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INTRODUCTION

The paper is not aimed at giving full insight into the up-to-date results in biped gait and posture mathematical modeling. Many researchers were successfully engaged with these problems during the past 15 years. They are mentioned here, though not all of them, but according to the authors' opinion the most important results from the field of mathematical modeling of the biped gait and posture dynamics. The first published paper, treating biped dynamics, appeared at the start of 1969 [1]. After that, other papers emerged, which widened the original model of biped gait at the basis of the same idea [2, 3, 4, 5, 6]. In parallel with biped gait mathematical modeling attempts were originated, trying to implement these results, at least partially, into the synthesis of complex active orthotic devices for producing the basic locomotor act of paralyzed persons [7, 8, 9]. Noted results in modeling the biped gait and posture, and in the synthesis of control algorithms were presented in papers [10, 11, 12, 13, 14], while the corresponding efforts to develop new orthotic systems for lower extremities were reported in [15, 16, 17], confirming a continuity of efforts to find at least partial application for the mathematical models of biped locomotion.

The interesting and very important problem stays, concerning the forming of some criterion for the validation of the biped gait at the basis of dynamic system properties, i.e. its mathematical model. Lack of such a criterion has been noted, which does not rely solely on the kinematic system properties and on the cosmosis of the realized gait. The synthesis of a criterion for gait validation, based on its dynamic data, presents of course a delicate task. This paper does not pretend to answer this question completely. It is intended to present the broad possibilities of the dynamic model of biped gait, or, the broadness and voluminosity of the simulation results concerning symmetrical and asymmetrical gait, as well all the relevant derived dynamic valuations, based on which can be contemplated about some validation criterion, which would be dynamic by its nature. The paper is also intended to demonstrate the purposefulness of using the biped gait mathematical model as a means for the generalization and systematization of the results of numerous investigations, by means of very modern equipment for measuring and recording the kinematical (gait pattern, gait speed, etc.) and dynamical gait parameters (dynamic support reactions, driving torques of the mechanism joints, acceleration of characteristic body points, etc.). Such a statement is primarily based on the earlier shown fact [17], that some simulation results of the dynamic state synthesis of the anthropomorphic configuration with identical parameters (geometry, segment masses of the body, upper and lower extremities, tensors of inertia of the segments), as well with identical imposed kinematical to the lower extremities (gait type), are practically identical with the measured and recorded values on the human being. This conclusion concerns first of all the characteristic and values of the dynamical reaction for in the foot-support contact, as well the driving torques in all the active joints of the extremities and of the body.

MATHEMATICAL MODEL OF BIPED DYNAMICS

In order to perform description of the biped motion dynamics, the known procedure based on the semi-inverse method is adopted [1, 2, 3]. Here are repeated only some basic postulates, on which are based the forming of the mathematical model of biped locomotion. It should be pointed out that the zero-moment point is of great practical significance. It is the conditional name for the instantaneous point of which the total reaction forces are acting, produced by man during gait. The trajectory of the zero-moment points can be recorded in a relatively simple way. Let be the zero-moment point (Fig. 1). According to D'Alembert's principle, the conditions of dynamic equilibrium, called the dynamic connections, are

\[ \begin{align*}
R_{xj} + F_{xj} & = 0, \\
R_{yj} + F_{yj} & = 0, \\
R_{zj} + F_{zj} & = 0.
\end{align*} \]
where $\mathbf{v}_s$ is the vector from point $s$ to the center of gravity of the $i$-th element, $\mathbf{v}_g$ are the natural vector and the moment of the inertial tensor of the $i$-th element; $m_i$ is the weight of particular elements; and $\mathbf{v}_s$, $\mathbf{v}_g$, $m_i$ are unit vectors of the coordinate axes at the Modeling point $s$.

The equilibrium equation with respect to the acting point of the resulting friction can be written as:

$$\mathbf{r} = \sum_{i=1}^{n} m_i \mathbf{v}_s \times \mathbf{v}_g + \sum_{j=1}^{n} \mathbf{f}_i \times \mathbf{v}_g = 0 \quad (2)$$

where $\mathbf{r}$ is the vector from the zero-moment point to the penetration point of axis $\mathbf{g}$ on the contact surface between the foot and ground, and $\mathbf{f}_i$ is the unit vector of axis $\mathbf{g}$.

Equations (1) and (2) represent the dynamic equilibrium conditions written in general form for an anthropomorphic system with fixed arms. If free arms were considered, this model would have to be extended by as many second-order differential equations as the "arms" have degrees of freedom. By fixing the arms, only three equations of dynamic connections are possible. Since the number of degrees of freedom of the anthropomorphic mechanism is considerably greater than three, for the (n-3) coordinates the motion program is prescribed, and the remaining coordinates are found from the equations of dynamic connections (1), (2).

In general, the above-described system of differential equations can be written as a linear form of generalized accelerations and a square form of relative angular velocities.

$$\begin{align*}
\mathbf{F} & = \mathbf{M} \ddot{\mathbf{q}} + \mathbf{C} \dot{\mathbf{q}} + \mathbf{Q} \\
\mathbf{Q} & = \mathbf{r} \times \mathbf{F} \quad (3)
\end{align*}$$

where $\mathbf{F}$, $\mathbf{M}$, $\mathbf{C}$, $\mathbf{Q}$ are functions of generalized coordinates. The prescribed $3$ coordinates are separated from the set of coordinates $\mathbf{q}$, and the $\mathbf{Q}$ coordinates computed from dynamic connections $\mathbf{F}$, $\mathbf{M}$, $\mathbf{C}$. Equations (1), (2) can be written in a concise form as:

$$\begin{align*}
\mathbf{F} & = \mathbf{M} \ddot{\mathbf{q}} + \mathbf{C} \dot{\mathbf{q}} + \mathbf{Q} \\
\mathbf{Q} & = \mathbf{r} \times \mathbf{F} \quad (4)
\end{align*}$$

where $\mathbf{F}$, $\mathbf{M}$, $\mathbf{C}$, $\mathbf{Q}$ are vector coefficients depending on $\mathbf{q}$, $\dot{\mathbf{q}}$, $\ddot{\mathbf{q}}$, $\mathbf{F}$. Eq. 4 with the imposed repeatability conditions $[1, 2]$: $\mathbf{Q}(t) = \mathbf{Q}(t + 1)$

$$\phi(t) = \phi(t + 1) \quad (5)$$

where $\phi(t)$ is the step period, gives the compensation symmetry of the single-support part of the body at the basis of perscribed synergy of lower extremities (gait pattern). Since the adopted

**SIMULATION RESULTS**

Based on such a program package, some characteristic results for gait upon ground level are presented. The parameters of the mechanical configuration in Figure 2 are given in Table 1.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Step length</td>
<td>0.5 m</td>
</tr>
<tr>
<td>Step time</td>
<td>1 s</td>
</tr>
<tr>
<td>Walking speed</td>
<td>1 m/s</td>
</tr>
</tbody>
</table>

Simulation has been performed with various gait parameters. As basic gait parameters were adopted:

- Step size (length)
- Step time (duration)
- Cadence
- Overlapping
- Double-support phase duration

Both symmetrical and asymmetrical gait were treated. In that way all basic gait parameters concerning only symmetrical gait types upon level ground will be presented. Also, asymmetrical gait cases are given with single-support and double-support phases. All analyses have been performed for prescribed values of gait parameters desired type of leg motion). Illustrated in Figure 3.

**REFERENCES**


T.I. Mass and Inertia DATA

<table>
<thead>
<tr>
<th>NR. of</th>
<th>MASS</th>
<th>LENGTH</th>
<th>Proper moments of inertia (kg·sec^2·m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>e</td>
<td>0.152</td>
<td>0.121</td>
<td>0.00096</td>
</tr>
<tr>
<td>e</td>
<td>0.612</td>
<td>0.397</td>
<td>0.00615</td>
</tr>
<tr>
<td>e</td>
<td>1.090</td>
<td>0.431</td>
<td>0.01725</td>
</tr>
<tr>
<td>e</td>
<td>1.222</td>
<td>0.192</td>
<td>0.02000</td>
</tr>
<tr>
<td>e</td>
<td>2.140</td>
<td>0.417</td>
<td>0.13400</td>
</tr>
<tr>
<td>e</td>
<td>0.640</td>
<td>0.785</td>
<td>0.00947</td>
</tr>
</tbody>
</table>

Fig. 1. Zero Moment Point (ZMP)

Fig. 2. Adopted Mechanical Configuration

Fig. 3. Adopted Trajectories of Legs

Maximal values of driving torques and mechanical work in the function of gait speed for different stride and T=1.5sec (symmetrical gait, single support phase)

<table>
<thead>
<tr>
<th>Case</th>
<th>t(sec)</th>
<th>∆x(m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>I</td>
<td>0 &lt; T/2</td>
<td>0.0</td>
</tr>
<tr>
<td>II</td>
<td>0.3 &lt; T/2</td>
<td>0.05</td>
</tr>
<tr>
<td>III</td>
<td>0.5 &lt; T/2</td>
<td>0.05</td>
</tr>
<tr>
<td>IV</td>
<td>0.7 &lt; T/2</td>
<td>-0.02</td>
</tr>
<tr>
<td>V</td>
<td>0.9 &lt; T/2</td>
<td>0.0</td>
</tr>
<tr>
<td>Y</td>
<td>0.61 T/2</td>
<td>0.35</td>
</tr>
</tbody>
</table>

Maximal values of vertical component of reaction force

Fig. 4. ZMP displacement

Maximal values of driving torques and mechanical work in the function of gait speed for different cadence and S=0.4 (symmetrical gait, different duration of double support phase)

<table>
<thead>
<tr>
<th>Case</th>
<th>∆x(m)</th>
<th>∆y(m)</th>
<th>∆y/∫L/sec</th>
<th>∆z(m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>I</td>
<td>0</td>
<td>0.15</td>
<td>0.0</td>
<td>0.0</td>
</tr>
<tr>
<td>II</td>
<td>0</td>
<td>0.15</td>
<td>0.0</td>
<td>0.0</td>
</tr>
<tr>
<td>III</td>
<td>0</td>
<td>0.15</td>
<td>0.0</td>
<td>0.0</td>
</tr>
<tr>
<td>IV</td>
<td>0</td>
<td>0.15</td>
<td>0.0</td>
<td>0.0</td>
</tr>
<tr>
<td>V</td>
<td>0</td>
<td>0.15</td>
<td>0.0</td>
<td>0.0</td>
</tr>
</tbody>
</table>

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Maximal values of vertical component of reaction force

Fig. 5. Compensating actuator in sagittal plane

Fig. 6. Total mechanical work

T: Step period sec, p: % of double support phase duration; d: Semi-distance of feet in frontal plane

Case: T = 1.5, S = 0.8

Fig. 7. Maximal values of vertical component of reaction force

Fig. 8. Compensating actuator in sagittal plane

Fig. 9. Total mechanical work

Fig. 10. Maximal values of vertical component of reaction force

Maximal values of driving torques and mechanical work during full step in the function of cadence difference between the left and right leg (AT) (asymmetrical gait, constant stride)

<table>
<thead>
<tr>
<th>Support phase of left leg</th>
<th>Support phase of right leg</th>
<th>AT</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.75 - 0.40</td>
<td>0.75 - 0.40</td>
<td>0.03</td>
</tr>
<tr>
<td>0.75 - 0.40</td>
<td>0.75 - 0.40</td>
<td>0.03</td>
</tr>
<tr>
<td>0.75 - 0.40</td>
<td>0.75 - 0.40</td>
<td>0.03</td>
</tr>
<tr>
<td>0.75 - 0.40</td>
<td>0.75 - 0.40</td>
<td>0.03</td>
</tr>
<tr>
<td>0.75 - 0.40</td>
<td>0.75 - 0.40</td>
<td>0.03</td>
</tr>
</tbody>
</table>

