gait of nineteen right handed and leg dominant subjects with average age (25.3±4.1 years), height (1.77±0.057 m) and mass (80.6±13.8 kg) was assessed using an eight camera video-based system (Expert Vision 3D system; Motion Analysis Corporation, Santa Rosa, California) synchronized to two AMTI force plates (Advanced Mechanical Technology, Inc., Newton, Massachusetts). Video data were recorded at 90 Hz while the synchronized force data were sampled at 360 Hz. The Direct Linear Transformation software of the Motion Analysis Expert Vision software was used to reconstruct the image markers into three-dimensional coordinates. Noise in video and force data were reduced by means of a fourth order zero-phase lag Butterworth filter, having a cut-off frequencies of 6 Hz and 30 Hz respectively (Winter, 1990). The temporal and phasic gait parameters were determined from the force plate and video data. The muscle powers and their related mechanical energy were calculated at each joint and in each plane of the lower limbs by means of the inverse dynamic technique. The Principal Component Analysis method was applied to reduce and classify 44 power related gait parameters for each limb (Sadeghi et al.1997). The interactions within a limb was determined by the relationship between any two power and energy pairs using Pearson correlation. Then, a Canonical Correlation Analysis was performed to establish the best linear correlation equation between all the right and left limb data sets. The first Canonical Roots (CR) which presented the most significant correlation coefficient between the weighted sum scores between propulsion and control parameters were chosen for further analysis ($R = 0.83$, $p<0.0001$).

RESULT and DISCUSSION: The 3-D muscle powers curves developed at the right hip, knee and ankle with their respective power peak activity were discussed in detailed by Allard et al. (1996), Eng and Winter (1995) and Sadeghi et al. (1997).

For the right and left limb, the Pearson correlation method was applied between all pairs of the gait parameters to determine how how their interactions contribute to forward progression.

The correlation for the first canonical root between the right and left limb data sets is shown in Fig 1, The scatter plot indicates a high positive linear contribution of the highest weighted values having a coefficient canonical root of 0.83 ($p < 0.001$). This supports a strong relation between the attributed functions of right limb propulsion and left limb control measured in two consecutive steps. The canonical loading values of the first canonical root are for the right and left limbs reflect the relative importance of each parameter within a data set with respect to those of the contralateral limb were used to interprete the nature of the interaction on gait parameters.

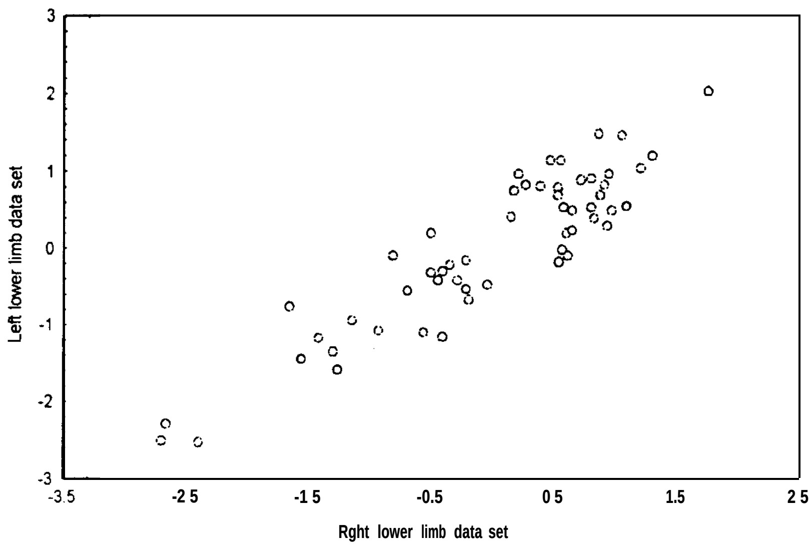

Fig 2: The first canonical correlation for the right and left limb interact having a coefficient canonical root of 0.83 (p<O.OOl).

CONCLUSION: The purpose of this study was to demonstrate that propulsion of the lower limb was associated to the interaction between muscle power bursts developed throughout the stance phase while control was achieved by the contralateral limb. For the right limb, propulsion was an activity initiated by the hip shortly after heel-strike and maintained throughout the stance phase. These results do not support the hypothesis that the ankle is a major contributor to forward progression. Control was the main task of the left limb as evidenced by the power absorption bursts at the hip and knee. The left limb power generations were generally secondary to control activities or possibly to correct for the right limb propulsion. Canonical Correlation Analysis was used to determine the interactions between the right and left lower extremity gait data sets. Inter-limb interaction further emphasised the functional relationship between forward progression and control tasks developed by each limb and highlight the importance of the frontal and transverse plane actions during gait.