S: Support phase of left leg; L: Support phase of right leg

Case: T = 1.5, S = 0.8

Fig. 5. Compensating actuator in sagittal plane


**M. Vukobratović: Biomechanics of Bipedal Locomotion**

![Graph](image)

Fig. 11. Compensating actuator in sagittal plane

![Graph](image)

Fig. 12. Compensating actuator in sagittal plane

![Graph](image)

Fig. 13. Total mechanical work

![Graph](image)

Fig. 14. Maximal values of vertical component of reaction force

### Table: Maximal values of driving torques and mechanical work during full step in the function of stride difference between the left and right leg (asymmetrical gait)

<table>
<thead>
<tr>
<th>Support phase of left leg</th>
<th>Support phase of right leg</th>
<th>ΔS</th>
</tr>
</thead>
<tbody>
<tr>
<td>t(sec)</td>
<td>ZMP low</td>
<td>t(sec)</td>
</tr>
<tr>
<td>0-0.75</td>
<td>0.8</td>
<td>0.75-1.5</td>
</tr>
<tr>
<td>0-0.75</td>
<td>0.8</td>
<td>0.75-1.5</td>
</tr>
<tr>
<td>0-0.75</td>
<td>0.8</td>
<td>0.75-1.5</td>
</tr>
<tr>
<td>0-0.75</td>
<td>0.8</td>
<td>0.75-1.5</td>
</tr>
</tbody>
</table>

J. Stefan Institute, E. Kardelj University, Ljubljana, Yugoslavia

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**ANALYSIS AND SYNTHESIS OF PARETIC GAIT - MEASUREMENTS AND ELECTRICAL STIMULATION**

Miroslav Klajčič, Matija Malešić, Uros Stašič

**INTRODUCTION**

For several years research on application of functional electrical stimulation (FES) to paretic patients enhancing externally evoked functional movements in walking has been going on in Ljubljana. For this purpose stimulators, together with the measurement equipment, were developed enabling the introduction at new therapeutic and orthotic method of rehabilitation. New methodologies have been always starting with feasibility studies comprising patient indications and selection, qualitative medical evaluation together with measurements of the biomechanical parameters, EMG activity etc., representing the analysis of the movements produced by the stimulation of the neuromuscular structures, and synthesis of externally evoked ones. The effects of FES were compared to the movements without stimulation, and further research work is planned on larger population of paretic patients to assure statistically significant improvements.

**METHODOLOGY AND INSTRUMENTATION**

For the rehabilitation of paretic gait single and multichannel electronic stimulators have been designed.\(1,2\) Starting in the early phase immediately after the lesion, sophisticated six channel devices have enabled initiating gait pattern, strengthening muscles and establishing antigravity support through six-channel stimulation therapy. Simple stimulators have been introduced for the orthotic use. After the conclusion of therapy patients have been issued single channel orthotic devices for the use in their home environments.\(3\) Surface as well as implantable technology have been developed for this purpose.\(4\)

Effects of stimulation have been measured and evaluated during and after the FES treatment. Planar goniometric system measuring angles in the six leg joints have been designed\(5\) together with ground reaction measuring system including force-shoes and crutches\(6\) have been designed and used in the measurements (Fig. 1).
**U Stanišić: Analysis and Synthesis of Paretic Gait.**

Specially developed software package has enabled on-line computing of the measured data on a mini-computer. (7).

**RESULTS**

The effects of multichannel electrical stimulation of the main leg muscles are shown on Fig. 2.

**Fig. 2:** Geoniograms of joint angles of hemiparetic gait (full lines) with and without stimulation together with normal ones (dashed line) and stimulation sequences.

With the simultaneous stimulation of m. tibialis anticus, m. biceps femoris, m. quadriceps, and m. glutaeus maximus, more normal angles and better repeatability have been obtained (8).

Common graphical representation of the measured data has been determined using a statistical approach over least 30 steps. Different computer print-outs are accessible to explicit the measurements. Time plots of vertical component of ground reaction force under both feet and crutch are shown in the upper part of Fig. 3 with the mean values (full line) and standard deviations (dotted line), while trajectories of zero moment point under both soles are shown in the lower part of Fig. 3 with mean coordinates (full line) and standard deviations in predetermined phases of stance (rectangles) (9).

**Fig. 3:** Ground reaction forces and their spatial distribution under the feet and crutch during gait.

Similar print-outs have been designed for geoniograms of joint angles (Fig. 4) with the mean values (full line) and standard deviations (dotted line), instead of commonly used foot-switch functions, force-basograms have been used for synchronisation (Fig. 4 - bottom diagrams) (9).

**Fig. 4:** Geoniograms of hip, knee, and ankle synchronised by force-basograms for left and right leg.

**CONCLUSION**

The approach to the analysis and synthesis of paretic gait has been discussed. The developed measurement instrumentation has been evaluated as useful tool for the estimation of gait and determination about the therapy. Besides, the stimulators have been designed for both therapeutic and orthotic applications during the rehabilitation of gait.

**REFERENCES**


A Kralj: Biomechanics aspects of External Control of Gait

Patient locomotion disorders may be classified in two ways:

A) Neuropathic patients (polio, cerebral palsy, multiple sclerosis, spinal cord injury) who have pathologically organized neural control of locomotion but intact natural structures. But also symptomatic patients (seizures, lower motor neuron or peripheral nerve lesions) but with intact normal neural control.

It is clear that the B group of patients is philosophically the least demanding group for external control. Here the goal is to substitute a biomechanical device for the missing structure and to use the patient’s natural control to achieve the desired function. In the A group, the specific pathologic organization must be understood and the transfer function of the neurocontrol known before the external control can be applied.

The external control system block C is composed of the external controller (1), the proprioceptive information pick up and processing unit (2), the voluntary control signal processing unit (3), and sensory feedback display (4). It is the last unit that offers many prospects for further development. At this point it is now known how the patient utilizes this information in the integration of the total control system. The biomechanical patient is to be analyzed by mechanical bracing, FES, or a combination of both. From a biomechanical aspect it is obvious that mechanical bracing provides support but does not use the “system” output effectively. It is more desirable to use the output of the system and to accomplish the restored locomotion. In this regard FES holds great promise as it provides the additional possibility of affecting the internal organization of the nervous system pathology and perhaps a much improved output beyond what would be expected from the initial goal. More on this later.

TASK ACCOMPLISHMENT

Without an argument from anyone in the field a rather simple approach to topics to ensure progress in the restoration of locomotion via external control could be produced. Given the uniqueness of each patient, the multiple disciplines involved in this complex problem and the rapidly developing technology the attempt to do so seems almost impossible. Yet in spite of this complexity the biomechanics approach allows a definite goal to be assessed and the aim of the success to be pursued. Numerous programs have been made in spite of the complexity and it is important for all critical tasks of common interest to a broad group of researchers. These come quickly to mind.

1. Understanding the biomechanical events in normal and pathological gait.
2. Understanding the neurophysiological bases for the biomechanical events.
3. Development of the technology for the external controller to achieve the integration of the biomechanical events and underlying neurophysiology for locomotion.

Despite our increasing understanding of each of the three task areas just cited, something seems to be lacking. In the process of patient assessment for the restoration of locomotion the physician is not at a loss for the availability of detailed data with regard to motion analysis or ever increasing insights into the neurophysiological mechanisms. The physical finds him/herself unable to use this information in such a way as to lead to clinical interpretation (9) specifically in answer to the question: "What does the individual patient performs a given way and the clinical researcher (it, why)? Because for a given motion deficiency of a given patient the number of possible medical reasons is responsible. For example: Inadequate analysis and interpretation of individual patient data is the result of lack of clear clinical interpretation. The clinician is not aware of the clinical condition of the patient’s gait. This is the major problem of how to use this knowledge will remain. We are at present limited in our external input capability to the patient as a system because we lack a thorough understanding of how the nervous system is integrated to produce normal locomotion. We know little about the organization in pathological conditions. Therefore selective correction and application of an external control is problematic. As a final comment to this area of physical assessment the question of an “optimal locomotion model” for a neuropathic locomotion disorder is still debatable, namely should the criteria be a comparison to the normal or in better to achieve maximal performance in a broader sense. One approach, FES, has produced particularly practical solutions in external control because its goal has been to achieve maximal performance in this broader sense.
FES AS AN EXTERNAL CONTROL

Biomechanical aspects of external control using FES are better recognized by illustrating the results of control obtained thus far. Liberson (1) demonstrated that FES can be used for provoking direct muscle contraction for functional use.

Figure 7, redrawn after (7) clearly illustrates that after only 30 minutes of FES for flexion reflex elicitation at the peroneal nerve resulted in improved joint compliance and diminished EMG of dorsal flexors. FES is an effective means for clonus reduction in patients suffering from cerebrovascular, cranio-cerebellar and other brain incidents as well as in multiple sclerosis patients (6, 8). It is interesting to note that in M.S. patients, spinal cord FES (8) and peripherally applied subcutaneous nerve stimulation do result in similar results of depression of spinal reflex excitation observable in clonus inhibition for up to 3 hours (6). The author of this work also demonstrated that FES applied to the wrist (median or radial nerve) can suppress ankle clonus and that the probable mechanism mediating the effect is centrifugal inhibition via long-loop reflexes. This is from a biomechanical aspect an interesting result showing that even the CNS organization can alter and hence controlled to a large extent using different input loci. FES can also produce therapeutic effects (19) and produce reorganization of the neuromuscular control as seen shown in homplastic patients (5). From a biomechanical point of view it is clear that FES is developing into an important external control means capable of altering in

a broad sense the neuromuscular control. The will enable the rehabilitation of neuromuscular patients by means of a new emerging field—restorative neurology. Therefore, continued research must be conducted to improve the technology of FES.

CONCLUSION

Restorative neurology is an emerging field and FES technology will contribute not only to the clinical restoration of locomotion but also holds forth promise for a tool to provide insight into the basic organization of the central nervous system's organization for locomotion. Much research needs to be done in the development of hardware, implantable batteries, multichannel controllers, etc.; neurophysiological insights are needed such as how to generate afferent FES extension and even eventual direct neural transfer of information. Much is hoped for by the patients who suffer; much more is needed by us who can contribute to the solutions of restoration of locomotion by external control.

LITERATURE


THE USE OF SURFACE EMG (PART II): WHICH PARAMETERS CAN BE USED IN ORDER TO DESCRIBE SURFACE EMG

Kees Boon et al. Which Parameter can be used to quantify Surface EMG

points indicate the d60 and d100 points in the spectrum.

**Characteristic power** (Prel 100): this parameter indicates the relative power (in %) in the spectrum above 100 Hz. Further more parameters are investigated that are based on the parameters already introduced, viz.

- **Reciprocal integration coefficient (RIC):** this parameter is the quotient between the SD of antagonist EMG activity and the SD of agonist EMG activity. This parameter is used, e.g., by Frero (1995).
- **EMG force gradient (EGF):** this parameter indicates the increase of EMG (expressed in the SD of agonist EMG activity) divided by the increase of force (in Newton).

Finally we studied also some parameters that are commonly used, viz.:

- **Integrated EMG (IMEM):** mean absolute EMG per second.
- **Mean top-top amplitude per second (V/100)**
- **Mean number of force peak seconds (NFP):**

**Methods and Materials**

**A subject is seated in a chair.** His untrained arm is held in a horizontal plane at the level of the shoulder, the angle between fore arm and upperarm is 90 degrees. Force transducers are placed at the wrist. EMG is measured with several different types of electrodes. A Medelec amplifier (M66) is used. The EMG is fed to a digital computer. The signal is sampled with a sample rate of 1000 Hz. During an interval of 2.048 seconds (12 bit ADC) the spectrum is calculated by means of a modified Blackman-Tuckey algorithm in which a Poppalaria window is used (band with 250 points).