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Head and Trunk Stabilization Strategies During Foward and Backward Walking in Healthy Adults

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INTRODUCTION

Human locomotion is a rhythmic activity that induces corresponding rhythmic oscillations of the trunk and the head [1]. An analysis of these oscillations in the antero-posterior and lateral planes of movements may provide information about how the nervous system deals with an increase in the level of equilibrium difficulty. The aim of the present study was to assess the head and trunk equilibrium strategies while walking forwards and backwards under different conditions. This was done using appropriate segmental anchoring indexes [2]. The balance difficulty while walking has been experimentally increased either with a soft support or with the eyes closed, in order to analyze the consequences of these constraints on speed, head, spine and pelvic range of motion, and stabilization strategies of the considered segments.

METHODS

Eleven healthy subjects, 5 males and 6 females [mean age (±SD): 27.6 (±4.2) years] participated in the study. The subjects were asked to walk forwards or backwards, either on a foam rubber support or on the hard ground, and with eyes open or closed. The subjects' locomotor movements were recorded using the E.L.I.T.E. system [3]. 3D-kinematic measurements of seven reflective markers, taped onto the skin over the head (right mastoid), the processus spinous of C7, T6, T12, L3, L5 and the right posterior iliac spine were obtained by means of two video cameras. From these data, the lateral angular movements of six segments around the antero-posterior axis (roll) were calculated: head segment (H) between head and C7; first thoracic segment ($_1T$) between C7 and T6; second thoracic segment ($_2T$) between T6 and T12; first lumbar segment ($_1L$) between T12 and L3; second lumbar segment ($_2L$) between L3 and L5 and pelvic segment (P) between L5 and pelvis. For each walking condition, three parameters were calculated to assess the head and trunk equilibrium strategies. First, the walking speed, which was obtained using the marker taped on the fifth lumbar vertebra. Second, the absolute angular dispersion around the roll axis of the different segments were computed each 10 ms during a trial. The mean dispersion gives a first indication of the absolute angular oscillations of a given segment in the frontal plane while walking. Third, the normalized anchoring index (Al) was used to compare the stabilization of a given segment with respect both to external space and to underlying anatomical segment [2]. The results of 2 to 4 trials were averaged in each experimental condition and subject. Descriptive statistics were calculated for all variables and a three-way repeated measures analyses of variance [2 (backward vs forward) X 2 (foam vs hard ground) X 2 (eyes open vs eyes closed)] were used to assess differences between the walking conditions. When the ANOVAs revealed significant differences ($P < 0.05$), the paired t-test with adjusted probability values were then performed to depict difference between the data.

RESULTS

The subjects walked at a faster speed when their eyes were open than closed ($P<0.001$). The highest speed (mean ± SD; $1.10 \pm 0.21 ms^{-1}$) was observed while walking under natural conditions (walking forward on hard ground, eyes open) whereas the slowest walking speed (mean± SD; $0.79 \pm 0.15\ ms^{-1}$) was observed for the most unusual condition (walking backwards on a foam, eyes closed). In general, the absolute angular dispersion around the roll axis for the head and trunk segments revealed that walking backwards reduced the angular dispersion of the spine segments; the smallest and highest angular dispersions were obtained for walking backwards on a hard ground, eyes open or closed, and walking forwards eyes closed on a foam rubber support, respectively. The absolute angular dispersions of the head and the pelvis did not vary significantly with any factors of the locomotor tasks with lateral oscillations lower than 2° for all conditions. For the trunk segments $_1T$, $_2T$ and $_1L$, the angular dispersion vary significantly with both the visual factor and the ground surface ($p<0.05$). The roll anchoring index (AI) associated with the lateral body oscillations while walking showed that walking backwards, instead of forwards, had, in general, little or no effect on the preferred stabilization strategy. Results also indicated that in many conditions, the H segment showed a positive roll AI indicating a good frontal stability of the head in space. The head stability in space tended to increase when the subjects walked on a foam rubber. By contrast, the AI of spinal segments showed a shift towards negative values indicating stabiion on the underlying segments. In particular, the good stabilization in space of the first lumbar segment observed while walking forwards disappeared during backward walking. Closing the eyes further decreased the AI of all spinal segments when the subjects walked on the foam rubber. Increasing locomotor difficulty thus induced generally an "en bloc" functioning of the spinal segments, and especially while waking backwards.