**RESULTS**

During a pilot investigation the EMG activity of biceps and triceps is registered with different electrodes (one subject). First one electrode is placed at the distal tendon of the biceps and the other is shifted towards the belly. In this case an increase of SD and a decrease of NFP is found. After this investigation a standard bipolar Medelec surface electrode (EL211M) is used. Again the electrode is shifted from tendon to belly (grounding at the wrist). Now a maximum of SD is found in a region of 3-5 cm from the distal tendon. The parameters NFP, d60, d100, F0, Prel 100 and Prel 100 show a minimum in this region. The electrode configuration indicated in the Figure 2 coincides with a distance of about 3 cm (centre electrode to tendon). After this investigation a group of 10 healthy subjects (aged between 20-30 years) is investigated.

**Methods**

- **NFP:** Number of force peak seconds
- **SD:** Standard deviation of force
- **Prel 100:** Preliminary 100
- **F0:** Fourier zero
- **d60:** D60
- **d100:** D100
- **IMEM:** Integrated Mean EMG
- **RIC:** Reciprocal Integration Coefficient
- **EGF:** EMG Force Gradient

**Figure 1**

**Figure 2**

**Figure 3**

**Figure 4**

**Table 1:** Variability between 10 subjects.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Prel 100</th>
<th>Mean (in % of mean)</th>
</tr>
</thead>
<tbody>
<tr>
<td>d60 (NFP)</td>
<td>74</td>
<td>30</td>
</tr>
<tr>
<td>d100 (NFP)</td>
<td>72</td>
<td>30</td>
</tr>
<tr>
<td>IMEM (NFP)</td>
<td>90</td>
<td>30</td>
</tr>
<tr>
<td>RIC (NFP)</td>
<td>74</td>
<td>30</td>
</tr>
<tr>
<td>EGF (NFP)</td>
<td>72</td>
<td>30</td>
</tr>
</tbody>
</table>

**METHODS AND MATERIALS**

- A rather linear relationship is found between IEMG, VLT and SD. Usually SD = k*IEMG with k = 0.0413 (force between 8 and 20N) and SD = k*IEMG with k = 0.00667. F0 is a rather independent parameter: it is related to the firing frequencies of motor units. d60 is considered as less important because it fluctuates too much between individuals. NFP and Prel 100 are very reproducible. Prel 100 can change significantly so will be shown in part II.

**REFERENCES**

J. Boucher Effects of Ground Electrode Position on Electromyographic Potentials

TABLE 1

<table>
<thead>
<tr>
<th>Electrode Position</th>
<th>AMPLITUDE (mV)</th>
<th>FREQUENCY (Hz)</th>
</tr>
</thead>
<tbody>
<tr>
<td>MVC</td>
<td>0.377</td>
<td>106</td>
</tr>
<tr>
<td>10% MVC</td>
<td>0.377</td>
<td>106</td>
</tr>
<tr>
<td>50% MVC</td>
<td>0.377</td>
<td>106</td>
</tr>
<tr>
<td>10% BONE</td>
<td>0.797</td>
<td>106</td>
</tr>
<tr>
<td>50% BONE</td>
<td>0.797</td>
<td>106</td>
</tr>
<tr>
<td>MUSCLE</td>
<td>1.154</td>
<td>111</td>
</tr>
</tbody>
</table>

The mean amplitude and frequency for all experimental conditions, Maximum voluntary contraction (MVC), 10% of MVC, ground electrode on a bony area (BONE) and on the belly of the muscle (MUSCLE).

DISCUSSION

The significant difference in myoelectric potential amplitudes between contraction levels was to be expected and is of less importance in the present study. The significant differences in potential amplitudes and frequencies associated with the electrode positions represent an important finding, which is in contradiction with Ramirez and Gluzenstein’s (1981) advice. These authors stated that the ground electrode location is not critical. According to the results the two electrode configurations represent two different systems (Boucher, 1981).

The one with the ground electrode on the bony area behaved as a narrow bandpass system, whereas the second one responded like a very broad bandpass system revealing a different peak frequency and different harmonics (Boucher, 1981).

It is obvious that this paper is only a first step toward the understanding of the effects of ground electrode position in surface EMG, and even though the equipment was rigorously calibrated, the possibility that the results were due to measurement artifacts is not yet eliminated. How, however, such drastic differences clearly revealed by the peak frequency of the bony area and secondary peaks at 120 Hz and secondary peaks at 50 Hz will be the need for further research on the topic of EMG nocod.
A Noujim et al. Surface EMG Spectral Analysis and its Application to Classification

The "Kennem" signal was used to help find the optimal filter order because the normalized error test (1 - VNF/Vp) fails. The predicted signal was subtracted from the original to obtain the difference (remnant) signal. The RMS (Root Mean Square) value was then calculated for the difference signal and its percentage of the original was also calculated. It was found that 20 All-Pole coefficients gave a reasonable segment approximation. The average of each of the 20 coefficients was calculated as a function of the segment number for deriving the number of segments that identify a record. The segment shift was statistically stationary at 500 msec. After 10 segments it was found that the coefficient average started to converge. 10 segments were used to identify the record.

For classification, the Fisher Linear Discriminant technique was used for the 20 x 10 coefficient matrix of each subject. Despite the relatively large dispersion in the model parameters of the same subject under the same recording conditions, major group classification was achieved. With 3 normals and 5 abnormalities sharing the construction of the Fisher Linear Discriminant Vector, 87.3% classification was achieved on a sample of 3 abnormalities and 5 normals using clinical needle electrode diagnostic classification as a reference. This result is demonstrated in Fig. 2.

REFERENCES


Fig. 1: Thirty all-pole spectrum (20 coefficients) as a function of the time.

Fig. 2: Fisher linear discriminant result with 5 normals and 5 normals sharing the construction of the discriminant vector.

Voluntary Contraction). The signal hounase was 10 Hz to 230 Hz where most of the surface EMG energy is located. The signal, sampled at 500 Hz, was found to have a changing spectrum, this is demonstrated in Fig. 1. Stationary segments of 500 msec were subject to linear prediction mathematics to model the surface EMG signal.

The segments were identified by calculating first the All-Pole, Autoregressive model parameters using the autocorrelation method.
M. Marzana Fegina, G. Feggino, L. Ferrari Strambi and G. Valli

INTRODUCTION

On the basis of aphasico-pathologic considerations, discrimination of acute neurogenic lesions from myopathic disorders should be assessed taking into account both the morphology of the motor unit potential (MUP) and the firing rate of the motor unit potential trains (MUPTs). However, in some cases the presence of short and phasic potentials of reduced amplitude in early stages of neurogenic atrophy as well as increased firing frequency in severe myopathies gives rise to misleading EMG patterns.

Recruitment studies are therefore important to differentiate a recent neurogenic lesion from a myopathic one. In the latter case there is a decrease in the density of electrical activity during voluntary contraction, while in the second case a full interference pattern appears already at submaximal contraction.

This work describes the behaviour of a number of Time and Frequency Domain parameters computed from the EMG signal, measured at 20-30-40s of the maximum voluntary effort in the case of myopathic and peripheral neurogenic patients.

MATERIAL AND METHOD

Experimental data have been collected from the Right Biceps Brachii of 14 subjects by means of a concentric needle electrode, using a Medelec MS6, four channel, two sweeps electromyograph. According to the CIDME findings each muscle has been assigned to one of the following two groups: 1) myopathic muscles 5); 2) peripheral neurogenic muscles.

The test was carried out under isometric non-fatiguing conditions with the arm flexed at 90° and the forearm in complete supination to avoid synergic contraction of the antagonist muscles. The force exerted by the contracting muscle was measured at 20°, 30° and 40° of the maximum by means of a dynamometer.

Before acquisition the EMG signal was low-pass filtered at 2000 Hz with an active analog equalizer filter and then sampled at 5000 Hz.

The recorded DEGs have been processed by a PDP 11/34 minicomputer provided with four disc units (two 10 Mybe and two 5 Mybe), an A/D converter, three visual displays as interactive terminals and a parallel printer as output peripheral unit. The experimental EMG signal file is stored in a library of 5 Mybe discs, each channel occupying 192 blocks of 512 byte each, approximately corresponding to 10s temporal epoch while the programs are resident on the 10 Mybe disc which is utilized in the multi-programming mode. The system is operated by the RSX11 OS. The processing programs are written in FORTRAN IV. Processing of the signals was carried out both in the Time and in the Frequency Domain.

a) Time Domain analysis

The mean value and the standard deviation have been computed for the following parameters over twelve blocks of 4096 points, each corresponding to a time epoch of 600 ms:

- number of zero-crossings
- number of positive maxima (T)
- mean amplitude of the rectified signal (A)
- the T/A ratio.

b) Frequency Domain analysis

Each 4096 points block has been subdivided into four blocks of 1024 points. The Fourier Transform has been computed by means of an FFI algorithm with decimation in time. A cosine window has been adopted. The mean value and standard deviation of the power spectrum for the following frequencies were calculated: 20, 30 and 40% of the maximum. Changes of their values from 20% to 30% and from 30% to 40% are given as \( \Delta_{23} \) and \( \Delta_{34} \) respectively.

Pathology refers to myopathic subjects where pathology PN refers to peripheral neurogenic subjects.

Table I summarizes the behaviour of Time Domain Parameters. As expected, when the contraction level is increased from 20% to 30% a larger number of MUs is recruited in myopathic patients whereas at a further increase from 30% to 40% recruitment cannot take place in the myopathic muscle while it occurs in the neurogenic muscles almost to the same extent as before. These phenomena are evident in the behaviour of the zero-crossings and the positive maxima increments \( \Delta_{23} \) and \( \Delta_{34} \).

Table I: Time Domain Parameters

<table>
<thead>
<tr>
<th>SUBJECT</th>
<th>Pathology</th>
<th>(T)</th>
<th>( \Delta_{23} )</th>
<th>( \Delta_{34} )</th>
</tr>
</thead>
<tbody>
<tr>
<td>T.C.</td>
<td>PN</td>
<td>4.60</td>
<td>8.50</td>
<td>7.00</td>
</tr>
<tr>
<td>T.P.</td>
<td>PN</td>
<td>20.00</td>
<td>20.50</td>
<td>20.30</td>
</tr>
<tr>
<td>T.S.</td>
<td>PN</td>
<td>24.70</td>
<td>25.00</td>
<td>25.50</td>
</tr>
<tr>
<td>T.C.</td>
<td>PN</td>
<td>20.00</td>
<td>20.50</td>
<td>20.30</td>
</tr>
<tr>
<td>T.P.</td>
<td>PN</td>
<td>24.70</td>
<td>25.00</td>
<td>25.50</td>
</tr>
<tr>
<td>T.S.</td>
<td>PN</td>
<td>24.70</td>
<td>25.00</td>
<td>25.50</td>
</tr>
</tbody>
</table>

The behaviour of the Frequency Domain parameters is less clear.

The large variances of the parameters as well as the large within-group variances make it difficult to comment on these results.

The reported values for the \( \Delta_{23} \) and \( \Delta_{34} \) increments are generally within 20% of the EMG average value of the parameters. This might support experimental and theoretical findings from other authors (Parker et al., 1977; Larsson, 1975) who provided evidence that the effort exerted by the contracting muscle does not influence the shape of the Power Spectrum.

REFERENCES


M. Maranaza Figini *, L. Ferini Strambi **, G. Ferrigno *** and G. Valli**

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*** I.U.D.O.M. Sezione di Milano, Milan, Italy

INTRODUCTION
Neurogenic pathologies are characterized by the morphology of the Motor Unit Action Potential (MUAP) which can be measured in the affected muscle: long duration polyphasic MUAPs are a sign of acquired peripheral neuropathies whereas giant MUAPs are dominant in motor neuron diseases and chronic peripheral neuropathies. It is important to the clinician to follow the progress of the neurogenic involvement and to check the success of the therapeutic treatment in the course of time especially when the development of the pathological conditions may be difficult to predict, as in hereditary and paramyoplastic polyneuropathies.

A technique able to assess increase or decrease of the number of giant as well as of polyphasic potentials is therefore of large utility. A study of the influence of the MUAP morphology on the compound EMG signal (Maranaza et al., 1981) which makes use of a simulation model, has shown that Frequency Domain Analysis is effective in detecting the presence of giant potentials. In particular the skewness of the Power Spectrum of the simulated EMG pattern increases with increasing number of giant potentials and decreases with increasing number of polyphasic potentials.

This work shows application of the simulated model findings to patients with evolutive neurogenic disease.

BACKGROUND
The model allows simulation of the EMG activity of the Biceps Brachii during moderate effort of short duration under isometric voluntary contraction of the muscle; the Motor Unit (MU) is recognized as the functional unit of the neuromuscular system; the MU activity (De Luca, 1979; Maranaza et al., 1981) is represented by a Motor Unit Action Potential Train (MUAPT). The interspike interval (ISI) is simulated by means of a stochastic variable. The EMG compound signal is obtained as the sum of the MUAPs. Normal amplitude, long duration, triphasic MUAPs were adopted to simulate acquired peripheral neuropathies and motor neuron together with hereditary peripheral neuropathies, respectively. Pathological changes in firing rate were also taken into account, according to experimental findings (Paltaj, 1974).

The value of the skewness was calculated for the Power Spectrum of EMG signals simulating measurements from territories of the muscle at different stages of evolutive neurogenic pathologies (Table I).

**Table I**

<table>
<thead>
<tr>
<th>MUAPT Composition</th>
<th>Skewness of the Power Spectrum</th>
</tr>
</thead>
<tbody>
<tr>
<td>2D N/O 0 GP</td>
<td>0 PGP</td>
</tr>
<tr>
<td>18 N/O 0 GP</td>
<td>1 PGP</td>
</tr>
<tr>
<td>14 N/O 0 GP</td>
<td>3 PGP</td>
</tr>
<tr>
<td>8 N/O 0 GP</td>
<td>6 PGP</td>
</tr>
<tr>
<td>2 N/O 3 GP</td>
<td>7 PGP</td>
</tr>
<tr>
<td>2 N/O 3 GP</td>
<td>5 PGP</td>
</tr>
<tr>
<td>0 N/O 6 GP</td>
<td>2 PGP</td>
</tr>
<tr>
<td>0 N/O 7 GP</td>
<td>0 PGP</td>
</tr>
</tbody>
</table>

EXPERIMENT
In order to validate the reliability of the skewness as an index to follow the evolution of neurogenic diseases, patients were selected who were affected by a well known peripheral nerve pathology: the idiopathic inflammatory polyradiculoneuropathy. In this disease it is often possible to have either a spontaneous remission or a good response to therapy. Four patients were examined two or three times, after a time interval of 0.7 to 1.2 years. The Right Biceps Brachii of the patients have been studied by means of Concentric Needle Electrode using a Medelec MS. The test was carried out under isometric non-fatiguing conditions with the arm flexed at 90° and the forearm in complete supination to avoid synergic contraction of the agonist muscles. The EMG signal was sampled at 5 KHz after low-pass filtering with a phase equalized 4th order Butterworth filter with 2 KHz cut-off frequency to prevent aliasing phenomena. The Fourier Transform has been computed by means of an FFT algorithm with windowing in time. A cosine window has been adopted. The mean value and standard deviation have been calculated over 60 blocks of 1024 points for the skewness of the Power Spectrum.

RESULTS AND DISCUSSION
The results are reported in Table II. For the first three patients the value of the skewness increased from the first measurement to the following. These findings agree with the model expectation since clinically an improvement was assessed showing a decrease of the number of polyphasic MUAPs. The high standard deviation in the case of patient D.M. at the second EMG test might be attributed to unsteadiness of the contraction and requires further investigations.

For Patient P.M. the first EMG test was performed some weeks after the onset of the disease. For this reason only the second and third tests should be considered. Also in these cases the value of the skewness agrees with the clinical findings which show the reaching of a stationary phase. In conclusion although the number of tests shall be obviously enlarged, these preliminary results are encouraging and suggest the convenience of adding the information provided by frequency domain analysis to the other parameters usually adopted.

REFERENCES


INTRODUCTION

Whatever the purpose might be, pertaining the many applications of the EMG signal, either in the field of myoelectric control or in the classification of pathological EMG patterns, the problem of data compression is a fundamental one. In this work the Concentric Needle EMG signal, both simulated and measured in well defined pathological classes is identified by means of linear prediction techniques, yielding a finite number of parameters. Identification aims at investigating whether the parameters relating to a specified model:

1) allow allocation of the muscle examined to a pathological class,
2) preserve all information necessary to obtain from the reconstructed EMG signal the time and frequency domain parameters which are of present clinical interest.

METHODS

By sampling the EMG signal a Time Series is obtained. The EMG signal sample at instant \( t \) is considered as the value taken by a stochastic variable in a casual experiment. The temporal series of the EMG samples is therefore considered as the realization of a stochastic process. By this approach the EMG signal after sampling and digitization is regarded as the output of a linear filter whose input is the white noise (Fig. 1). The problem then is a classical Identification problem: a model must be determined which is able to describe the behavior of the system (in this case the nervous system) starting from experimentally observed input and output data (in this case the input is assumed to be the white noise).

To solve the identification problem it is therefore necessary:

1) to select a class of models
2) to adopt an adequacy criterion for the model.

The criterion selected is based on the one-step prediction error:

The classes of models selected are both the Auto-Regressive (AR) and the AutoRegressive-Moving-Average (ARMA).

The AR(n) model describes the process in the following way:

\[
y(t) = a_1 y(t-1) + a_2 y(t-2) + \ldots + a_n y(t-n) + e(t)
\]

with:

\( a_1, a_2, \ldots, a_n \) AR parameters
\( n \) model order
\( e(t) \) white noise

The ARMA model describes the process in the following way:

\[
y(t) = a_1 y(t-1) + a_2 y(t-2) + \ldots + a_p y(t-p) + e(t)
\]

\[
e(t) = c_1 e(t-1) + c_2 e(t-2) + \ldots + c_q e(t-q)
\]

with:

\( a_1, a_2, \ldots, a_p, c_1, \ldots, c_q \) AR and MA parameters
\( p, q \) model order
\( e(t) \) white noise

RESULTS AND DISCUSSION

1) Classification of cases into pathological classes using AR models.

Both simulated and experimental signals were identified with AR models of order from 1 to 9 yielding the following results:

Simulated signals: the simulated signals were obtained by means of the simulation model described in (Maranzana and Fabbri, 1981). Taking into account the partial autocorrelation function, the variance of the estimation error as well as the stability of the parameters, the model which best identifies the simulated EMG signal is the second order model. In view of a possible pathological classification the AR(2) model was estimated for the experimental signals.

The Right Biceps Brachii of 34 subjects were studied. According to the CNEMG findings each muscle was assigned to one of the following groups: normal, myopathic, recent neurogenic and chronic neurogenic. The AR(2) model parameters did not show any significant among group difference. The derived all-poles description of the system, however, showed possible inadequacy of the adopted sampling frequency.

Identification was therefore attempted at modified sampling frequencies. Interesting discrimination of simulated myopathic from simulated normal EMG patterns was obtained, as well as of simulated neurogenic from simulated normal patterns when the sampling frequency was 2.5 kHz.

1) Reconstruction of the EMG signals.

Diagnostically checking on the estimation error for the AR(2) model proved it not to be white. Further investigations were therefore carried out on the simulated signals using both AR and ARMA models (Marianzana, 1974; Bittanti et al., 1982). In particular AR models of order from 1 to 9 and ARMA models of order from (1,1) to (3,3) were identified (Anderson, 1979). After diagonal checking of the prediction error (Anderson and Portemanteau tests), the optimal model was selected on the basis of the prediction error variance, the Final Prediction Error (FPE), the Akaike information criterion (AIC) and the minimum length criterion (Rissanen). From the parameters of the optimal model and newly generated white noise reconstruction of the signal was obtained (Fig. 2). Time Domain parameters such as number of turns (T), number of zero-crossings, mean amplitude (A), the T/A ratio, and Frequency Domain parameters such as mean and median frequency, skewness and kurtosis of the Power Spectrum were computed both for the original signal and from the reconstructed one. Comparison showed a generally good agreement as for the Frequency Domain parameters. This was confirmed by comparison of the maximum entropy spectrum computed from the identified parameters to the power spectrum computed by means of the FFT algorithm. Results obtained for the Time Domain parameters were less satisfactory.

REFERENCES


ACKNOWLEDGMENT

The authors are deeply indebted to Sergio Bittanti, Centro Teoria Sistemi CNR, Milano, for helpful discussion and challenging suggestions.
A biomechanical approach of sport activities as been made, in recent time, by many authors. In particular the ski jump has been investigated by different techniques. An agreement between the Centro di Biomeccanica and the Italian Federation of Winter Sports (F.I.S.I.) has been performed in order to obtain results able to evaluate the performance of the athlete related with the kinematic and dynamic variables, the equipment adopted and the take-off events.

An useful mathematical model, implemented in a PDP 34/11 computer, describing the ski jumper has been developed. The model consists of two main parts: - the first analyzes the Innuring phase; - the second analyzes the airtime and landing phases. The first part allows to compute the velocity of the athlete during the Innuring and in particular at its end, so that at the first beginning of the flight. The velocity is computed when the geometrical parameter of the outline and the kinematic and dynamic parameters of the athlete are assigned.

The velocity at the beginning of the flight is not the actual of the athlete, as the program neglects the variations induced with the take-off.

The second part allows to predict the trajectory, the instantaneous velocity and the length of the jump. These results are obtained when the first velocity, the dynamical parameter of the athlete and the geometrical parameters of the

The existence of postural activity prior to and during a movement was established in man by BECKNILL et al. (1967). However, neither a complete description of the EMG anticipatory sequence nor a satisfactory interpretation of the biomechanical organization were given. It is the aim of the present research program to answer these questions. The first results have already been published (ZATIARA et BOUSSET, 1980; BOUSSET et ZATIARA, 1981).

Movements of antepulsion-flexion of the upper limbs have been investigated according to three conditions: unilaterial flexions with an adduction of the humerus (UF) and with an additional external rotation (UF), unilaterial flexions (UF) and unilaterial flexions (UF); the dynamic asymmetry of the movements increased from UF to UF and to UF.

**EXPERIMENT.**

Subject stood on a force platform which made it possible to measure the acceleration of the body's center of gravity, according to the antero-posterior, lateral and vertical axes (BOUSSET et al., 1981). Accelerometers fixed on the splints were bound to the wrist of the moving upper limb and to the contralateral side. This made it possible to measure the tangential acceleration of the arm (N) and to determine the on-set and the sign of the anticipatory antero-posterior local accelerations. The activity of the Anterior Pectoralis (AP) and of the main muscles of the lower limbs, pelvis, trunk and scapular girdle and the iliopsoas (IP) and the iliac (IL) lateral sides, were recorded by surface electromyography (EMG). The instructions given to the subject was to point as quickly as possible, with his upper limb stretched out, at a light located in front of him at shoulder level. The lighting up of the target was the signal that the instructions should be carried out. Fifteen healthy adult subjects were tested during experimental sessions of 40-70 movements carried out in series of 3-5 movements of each type. During each trial, EMG activity of the AP and/or IP were recorded. The other muscles and accelerations were recorded by rotation in order to make the relevant comparisions possible. The onset of the activity of AP and the onset of IP were used respectively as time origins for the EMG and biomechanical activities.