DISCUSSION

This study indicated that walking backwards generally reduces rather than increases the trunk angular ***ranges*** of motion in roll. This may simply be due to the decrease in the walking speed, the first strategy to occur when encountering locomotor diiculty. Similar decrease in locomotor speed has already been described in humans while walking on a horizontal ladder or a narrow beam [1]. The fact that head and pelvis ranges of motion in roll did not display any change with the experimental condition is a first indication of their lateral stabiion in space while walking. Although there was no significant difference between head anchoring indexes with eyes open and closed in any experimental condition, these indexes became significantly di&rent from zero without vision, which was not the case with vision, except while walking backwards on the soft support (the most unusual situation). There was thus no increase in intensity of the head stabilization in space without vision, as was previously reported while walking on a narrow beam [2], but an increase in the efficiency of this head stabilization strategy (less variability). This result confirms that this strategy is specifically aimed at controlling balance, since locomotor equilibrium difficulty is increased with eyes closed. At the spinal level, walking backwards mainly suppresses the stabilization in space of the first lumbar segment and induces "en bloc" functioning of the other spinal segments, although locomotor speed was reduced. This decrease in the efficiency of segmental stabilization at the spinal level while walking backwards as compared to forward walking may be mainly due to the fact that backward walking is less familiar and trained than forward walking. This supports that segmental lateral stabilization strategies while walking need to be trained, especially during ontogenesis, in order to be well mastered and efficient.

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Correlation Between the Neurological Findings and the Thermographic Changes of Fingers in Diabetes Mellitus

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INTRODUCTION

The cold water immersion test has been used for indicating peripheral circulation disorders and disorders of the autonomic nervous system in diabetes mellitus. To clarify the effect of autonomic nerve function, changes in the temperatures of both hands were analyzed following immersion of one hand in cold water.

METHODS

Thermographic measurements were performed in an air-controlled room(25℃, 45%, no wind) on both hands of 60diabetesmellituspatientsand 12normal controlsafteronehand (the left hand) was immersed in 15°C water for 3 minutes. And we calculated the corelation rates between the thermographic changes [illegible] neurological data(nerve conduct [illegible] autonomic nerve function and disease severities).

RESULTS/ DISCUSSION

The finger temperatureontheimmersed (left) side of the diabetes mellitus patients recovered more slowly than that of the normal controls. Whereas the finger temperature of the right (non-immerse) side of uiabetes mellitus patients remained unchanged, the right finger temperature of the normal controls decreased significantly during the immersion. It was concluded that disorders of the autonomic nervous system in diabetes mellitus decreased the temperature change in the side not undergoing immersion. We found the statistical correlation between the temperature of fingers and many clinical parameters.

The Effect of Oral Creatine Supplementation on Muscle Activity (iEMG) in a Vegetarian and a non-Vegetarian Population: a Double Blind Study

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INTRODUCTION:
PCr (phosfocreatine) plays an important role in muscular contraction, especially during intensive short efforts. It acts as a buffer for the ATP/ADP ratio and as a mediator for energy translocation from the mitochondria towards the ATP utilization sites (Meyer et al, 1984). PCr is the limiting factor during maximum efforts. Literature nevertheless is inconclusive about the influence of PCr supplementation upon physical performance. Vegetarians who do not have external sources of Cr, are believed to have low initial muscle Cr levels and are therefore believed to be weak performers in high intensity exercise as opposed to non-vegetarians (Balsam et al, 1993, Casey et al, 1996).

PURPOSE:
This study tries to find out whether there are changes in muscle activity (iEMG) during short-term concentric and eccentric isokinetic work in both vegetarians and non-vegetarians.

METHODOLOGY:
A group of lacto(-ovo-)vegetarians (n=lO) and a group of non-vegetarians (n=lO), matched for age, gender and life-style (non competitive athletes) volunteered to participate and were asked to refrain from caffeinated drinks on the day of the experiment. The KinCom (isokinetic dynamometer) and Active Bipolar Surface Electrodes (Sample Frequency 1000 Hz) were used to measure and calculate integrated EMG of the quadriceps muscles. The subjects were positioned in a sitting 120 degrees angle between trunk and upper leg. A standardized warming up and adaptation procedure was effectuated. The experimental procedure consisted of 5 bouts of 20 repetitions at 3.14 rad/s interspersed with 60 s rest intervals. Under double blind conditions the subjects were given either oral Cr supplementation (20g Cr-monohydrate) or placebo (glucose) for 6 days previous to the lab experiments. A 6 weeks wash-out period was respected before te second supplementation of 6 days period started. The day after finishing the supplementation the tests were performed.