**RESULTS.**

Before the activation of the AP, a sequence of inhibitions (+) and excitations (-) concerning muscles of lower limb, pelvis and trunk. These anticipatory EMG activities were organized according to a pattern reproducible for one subject and consistent from one subject to another (see Fig. 1 and 2). Moreover, this pattern was specific to the forthcoming voluntary movement: 1) for the UF, the chronology of anticipatory EMG activity was Selons (+), Pectoralis (+), Erector Spinae (+), Iliopsoas (+), Erector Spinae (+); 2) for the AP, anticipatory EMG only concerned the muscles of Selons, Iliopsoas, Gluteus Maximus and Erector Spinae; 3) for both types of movement the anticipation of the EMG activities of the lower limbs and pelvis increased with the dynamic asymmetry of the forthcoming movement.

The onset of IP was also preceded by anticipatory acceleration at the general level as well as at the local one. The shape of these anticipatory accelerations was reproducible for one subject and consistent from one subject to another. The body's center of gravity was always subjected to a downward and upward anticipatory acceleration. The duration of this general anticipatory acceleration increased with the asymmetry of the forthcoming movement. The local anticipatory accelerations were organized according to a pattern specific to the forthcoming movement: 1) for the UF, the local anticipatory accelerations of the ipsilateral side were different from the contralateral side; 2) for the AP, identity was the rule; 3) the anticipation of these local anticipatory accelerations increased with the asymmetry of the forthcoming movement. So the anticipatory local accelerations pattern showed the same modifications with respect to the type of the forthcoming movement as the anticipatory EMG activities pattern.

**DISCUSSION.**

The present results establish that volu-
S Buisset Anticipatory EMG Activities Related to Voluntary Movement

tory elevation of the upper limb is preceded by movements of the body's center of gravity and that these general movements are related to postural anticipatory movements of the lower limbs, pelvis, trunk and shoulders which are due to anticipatory muscular activities. Moreover they show that these postural adjustments are specific to the forthcoming movement.

The study of both anticipatory EMG activities and local accelerations makes it possible to partly determine the anticipatory postural movements: an extension of the spine occurs with, for the BF, an extension of knees and hips, and, for the QT, a rotation of the trunk about the vertical axis (from the ipsilateral to the contralateral side), due to a flexion of the ipsilateral knee and hip and an extension of the contralateral knee and hip.

The analysis of the system of forces corresponding to the subject at the onset of the upper limb movement makes it possible to consider that anticipatory postural adjustments would create a suspension of the body a movement whose forces of inertia would balance, when time comes, the forces of inertia due to the voluntary movement, which are disturbing for the postural equilibrium.

The reproducibility of the EMG and accelerometric activities for one subject and from one subject to another and their specificity to the forthcoming movement makes it possible to consider that voluntary movement and postural adjustments are part of a same motor program. Moreover, the anticipatory postural adjustments may be considered as programmed, a result which is in accordance with HELMSTET et al. (1997) and with MEISS and HAINES (1979 personal communication). Also,


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![Fig. 1](image1.png)

**EMG activities of the main muscles of the trunk and the scapula-pelvis for unilateral flexions with additional inertias (UF) and for bilateral flexions (BF).**

The figure represents, for one subject, the rectified and smoothed EMG activities of five trials, superimposed by synchronising the records on the onset of the Anterior Deltoides (DA). From bottom to top: Erectores Spinae (ES), Latissimus Dorsi (LD), Obliquus Extremus (OE), Rectus Abdominis (RA), Serratus Anterior (SA), Tripecoidea Superior (TS), Pectoralis Major, sternal portion (PM), i and c: ipsilateral and contralateral muscles with respect to the moving upper limb.

![Fig. 2](image2.png)

**EMG activities of the main muscles of the lower limbs and pelvis for unilateral flexions with additional inertias (UF) and bilateral flexions (BF).**

As for figure 1, the figure represents, for one subject, the rectified and smoothed EMG activities of five trials, superimposed by synchronising the records on the onset of the Anterior Deltoides (DA). From bottom to top: Soleus (SO), Tibialis Anterior (TA), Vastus lateralis (VL), Rectus Femoris (RF), Semitendinosus (ST), Tensor Fascia Latae (TFL), Gluteus Maximus (GM).

INTRODUCTION

The knee joint depends on muscles to provide stability as well as movement. As such, the muscles contribute a major portion of the forces acting at the knee. The muscles control motion through a combination of agonist and antagonist activity and stabilize the joint by producing moments to maintain equilibrium. An understanding of the factors that influence the level of muscle contraction, muscle synergy, and the presence of antagonist activity about the knee is an important component in the analysis of internal mechanics of the joint. The purpose of the present study was to investigate the relationship between external moments acting at the knee, knee flexion angle, and myoelectric activity of twelve muscles surrounding the knee joint.

MATERIALS AND METHODS

The twelve muscles included were the biceps femoris, semi-tendinosus, semimembranosus, gastrocnemius, sartorius, tensor fascia latae, gracilis, vastus lateralis, vastus intermedius, vastus medialis, and rectus femoris. Fine wire electrodes, 0.5 mm in diameter, were inserted into each muscle. The electrodes were connected to preamplifiers near the electrode location. The signals were further amplified in main amplifiers and recorded on magnetic tape together with a knee-angleometer signal. For evaluation, the recorded signals were played back and fed to multichannel r.m.s. detectors, and further to the analog-to-digital converter of a computer. The r.m.s. detectors included low-pass filters with a 3 dB frequency limit of 0.8 Hz. The conversion rate was 6 Hz. The mean r.m.s. value was calculated over a 15 second period for which the load was held constant. Loads were applied at the knee joint in directions tending to produce flexion, extension, and combined loads of flexion-adduction, flexion-abduction, and extension-adduction, and extension-abduction. The tests were performed at 10, 20 and 40° of knee flexion.

RESULTS

The moments tending to flex the knee produced significant activity in 8 of the 12 muscles (VI, VM, VL, RF, Tt, GR, GM, and GL) at all three angles of knee flexion. The maximum extensor muscle activity was found when the knee was at 10° of flexion. The muscle activity decreased by nearly 50% for the same external moment when the knee angle was changed from 10° to 40° (Figure 1). Three of the quadriceps muscles (VI, VM, and Vt) had a linear increase in muscle activity with increase in applied moments, while near full extension the vastus medialis and lateralis muscles had a cubic trend indicating a saturation of muscle activity at the highest load levels. When muscle activity was near maximum, there was not always a linear relationship between changes in myoelectric activity and in joint moment.

There was also significant activity in muscles not normally extended with the knee in the knee moments tending to flex the joint were applied. The tensor fascia latae, sartorius, and gracilis were all active. Their actions may be present to stabilize the knee by producing loads balancing some of the action of the quadriceps mechanism. As expected, the flexor muscle groups primarily responded to the external loads tending to extend the joint. In general, the knee flexors appear more efficient at 10° of flexion than at higher flexion angles. The hamstrings muscles showed a significant increase in

T ANDRIACCHI ET AL. RELATIONSHIP OF APPLIED EXTERNAL MOMENTS AND MYOELECTRIC ACTIVITY MUSCLE ACTIVITY AS THE KNEE FLEXED FROM 10° TO 40° OF FLEXION.

DISCUSSION

The results of this study tend to indicate that the level of muscle activity cannot be predicted from external moments acting about the joint in a simple manner. Some of the explanations for this phenomenon can be described in purely mechanical terms. First, the patellar mechanism changes the moment arm of the quadriceps mechanism with changes in the knee flexion angle. Secondly, the contact point between the femur and the tibia can be moved posteriorly with increasing knee flexion, increasing the mechanical advantage of the extensor mechanism and reducing the mechanical advantage of the flexor mechanism with increasing flexion angle. Thus, there is not necessarily a linear change between the extrinsic moments and muscle activity. This non-linear response seems to be influenced to some extent by a combination of the presence of antagonist muscle activity as well as changes in the moment arms of the individual muscles.
MUSCLE ACTIVITY VS. EXTERNAL FLEXION MOMENT AT KNEE

INTRODUCTION
Biomechanical models of joint systems have proven to be important tools for research and practical ergonomic applications (Andersson et al., 1977). One important part in the construction of a biomechanical model is the measurement of moments created by different muscles around a joint. The muscle force is often estimated from the RMS value of the EMG signal (Kajdes, 1978), although the RMS value is considered to be unreliable for this purpose under fatigue.

In situations where the RMS value is unreliable in the estimation of muscle force it is of interest to use another measurable parameter related to the muscle force. One such parameter is the intramuscular pressure. It is well known that this pressure relates to muscle contraction (Mubarak et al., 1976). Unlike the EMG signal the pressure is a mechanical parameter resulting from mechanical events taking place in the muscle during contraction. It is therefore reasonable to believe that the pressure in some way relates to the mechanical force generated during contraction even if several other factors may contribute to the build-up of the pressure.

MATERIAL AND METHODS
The data were obtained from experiments with twenty-six healthy male subjects aged twenty to forty-five years who volunteered for the investigation. A pressure catheter was introduced into the biceps brachii muscle under local anaesthesia. In seven subjects wire-electrodes were used (Mubarak et al., 1976) but in the remaining nineteen subjects infusion type catheters were used. The pressure catheters were connected to a Bentley Trantec Electronic pressure transducer and the signal amplified and recorded. The EMG signals were picked up from the three elbow flexors (biceps, brachialis and brachioradialis) by monopolar wire electrodes. The EMG signals were amplified with a bandwidth of 30 to 1000 Hz and the RMS value with a 100 ms filter time constant as well as the raw EMG recorded.

The subjects were told to perform isometric elbow flexions with the hand in full supination and the elbow in ninety degrees of flexion. The arms of the subjects were fixed in a special set-up in order to enhance reproducibility in the loadings. The load generated at the wrist was measured by a force transducer and fed back to the subject using a CRG. The force, pressure and the RMS values of the EMG signals were displayed on a chart recorder.

The subjects were told to generate different load levels and to maintain each load for five seconds. In twenty-four experiments the loads were between one and ten kg with one kg increments but in five experiments the loads were between one and forty kg, the maximum load being dependent upon the maximum contraction capacity of the subject.

RESULTS
In all experiments there was a very close correlation between the loading of the supinated wrist and the intramuscular pressure in the biceps. The relation is very nearly linear and a typical result is illustrated in Figure 1. The absolute value of the function was different between individuals but could also vary in the same individual because of changes in the mode of lifting and in the depths of the measuring catheters in the muscles. In eight experiments the pressure gradient in the muscles was studied. The decrease in pressure between centre and periphery of the muscle was 80 to 90 per cent. There was no apparent effect of muscle fatigue on the pressure. The linear correlation coefficient between wrist load and biceps pressure in 92 series of loadings averaged 0.963 (SD = 0.041).

The well-known relation between wrist load in supination and the RMS value of the biceps EMG signal was demonstrated also in these experiments. Figure 2 shows this relation from the same experiment as in Figure 1. Also, the linear correlation coefficient between load and EMG was high averaging 0.968 (SD = 0.027) in 79 series of loadings. As both the pressure and the RMS value were linearly related to load consequently also the relation between pressure and EMG value was very linear with an average correlation coefficient of 0.964 (SD = 0.038) in 79 series of loadings. However, in cases where the pressure and the EMG were poorly related to the load because of different modes of lifting at different load levels the relation between EMG and
P Parker et al. Effects of Load on Skeletal Muscle Interstitial Pressure and Myoelectric pressure was still very linear. Figure 3 shows the relationship between wrist load, biceps pressure, and biceps EMG during a sequence of dynamic isometric loading. The correlation between the three parameters is very good, with an average correlation coefficient during five minutes of loading amounting to 0.971 (P < 0.02) and 0.916 (P < 0.04) for load/EMG. Thus, it seems as if both the RMS value of the EMG signal and the pressure are directly related to the mechanical function of muscle.