RESULTS:
The mean iEMG parameters calculated form the different bouts of 20 repetitions showed no relation with the supplementation of Cr nor with placebo intervention, in both vegetarian and non-vegetarian groups. There is no statistical significant difference in iEMG after treatment with Cr or Placebo. Non-vegetarians compared with vegetarians do not reveal any significant differences. Body weight in both the vegetarian and non-vegetarian group was not significantly influenced by the Cr supplementation.

DISCUSSION & CONCLUSION:

No studies have been introducing EMG to measure the influence of PCr onto vegetarian an non-vegetarian populations. The present study does not show any differences of iEMG between a vegetarian or non-vegetarian group, both in concentric and eccentric short maximum contractions. These findings do not indicate a substantial shortage of Cr in vegetarions who solely can rely on endogenous biosynthesis of Cr. This EMG study does not confirm the isokinetic torque results where a tendency to produce higher torque values after the intake of PCr in both groups was noticed (Clarijs et al, 1997).

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EMG Signal Parameters and Their Relation to Psychosocial Factors in a Study of Muscle Tension Patterns Among White Collar Workers

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INTRODUCTION

High levels of stress and a high prevalence of shoulder/neck pain had been observed among office employees at a large industrial company. In an attempt to alleviate these problems, an intervention programrne was initiated by the company, union representatives and the occupational health care centre. As it was hypothesised that the ailments in part had a background in adverse muscular tension patterns, monitoring of ElectroMyoGraphic (EMG) activity was undertaken for entire working days.

Recently, much interest has been focused on EMG-gaps, i.e. short periods where the muscle is totally relaxed. Both Veiersted (1995) and Hagg (1997) have found significantly fewer EMG-gaps among female employees with complaints from the neck and shoulder region. Lundberg et al. (1994) have shown that the EMG activity of the trapezius muscle is significantly increased when performing a test contraction under psychologically stressful conditions compared to the same test in a non-stressful situation. In the analysis, parameters representing these load and temporal aspects of the EMG signal were applied.

METHODS

Measurements of myoelectric activity, psychosocial factors at work, neck/shoulder pain and a number of stress related physical and psychological symptoms were carried out. A stress handling course was offered to everyone interested, but those having high stress levels and/or unfavourable trapezius muscle activity were requested to take part (n_i=19). Those who did not participate in the course, were used as controls (n_c=29). The groups did not differ in terms of psychosocial factors and amount of computer work. Six to eight hours of EMG were recorded from each employee before the intervention. In the follow-up, only four hours were recorded, as it was seen that the muscle activity was practically the same when morning and afternoon sessions from the first round were compared. Comparisons between initial and follow-up data were made by comparing morning sessions to mornings and afternoons to afternoons.

Surface EMG were recorded bilaterally from the trapezius descendens using the MyoGuard monitoring device (Sandsjo, 1997), which presents averaged rectified values (ARV) and mean power frequency (MPF) of the muscle activity once every second. The electrodes were placed 2 cm lateral to the midpoint between C7 and the acromion, with an interelectrode distance of 20 mm. At the start of each measurement, a normalisation procedure consisting of four, 20 seconds long, reference contraction (arms held straight and horizontally in the frontal plane corresponding to about 15% MVC) and muscle relaxation recordings, were performed. The normalisation procedure was also repeated at the end of each measurement allowing comparisons of muscle fatigue. All data were normalised, where the most stable reference contraction was used as reference and set to 100% (Reference Voluntarily Electrical activation, RVE).

In order to evaluate if those of the employees taking part in the course, managed to change their muscle activation during work, the outcome of a set of EMG parameters, applied on recordings made before and after the intervention, were compared. Parameters applied included; MeanARV , i.e. mean muscle activity; *MeanRestLth,* i.e. mean length of muscle rest periods; and *RestTime,* i.e. percentage muscle rest-time of total time. The threshold used to discriminate between muscle activity and the muscle at rest was individually set to the mean value of the best, i.e. lowest, initial relaxation registration with 10% RVE added.