DISCUSSION

The results of the present study show that both the intramuscular pressure and the RMS value of the biceps EMG signal are linearly correlated to the external load on the supinated wrist. In fact, in cases where the subject was unable to perform isometric contractions with great reproducibility, the EMG signal and the pressure correlated better than either of them did to the load. Therefore, there is no doubt that both parameters are reflecting the muscle mechanical function. The RMS value of the EMG signal is known to be related to the force of a muscle contraction.

However, the relation has defined limits under muscular fatigue and other conditions. The pressure, being a mechanical parameter may be more closely related to the force than the EMG signal. The pressure is also affected by several factors which limit its use for the estimation of muscle force. Therefore the pressure can only be used as an estimator of muscle force in situations which permit reproducible isometric conditions.

To conclude, the intramuscular pressure can be used to approximate the muscular force in biomechanic studies with the same precision as the RMS value of the EMG signal under constant, reproducible, isometric conditions. When fatigue affects the reliability of the RMS value the intramuscular pressure is probably more accurate for the estimation of force than the EMG signal. There is however no evidence that there exists a constant relationship between muscular force and intramuscular pressure.

REFERENCES


RELATIONSHIP BETWEEN KINEMATIC MUSCLE FORCES AND EMG (ABGT)

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The relationship between the isometric muscle forces and the integrated EMG is reported by many researchers to be a linear one. Some, however, indicate that this might not be true for very high contractions. Few findings have been reported for the relationship between the EMG and the dynamically changing muscle forces. This paper presents early results of a study about the relationship between the EMG and dynamic muscle forces for the elbow flexion. The relationship is based on a model of the EMG in this study. Additional parameters from the measured EMG, the kinematic information of the prevailing muscle lengths, and the subject's specific physiological data are incorporated in the model. Elbow flexion in a neutral position was chosen for this study so as to see only a few muscles involved. The evaluation for the additional parameters of the model was based on data from experiments with ten subjects. Each subject was asked to flex the arm quickly with different external loads attached to the wrist. The recordings of the EMG, the external load and the kinematics were synchronized. The validity of the model for the measured conditions and the application of the model to other muscle groups are reported in this paper.

RELATIVE ACTIVITY OF THE DELTOID, THE PECTORALIS MAJOR AND THE INFRASPINATUS MUSCLES IN DIFFERENT POSITIONS OF ABDUCTION.

J.A. Kielberg, Department of Anatomy and Embryology, University of Groningen, The Netherlands.

INTRODUCTION

In a preliminary study to investigations concerning the relative contribution of some shoulder muscles to the abduction we first tried to answer the following questions:

1. What is the variation of the EMG measurements? (Penzchler and Wunder, 1981)
2. In what plane should the abduction be performed? innan et al. (1944) did their investigations on abduction in the frontal plane. Johnson (1937) found the scapular plane to be the more physiological. Freedman and Munro (1966) investigated abduction in the scapular plane in 61 persons. They stated that the scapula, humerus and attachments of humeroulnar muscles in a single plane. Poppen and Walker (1978) reported about EMG investigations concerning abduction in the scapular plane.

In our study surface EMG was recorded in different isometric positions of abduction, in the frontal as well as in the scapular plane from the deltoid, the pectoralis major and the infraspinatus muscles. In order to get a clearer idea about variation and the influence of the plane in which the abduction is performed.

MATERIALS AND METHODS

Surface EMG recordings were obtained from four healthy men who were subjected to the protocol on three different days (see Table 1) for muscles positions and technical data). Electrode placement was standardized. For the plane of the scapula a plane was taken with a 30° angle to the frontal plane. In the externally loaded situation 1 kg was added at the wrist. EMG was recorded during five seconds for each position. The EMG signals were filtered, amplified and stored on a Bell and Howell tape recorder. Later on raw EMG signals were analyzed by using a PDP-11 computer. The rectified EMG was integrated over a one second period. Thereafter the absolute EMG values were related to the maximal EMG (100%) found for each muscle during MVC in one session.

TABLE 1 Muscles and standardized electrode placement

<table>
<thead>
<tr>
<th>Muscles</th>
<th>Electrode placement at 90° abduction</th>
</tr>
</thead>
<tbody>
<tr>
<td>Deltoid anterior part</td>
<td>at 90° abduction: along line a.c.junction-medial epicondyle</td>
</tr>
<tr>
<td>middle part</td>
<td>0°:along line mid-scarpeion-lateral epicondyle</td>
</tr>
<tr>
<td>posterior part</td>
<td>at 90° abduction: along line posterior: ant-scarpeion-lateral epicondyle</td>
</tr>
<tr>
<td>Infraspinatus</td>
<td>below a.c.joint and medial from axilla</td>
</tr>
<tr>
<td>Pectoralis major clavicular part</td>
<td>below midlualcular</td>
</tr>
</tbody>
</table>

TABLE 2 Positions during one session, including external loading at the wrist.

<table>
<thead>
<tr>
<th>Abduction Angle</th>
<th>Frontal plane</th>
<th>Scapular plane</th>
<th>Frontal plane + 1 kg</th>
</tr>
</thead>
<tbody>
<tr>
<td>30°</td>
<td>xx</td>
<td>xx</td>
<td>xx</td>
</tr>
<tr>
<td>45°</td>
<td>xx</td>
<td>xx</td>
<td>xx</td>
</tr>
<tr>
<td>60°</td>
<td>xx</td>
<td>xx</td>
<td>xx</td>
</tr>
<tr>
<td>75°</td>
<td>xx</td>
<td>xx</td>
<td>xx</td>
</tr>
<tr>
<td>90°</td>
<td>xx</td>
<td>xx</td>
<td>xx</td>
</tr>
</tbody>
</table>

3x MVC Frontal plane - - - xx
3x MVC Scapular plane - - - xx

TABLE 3 Technical data

| Beckman surface electrodes, 11 mm diameter, recording surface 2 mm, interelectrode distance 11 mm, amplifier bandwidth 30 Hz-10 kHz, input impedance 100 M, CMRR 50 dB-90 dB, tape recorder 5 kHz-10 kHz, sampling rate A/D conversion 5 kHz |

REFERENCES

In figure 1-4 the mean EMG values (x±S.D.) are presented as a percentage of the MVC found during MVC.

For all muscles (except pectoralis) a linear relation was found between EMG and angle of abduction (r=0.94-1.00). The pectoralis major showed no activity in frontal plane abduction. There was little activity in the scapular plane by amplifying the signal 5000 mV. The signal/noise ratio was insufficient to quantify this EMG. This...
J Ringelberg et al. Relative Activity of the Deltoid....

MVC was 9.99 kg (mean) ± 2.13 kg (S.D.), registration at the wrist.

Standard error of Mean was between 5 and 15%.

DISCUSSION

1. S.E.M. = 5-15% - is acceptable for measurements under these conditions.
2. From figures 1-4 it may be concluded that the EMG activity of the three parts of the deltoid muscle is lowest in scapular plane abduction, except for the anterior part between 30° and 60°. The infraspinatus shows more activity in the scapular plane. In our opinion the optimal situation will be found for abduction in the scapular plane, because the most important agonist, the deltoid muscle is least active. The supraspinatus muscle was not investigated. Judging from its anatomic position, it will have a more comfortable situation in scapular plane abduction because of its origin in the supraspinatus fossa.

REFERENCES


ANALYSIS OF FORCE TRACKING AND EMG INDICATING SENSORIMOTOR PERFORMANCE

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INTRODUCTION

Tracking performance has been investigated extensively either to detect and describe human operator performance in terms of control theory or to analyse properties and disturbances of the neuromuscular system for medical diagnosis. Many experiments have been described in which movements of limbs were involved. The interpretation of these results is very difficult since (1) different types of sensor information (position, velocity, acceleration, force, pressure, kinesthesia and visual input) is processed by the human operator, and (2) the force transmission via the muscle groups involved causes a resulting movement which is difficult to be interpreted. A strong restriction is to prevent movements and design a force tracking task in isometric conditions. In this situation EMG activities of the muscle involved may give a deeper insight in neuromuscular control than the mechanical output since this source of information is one step closer to the CNS and the muscles. The present study was performed to find a well-defined biomechanical situation for isometric force tracking where the EMG of the most important muscle groups can be recorded by means of surface electrodes. Different force tracking tasks are designed including a critical tracking task (1/).

EXPERIMENT

The tracking force as the output signal of the human operator was the flexion/extension torque of the hand which was exerted in a set-up indicated by Fig. 1. The ankle transducer a voltage proportional to the torque was produced. In the target tracking task the feedback element had a proportional characteristic. The task of the human operator was to make the horizontal cursor line on an electronic display follow a target line the position of which changed according to a selected function of time. The control loop is closed by the visual feedback path.

Fig. 1: Target Tracking Task: Experimental Set-Up

In the critical tracking task the dynamic characteristic of the feedback element was different: there was an integrating effect combined with a constantly increasing gain. The constant set point given by the force has been recorded. In addition, the EMG activities...
Analysis of Force Tracking and EMG Indicating Sensorimotor Performance

A critical tracking task requires a high degree of sensorimotor performance. The human operator must continuously adjust the position of a cursor on a computer screen to track a target. This task is performed by varying the frequency of EMG signals, which are recorded and amplified.

Fig. 2a: Example for Target Tracking: triangular wave form of target position

Fig. 2b: Critical Tracking: Force and Averaged EMG

DATA ANALYSIS

a) Target tracking task: In the example of Fig. 2a, a triangular function of low frequency (0.1 Hz) has been followed. Since the averaged EMG is good, the approximation proportional to force can see an alternating contraction of agonist and antagonist. In addition, there are components of higher frequency which are also to be seen in the force.

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Electromyograms show periods of activity. Combined with a photographic registration of movement of body segments and joints, these signals can be divided into periods of concentric and eccentric contractions. Prerequisite for this is that a link is established which synchronizes the time axis on the two sets of registration.

When rising and seating oneself, the movements of the spine are initiated and regulated by short and long trunk muscles. For the hip joint it is mainly its extensors which are important. The purpose of this paper is to report the periods of concentric and eccentric contractions in the pelvis, hips, and knees.

Methods

For electromyography the rectus femoris, the gluteus maximus, the rectus abdominis, and the erector spinae muscles were selected. Their activity was recorded by surface electrodes, and the signals were recorded rectified and averaged.

The movements, seating oneself and rising from a chair, were registered photographically with strain gauge and lighttracks. The lighttracks were formed by diodes placed on the shoulder, on the anterior and posterior midline of the body. The frontal lighttracks were viewed by the eye

with the electromyograms.

RESULTS

The photographic registration of the movements seating oneself, is shown in Fig. 1a with the corresponding EMG (ib) and analysis (ic) added. In the photograph each light-dark period lasts 20 msecs. The missing signal in this case found after six periods, i.e., at 140 msecs from the beginning of the movement (small arrows). Seat contact is established at 760 msecs (large arrow).