RESULTS

The work-load, as measured by *MeanARV, RestTime* and *MeanRestLth, was* significantly higher on the right compared to the left side throughout the recordings (p<O.OOl). Before the intervention, attitude to work (lust/no lust) correlated to *MeanARV* (p<O.Ol) and to *RestTime* and *MeanRestLth (p<0.05).*

Follow-up results showed reduced neck/shoulder pain, lower self-reported stress and less stress related psychosocial symptoms. Among the EMG parameters, significant change was found only in the *RestTime* parameter of the left trapezius ($p<0.05$). *RestTime* did not change in the control group. The groups did not differ in terms of psychosocial factors and no changes in this respect could be observed during the intervention period. Signs of muscle fatigue, both as heightened ARV- and lowered MPF-values, could be found before and after the intervention (Table 1). There was a tendency that the intervention group had less pronounced fatigue signs then the control group after the intervention.

Table 1. Group mean values of the Mean ARV and MPF values from the four reference contractions performed at the *start* of each recording and the *test* contractions performed at the *end* of each recording. The table shows the outcome of these parameters and significance levels *before* and *after* the intervention when tested with a 2-sided, paired student's T-test. The ARV and MPF values were normalised, where the most stable reference contraction of the four performed at the start of each recording was used as *the* reference and set to 100% (Reference Voluntarilv Electrical activation).

		Before intervention			*Afier intervention*		
Group	**Param.**	Reference	Test	D	Reference	Test	D
All (n=48)	ARV right	102%	114%	<0.05	103%	111%	<0.005
	ARV left	102%	101%	-	101%	106%	<0.05.
	MPF right	99.5%	98.0%	<0.05	100%	99.8%	-
	MPF left	99.7%	97.0%	co.00 1	100%	98.6%	<0.0 1
Intervention (n_i=19)	ARV right	105%	114%	-	104%	107%	-
	ARV left	101%	97.0%	-	101%	101%	-
	MPF right	99.3%	97.9%	-	101%	101%	-
	MPF left	99.8%	97.4%	<0.05	100%	99.1%	-
Control (n_c=29)	ARV right	100%	110%	<0.1	103%	113%	<0.005
	ARV left	102%	102%	-	101%	108%	<0.05
	MPF right	99.5%	98.4%	-	100%	99.3%	-
	MPF left	99.5%	96.8%	<0.00 1	100%	98.6%	<0.05

CONCLUSION

The results demonstrates that the used parameters, representing both load and temporal aspects of the myoelectric activity, reflects workload, psychosocial factors as well as the intervention programme, according to the hypothesis.

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Comparison of the Median Frequency of the EMG Power Spectrum During Ramp Versus Step Isometric Contractions of the Biceps Brachii

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INTRODUCTION

The median frequency of the EMG power spectrum as been related to muscle conduction velocity and fiber recruitment. It has also been found to be proportional to muscle force, up to a level of 20-30% (Hagberg & Ericson, 1982) or 80% (Moritani & Muro, 1987) of the maximal voluntary contraction (MVC) when using either step or ramp contractions, respectively. Another finding is that the median decreases during a steady contraction, which has been associated with fatigue (De Luca et al., 1983). It is therefore possible that the discrepancy between studies of the median/force relationship may be due to the time at which the measurement is taken, for step contractions. The objectives of this study were therefore twofold: (1) to compare the median of EMG power spectra obtained during an isometric ramp contraction to those obtained at diierent points in time during a step contraction; and (2) to verify that the decrease in the value of the median is indeed due to fatigue.

METHODS

Ten male subjects, aged 34.0 (± 4.6) years, participated in the study after giving their informed consent. Subjects were seated with their shoulder at 15° abduction, their elbow flexed at 90° and their forearm in mid-pronation. Active bipolar electrodes with a 10 mm inter-electrode distance were placed on the biceps brachii. A *strain* gauge was used to measure vertical force and visual feedback of the exerted force was provided through a computer monitor. Maximum voluntary contraction (MVC) was first measured by taking the highest value from two trials.

Subjects were instructed to produce muscle contractions by following a target trace on the computer monitor, in the following order: (1) two ramp contractions of linearly increasing contraction to 100% MVC (5 seconds). (2) two step contractions to 40, 20, 80, 10 and 60% MVC (8 seconds). (3) two ramp contractions, as in (1).