With the diodes placed on the knee, hip, and shoulder the hip flexion and the forward inclination of the trunk could be measured at each 20 msecs (ic). The two diodes on the tibia showed the inclination of the pelvis while alterations in the lumbar lordosis appeared as the difference between this and the trunk inclination. In the beginning of the movement, the center of gravity must be kept within the area of support. The movement is therefore initiated by pulling the pelvis backward and the shoulder and knee slightly forward. Hip flexion and trunk inclination reach their maximum at the time of seat contact, while a 15-20° flattening of the lordosis is kept as part of the sitting position (ic).

A part of the quadriceps the rectus femoris (r.f.) shows two maxima: at the deceleration of the downward movement and when the body moves backward to approach the seat.

The gluteus maximus (g.m.) contracts eccentrically and isometrically during the downward movement. Its peak activity is registered immediately after touchdown where the muscle with a concentric contraction erects the trunk.

The rectus abdominis (r.a.) initiates the forward inclination of the trunk, supporting the flattening of the lordosis and the abdominal wall and ends with an eccentric contraction.
A Christiansen et al. Concentric and Eccentric Contractions of Hamstrings

The erector spinae (e.s.) adjusts by an eccentric and isometric contraction the flattening of the lordosis.
In the action of rising oneself (fig. 2) the center of gravity is brought forward over the feet. The hip continues to flex after the seat has been lifted.
A peak activity in the gluteus maximus starts its concentric contraction and the rising of the trunk.

INTRODUCTION

During his first school — year the child carries a satchel with school accessories weighing up to 5.5 kg. There exist contrary opinions concerning permissiveness of loading the child in this way and the influence there of upon the development of a bad posture and deformations in the locomotor system. Therefore, we undertook to analyze the influence of loading the child upon the activity of single muscles of the trunk and those of the lower extremities, and also the influence of shifting the center of gravity occurring at carrying the satchel on the back, in the right or in the left hand, an experiment was performed also in the absence of loading. These influences may play an important part in the mechanism of maintaining an upright posture or maintaining single segments of the locomotor apparatus in physiological relationships. On the basis of the data obtained we shall try to clear up this dilemma experienced by parents and all those concerned with a normal development of the child.

EXPERIMENTAL CONDITIONS

On the occasion of standard medical examinations at the children's clinical hospital there were chosen seven children (4 boys, 3 girls) seven to nine years old, with whom no clinical symptoms of bad posture, spine deformities or changes in the structures of the locomotor system had been noticed.

In the course of the experiment we recorded muscle activity (electromyographic activity — EMG) and the shift of the center of gravity while resting without loading, with the satchel on the back after a twenty minutes' walk with the satchel, and the satchel in the right and the left hand respectively. The EMG activity was detected via surface disc electrodes in the paravertebral muscles of the thoracic and lumbar region, oblique abdominal musculer, flexors and extensors of the knee. Recording by a Van Gogh polygraph (amplification 175 mV/1mV, filter 35, time constant 0.01). The shift of the center of gravity was analyzed by a statokinesiometric method (Grocanin et al., 1977). The obtained EMG records are assessed as: 0 — no activity; 0.5 (1) — poor EMG activity; 1 (+) — moderate EMG activity, and 2 (+) very strong EMG activity. With respect to the shifts of the center of gravity we assessed the changes in the medium radius of movement and direction in which the center of gravity is shifted (there exist data on the shifts of the center of gravity on X and Y axes, and the maximal movement radius).

RESULTS

During the experiment with and without loading a minimal EMG activity was recorded with five children, whereas it was slightly more expressed with two children. Table I shows the common EMG activity of all seven children with respect to single muscles in different conditions. As evident from the Table, the greatest muscle activity occurs at the beginning of the experiment irrespective of the fact whether a child is loaded with a satchel or not or whether he is holding it in his left or his right hand. The greatest activity is obviously displayed in m. obliquus abdominis sin., and m. paravertebrales rect. thoracallis sin., however, it is somewhat lesser here. An interesting relationship is to be noticed.
between the magnitude of the activity of \( m. \) quadriceps femoris sin. and \( m. \) biceps femoris dext., which display re-
ciprocity. The distribution and inten-
sity of the EMG activity according to
Figure 1 show that:
- in general the EMG activity is grea-
ter on the left than on the right;
- it is relatively maximal at the be-
ginning of the experiment;
- while holding the satchel in the
right hand there occurs, beside a mo-
re strongly expressed EMG activity of
the left side muscles, an increased
activity in \( m. \) paravertebrales reg.
dorsalis. The situation is
reversed when the child holds the
satchel in his left hand;
- at the end of the experiment, in the
standing position without load, the
EMG activity is minimal.

By means of staticsometry there was
concluded that with two children no
essential changes could be noted in
the medium radius of the movement of
the center of gravity as well as no la-
teral shifts. With others we noticed
differences in the medium radius of
the movement under various test con-
titions, and with four children also a
lateral shift of the center of gravity
to the opposite side while holding the
satchel in the right hand and the left
hand, respectively.

**DISCUSSION AND CONCLUSIONS**

Due to a relatively small number of
children participating in the ex-
periments the data obtained can only be
treated as preliminary. However, we
can state that the distribution pa-
tterns of the EMG activity are constant;
the difference is merely in the degree
of activity with each child. On the ba-
sis of these analyses we can draw a
conclusion that it is best for the
child to carry the satchel on his back,
whereas the carrying of the satchel in
one of the hands causes changes to a
degree as observed with the child in
new situations and connected with un-
certainty as noticed at the beginning
of the experiment.

An interesting correlation is to be no-
ted between the distribution, the de-
gree of the EMG activity, the medium
radius of the movement and the situ-
ation of shifting the center of gravity
showing to what extent the child is able
to compensate the influence of an
outer force (the weight of the satchel)
in maintaining an upward posture. Thus,
we can see that a group of children
can maintain the initial position with-
out considerable shifts of the center
of gravity and the EMG activity without
taking into account the manner of
loading (on the back, in the left or the
right hand). The second group displays
a greater EMG activity and changes in
the medium radius of the center of
gravity movement, whereas there are no
substantial lateral or backward shifts;
and finally, a group with which there
is no difference can be observed, beside a
lower EMG activity, also a contralateral
shift of the center of gravity while hold-
ing the satchel in the left hand.

We could say that with all of
these children, in spite of different
reactions to a newly arisen situation,
a compensation to the influence of an
outer force is clearly evident. The best
of these is undoubtedly the group show-
ing no shift for a greater activation
of muscles and maintaining the center
of gravity in its lateral position.

With the children with whom the EMG
activity is more stressed and the cen-
ter of gravity is subject to compen-
satory shifting, there can be foreseen a
more rapid occurrence of fatigue and
weakening, which due to repeated loa-
ding can result in bad posture and sub-
sequent deformations. However, the shift
of the center of gravity without the
occurrence of the EMG activity which
was not detected in our group of chil-
dren represents a major threat to the
population of children since with them
the mechanisms responsible for main-
taining an upward posture behave contrary
to one's expectations. The reasons the-
reto are different, to be sure, but it
that the question is of an insufficient
control of postural mechanisms, insuf-
ficient functions of motor units irre-
spective of the etiology (cerebral, neu-
ral etc.). Anyway, it would be advanta-
geous if the children suffering from
neuromuscular diseases were subjected to
polyelectro-
myographic analysis under loading to re-
veal the possibility and degree of com-
pen
sation. In this way a quantitative
influence of loading upon posture and
development of deformations could be
avoided with these children.

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myographic analysis under loading to re-
veal the possibility and degree of com-
pen
sation. In this way a quantitative
influence of loading upon posture and
development of deformations could be
avoided with these children.
Hengi Lumped - Parameter model of the human thermoregulation system

**ABSTRACT**

Body temperature in man is kept in its normal range as a result of the combined responses of various physiological systems involved in thermoregulation. The regulation system under study is basically a non-linear negative feedback control system with constant reference input. The aim of this study is to construct a mathematical model of the temperature regulation in human body which can be used to study the certain properties of the real system. For this purpose lumped-parameter models, and their electrical analogues, have been constructed and simulated on a digital computer. Simulation of the models have been based on the numerical data gathered from the published works of various workers. Validity of the models have been verified for various environmental conditions.

**MATHEMATICAL MODEL**

While constructing the model the human body has been partitioned into four simple geometrical segments as follows: Head as a sphere; trunk, upper extremities, and lower extremities each as a cylinder. Furthermore, each body segment has been assumed as consisting of three concentric (axial) compartments with uniform thermal properties Core, muscle, and skin and fat (Fig. 2). Blood has been considered as a body compartment interconnecting each body segments and compartments. Since the thermal properties of the tissues/which compartment do not differ very much, each compartment is assumed uniform throughout. Thus it is possible to represent a compartment with a lumped model simulating heat storing (C) and heat conducting (k) properties.

**ELECTRICAL ANALOG MODEL**

Using the analogy between temperature and electrical potential as "across variables" and heat flow and electrical current as "through variables" it is possible to construct an electrical analog circuit for the temperature regulation system in human body (Fig. 3).

In the analog circuit: (a) Heat generation by metabolism and/or muscular work, and heat loss by evaporation are represented by controlled current sources; (b) Total metabolic heat generation is divided into all compartments except blood compartment according to the weight percentages; (c) the temperature gradient dependent heat exchange mechanisms between skin and environment (i.e., radiation, convection, conduction) is modeled by equivalent variable thermal conductances; (d) ambient air temperature is represented by a voltage source connected to skin conductances of each segment; (e) all temperatures are measured with respect to a reference point which is taken as an 0°C Centigrade in the model; G's represent thermal capacitances; G's thermal conductances; k's metabolic heat generation; k's muscular heat generation; k's evaporative heat loss by sweating.

**RESULTS**

The computer program has been organized in a manner to be suitable for various environmental and experimental conditions to the system. Two sets of computed results are shown in Fig. 4 and Fig. 5, which appear to lend support to the validity of the derived lumped-parameter model of the human thermoregulation system (shown in Fig. 3)

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POSTURAL CONTROL IN INITIATION OF GAIT

M. Gregorič et al. Postural Control in Initiation of Gait

INTRODUCTION

Apart from the work of Carlsson (1966), R. Herman et al. (1973) and Cook and Cozzens (1975), the postural mechanisms in initiation of gait have not been extensively studied in man. In this study we wished to determine the changes in the floor reaction force and in related electromyographic activity of pretibial and calf muscles when gait is initiated from a symmetrical standing position in healthy subjects and in patients with Parkinsonism.

METHODS

Six adult healthy subjects and 4 patients with Parkinsonism (bilateral involvement, hypokinesia, rigidity, and disturbances of posture and gait) were included in the study. The subjects were standing on the Kistler force-platform in a symmetrical "easy" standing position. They were instructed to start walking after the starting signal (flash) with ordinary, fast and slow velocities of gait. The displacements of the centre of force (CF) were computed up to the moment the subjects stepped out of the platform. The electromyographic activity (EMG) of the tibial ant. and triceps surae muscles was recorded simultaneously on both sides with surface electrodes. Phases of gait were recorded by footswitches. The technique is described by Lj.R. Larsson et al. (1980).

RESULTS

In healthy subjects the CF moved with a delay after the starting signal at first backwards and slightly laterally to the side of the first "swinging leg", then to the other side, toward the "stance leg" and finally forwards (Fig. 1). The EMG activity of pretibial muscles on both sides was shortly preceding the changes in forces. This activity was sustained during a backward shift

and it stopped when the CF was displaced forwards. At the first push off phase (heel up-ball up) pretibial muscles were still active, while there was little or no activity of calf muscles and the CF was in the backward position; however, it was shifted to the side of the "stance leg" (in order to unload the "swinging leg"). The CF moved forwards not later than at the end of the first swing phase. The first push off (heel up-ball up) phase recorded by footswitches was of a comparatively short duration. In patients with Parkinsonism the movements of the CF often showed some oscillations of the lower amplitude before the final forward displacement (Fig. 1). The start of the first step following the signal was delayed in comparison to healthy subjects. Coactivation of pretibial and calf muscles was often observed in the patients (Fig. 2). The initial backward and lateral displacement of the CF was larger and the EMG activity of pretibial muscles was higher at faster than at slower velocities of gait.