EMG data were sampled at 2000 Hz then filtered using a l0-500 Hz bandpass filter. The RMS of the signal and the median frequency of its power spectrum were calculated every 50 ms using 256 ms windows. Data for the median and RMS were then sampled for the ramp contractions at points corresponding to 10, 20,40, 60 and 80% MVC. For the step contractions, the time at which the force stopped oscillating was visually determined. This point, the stabilization time (ts), always occurred within 1 second of the subject first reaching the target force level. Median and RMS were then sampled at ts, at ts+3 seconds, and ts+6 seconds.

RESULTS

A good match between the median in ramp contractions and the median at ts in step contractions was observed (Fig. 1). Both measures showed a linear increase with force up to 40-60% MVC. For step contractions the median ts+3 and ts+6 did not vary or decreased with increasing force. Also, a clear decrease of the median with time during 80% MVC step contractions was observed.

Subjects were then divided into 2 groups: subjects that displayed RMS increase during step contractions for most to all trials (fatigue group; n=4); subjects with no RMS increase for none or few trials (no fatigue group; n=4). A good correspondence between the medians measured during ramp contractions and the medians measured at ts during step contractions was observed for the no fatigue group, but not for the fatigue group (Fig. 2). The values of the medians for the no fatigue group were lower than those of the fatigue group.

CONCLUSION

This study has revealed that the median varies during an 8 second step contractions and that the point at which the measurement is taken is important. More importantly, fatigue may not be the only factor causing a decrease in the median observed in step versus ramp contractions. Factors such as muscle fiber type, involvement of synergists or biomechanical properties of muscles may possibly have played a role.

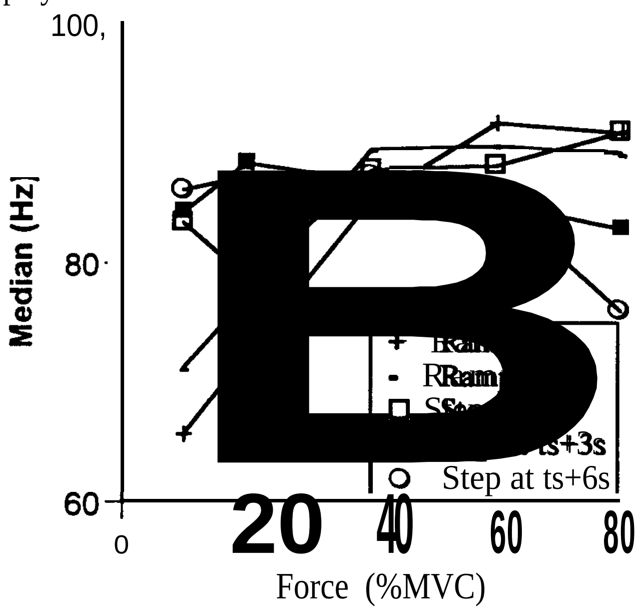

Fig. 1: Median values of EMG power spectra during isometric biceps ramp contractions, and at stabilization time (ts), ts+3 and ts+6 seconds for step contractions; values are averages for 10 subjects.

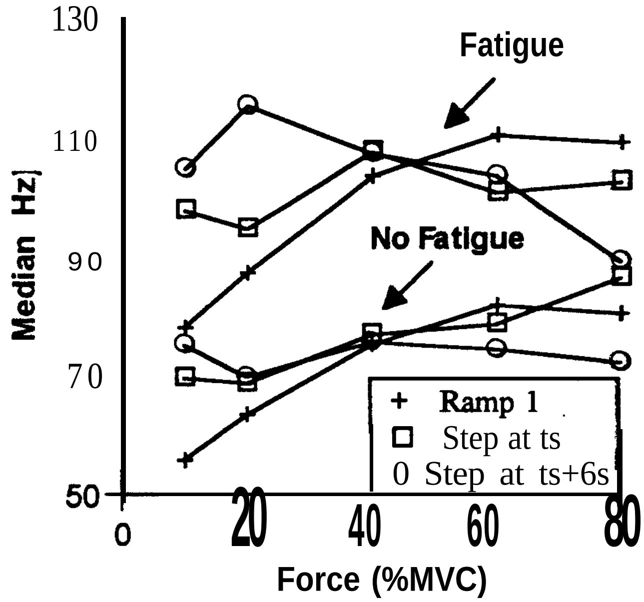

Fig. 2: Median values of EMG power spectra for ramp and step contractions for subjects showing signs of fatigue during step contractions (top, n=4) and subjects showing no signs of fatigue (bottom, n=4).

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