CONCLUSIONS

The described changes in the position of the centre of force and in the EMG activity of leg muscles on starting the gait from a symmetrical standing position seem to reflect the complex postural mechanisms by means of which the body is increasing the impulse of force and the acceleration of the initial movements. These mechanisms are disturbed in Parkinsonism which points to an important role of the basal ganglia in the neural control of initiation of the stepping cycle.

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LITERATURE


BODY SWAY IN DIFFERENT STANDING POSTURES

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INTRODUCTION

The platform stabilometry (statokinestimetry) is the recording of body sway by measuring the displacement of the centre of force on a force measuring platform during quiet standing. Since visual inspection of the output curves (statokinestigrams and stabiliograms) does not give us enough quantitative data on body sway, numerous computer-based processing techniques have been introduced to process these signals (Njolstam 1973, Hafschmidt 1979). The results of this computation involve a set of parameters, which describe the characteristics of the body sway. For the study of postural subsystems (for diagnosis or follow up) these data should be somewhat compared with the "normal" values.

A non suitable model for normal posture is available, we have to evaluate each parameter by its normal values. The Hamming distance to the "normal" vector (total score) gives us an impression of equilibrium function.

The purpose of this study was to point out that the "normal" vectors depend on the orthostatic elements of posture (not only static, but also dynamic ones). Since the subjects adopt different postures during tests, even when quiet easy standing is strictly requested, these static factors should be estimated. To preserve the simplicity of the method, we tried to estimate the static elements of posture from the measured forces using the inverse pendulum model, which has been proposed by many authors (Kapteyn 1973, Gourion et al. 1975).

METHOD

The subjects (6 male and 1 female, aged 30 to 35 years) were asked to stand quiet on the Kistler force-plate (9291A and 9281A11) barefooted with the heels put together to form an angle of 25 degrees in different positions:

- forward leaning (maximum and moderate), neutral and backward leaning; left and right leaning and symmetrical position.

To verify the inverse pendulum dynamics we fixed the subject to the bar by a bandage with the ankle joint free. At the end of each trial two photos were taken by two cameras placed on the left and at the back of the subject. Electromyograms were recorded simultaneously on six muscles groups of the leg and the back (tibialis anterior, triceps surae and erector spineo on both sides) using standard bipolar surface electrodes. From the forces the body sway parameters were computed. The mean speed (m/s), the RMS value (in sagittal and lateral directions) with regard to the mean position and the mean position (NP) of the centre of force and the RMS value of the dynamic part of vertical forces were studied in detail.

RESULTS

All subjects showed increased amplitude of sway (mean speed and RMS values) when not in neutral position (Table 1).

<table>
<thead>
<tr>
<th>LEFT LEANING</th>
<th>SYMMETRICAL</th>
<th>RIGHT LEANING</th>
</tr>
</thead>
<tbody>
<tr>
<td>MAXIMAL</td>
<td>FORWARD</td>
<td>13.9 (±4.6)</td>
</tr>
<tr>
<td>MAXIMAL</td>
<td>FORWARD</td>
<td>13.2 (±4.7)</td>
</tr>
<tr>
<td>MAXIMAL</td>
<td>BACKWARD</td>
<td>21.5 (±3.5)</td>
</tr>
</tbody>
</table>

The sway was more pronounced in the direction of leaning but not in the extreme leaning positions (Fig. 1). On the other hand, the RMS value of vertical forces was not so closely related to the mean position (Fig. 2). Two subjects even had decreased RMS values of vertical forces when not in neutral positions. In passive posture (no EMG activity in tibialis anterior and triceps surae) as well we found increased mean speed and RMS values as compared to the neutral positions. No significant differences were found for RMS values of the vertical forces. With the subject fixed to the bar, similar amplitude position relations were found, the RMS values of vertical forces being also in close relation to the position.

DISCUSSION AND CONCLUSIONS

The minimum amplitude of sway of the centre of force is reached when the body is in alert neutral position. This position can be adopted by following the instructions during the test. It can be checked by EMG (minimal but constant activity in triceps surae), by mean position of the centre of force (about 50 mm in front of the ankle joint axis), and less accurately, by the plumb-line. Since all dynamic parameters are very sensitive (in the sagittal direction at least) to these static elements of posture, this correlation should be taken into account when comparing the data with the normal values.

According to the principles of mechanics, for the inverse pendulum (weightless rod with a point mass (body mass) and with the rotational axis (ankle) at the other end), oscillating with the constant acceleration amplitudes, the amplitudes of oscillations of X forces are proportional to the displacement from the vertical (if the amplitude of oscillations and displacements are small comparatively to the rod length). This relation was observed with the body fixed to the bar. In subjects standing free there was no regularity in relation between the vertical forces (normalized to the same mean speed of the centre of force) and leaning in sagittal direction. Hence the inverse pendulum model can not be accepted for body sway in sagittal direction (at least for the high-frequency range).

Fig. 1 Typical SKG for forward leaning position.

Fig. 2 Typical position-body sway diagram. The mean speed of centre of force (crosses) is more pronounced when the body is not in neutral position. No correlation can be found for RMS values of vertical forces normalized to the body weight and mean speed (circles). For inverse pendulum model this relation is linear (lines).
THE FEEDBACK IN THE SYSTEM OF CORRECTIONS MEANS OF THE SPORTSMEN MOVEMENT DISTURBANCES

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ALL-UNION SCIENTIFIC-RESEARCH INSTITUTE OF PHYSICAL CULTURE

Two levels of regulation of the motor function are known: to be open-loop, based on the theory of the motor programing and close-loop based on the functioning feedback differentiation. In accordance with the close-loop regulation there is the internal and external ring of movement regulation. Taking into consideration the theoretical data, the correction of the motor disturbances caused by deautomatization is proposed. For this purpose the parameters of the external features or internal structure is presented on the graphical display, which gives us the immediate information during the execution of motor action.

In this article the defending action of fencers with technical errors caused by deautomatization of skills were taken as a motor model. Mechanograms and the EMG of deficit muscles are presented on the graphical display. The subject task was to render the movement pattern with the different speed. After the training according to some concrets programus the stabiliti of skills which is characterised by recovery of movement disturbances is observed.

That procedure may be successfully used in the cases of the motor disturbances as a result of a stressful situation of the competitive activiti.

Experimental data and its explanation in the report are presented in detail.

SEP CHANGES BY RETRACTION OF THE THORACIC CORD WITH OR WITHOUT RHIZOTOMY

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INTRODUCTION

As the surgical approaches for the spinal thoracic tumor, anterior decompression or lateral rhachiotomy or centro-transversectomy are generally indicated. But in cases that tumor invaded not only the vertebra but also the epidural space, there are no technique favorable to acquire one stage decompression.

On the other hand, laminectomy with one or two rhizotomies makes it easy to approach the tumor invaded vertebral body and its epidural space. The purpose of this research is to study the effect on spinal cord by lateral retraction with or without rhizotomy utilizing computer-averaged spinal evoked potentials (SEP).

EXPERIMENTS

Two groups: group 1 with rhizotomy (L1, L2), group 2 without rhizotomy. A retractor connected to a micromanipulator was set at the center between L1 and L3 roots and just contacted with dura mata.

Before and during lateral retraction, SEP was measured every 2 minutes. Retractor was driven 1mm every 1 minute.

RESULTS

SEP have been clinically applied to evaluate the neurophysiologic function of spinal cord. Spike potentials in short latency of SEP are interpreted as action potentials pass through the superficial posterolateral tract (1st Negative deflection; 1st N.D.) and posterior tract (2nd Negative deflection; 2nd N.D.) of spinal cord. The significant SEP difference between group 1 and 2 was in the decreasing ratio in the amplitude of both 1st N.D. and 2nd N.D. In group 1 the amplitude decreased to half by 3 or 4 mm retraction, but in group 2 by 6 mm retraction. (Fig.1)

The appearance compared between two groups at the same retracted length was remarkably different. In group 2, the nerve roots of the retracted side were more tense and spinal cord was highly strained than in group 1.

CONCLUSIONS

In our series we observed SEP difference between group 1 and 2. It suggested that the impairment of the spinal cord function due to lateral retraction could be reduced by rhizotomy.
SEP Changes due to Lateral Compression

with Rhizotomy
without Rhizotomy

control
1 mm
2 mm
3 mm
4 mm
5 mm
6 mm

100μv
1ms

100μv
1ms

Fig 1

Aetiology of Idiopathic scoliosis still today retains its doubts and its uncertainties.

In principle the integrity of the muscular system of the scoliotic column may be an useful information for evaluating the malformation. The more and more thorough study of this system is therefore fully justified, with all technologically most advances methods, with quantitative electromyographic study of the paravertebral muscles.

Our research, in fact, provides for picking-up and elaborating in real time the EMG signal from scoliotic subjects.

The study consists of the quantification of the asymmetry of the electromyographic signal which exists between the right and left sides, through the definition and the comparative evaluation of appropriate indexes, the asymmetry index and the percentual difference of power and their successive variations during the possible therapies.

A program, therefore, has been implemented on the 5451B HP Fourier System for computing all above indexes record.

In the paper are presented the procedures of computations and clinical results.

The results of the spectral analysis were similar for the scoliosis and the control groups. The center frequency curves, when plotted versus time, showed a consistent decrease during the loading period, in both subjects and at all electrode locations. A shift towards lower frequency curves of the center frequency curves occurred (figure 4).

There were no significant differences in the fatigue index values between either the sides of the back or between the two investigated groups.

DISCUSSION

A comparatively higher myoelectric signal amplitude was found in the paraspinal muscles on the convex side of the scoliosis curve. This is consistent with previous studies of scoliosis patients, (Brusseatis 1962, Buk 1962), the force required to balance the spine is greater on the convex side for mechanical reasons. We found that the myoelectric amplitude difference increased with increasing degree of scoliosis. Therefore it is probable that the amplitude difference between the sides is caused by the curve.

The center frequency curves showed a decrease during the loading period in both subject groups and at all electrode locations. The fatigue index values also were the same throughout. This indicates a similar rate of ongoing fatigue. Thus, there was no indication of a different response to load of the paraspinal muscles when the patients with scoliosis were compared to healthy controls. The results obtained indicate that the load on the paraspinal muscles on the convex and concave sides were in proportion to their capacity. An adaptation of the convex sides muscles appears to have occurred to the higher load demand.

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Figure 3. The myoelectric amplitude quotient in lumbar scoliosis in relation to curve degree. Mean values ± SEM. N = 41. All curves were convex to the left.

Figure 4. The center frequency curves in Hz during the 2 minute study period. Mean values (SEM). 40 scoliotic subjects and 19 controls.

EQUILIBRIUM REACTIONS IN IDIOPATHIC SCOLIOSIS (ABT)

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CHILDREN'S HOSPITAL AT STANFORD

One hundred and twenty cases of idiopathic scoliosis and fifty age-matched controls were tested by a computerized statometer system to measure postural sway in the standing position. A test of eye-hand coordination by computerized video game was also performed by subjects and controls. Both statically and during rhythmic tilting data describing the amount of sway during eyes open and eyes closed intervals were obtained and compared between subjects with progressive and nonprogressive histories. There were significant differences seen between subjects that were progressive and those who had shown a history of being nonprogressive. Subjects who were progressive despite bracing treatment had the lowest amount of sway of any of the groups and this was significant at .01 level. There were some significant differences between scores on the eye-hand coordination tests between progressive and nonprogressive curves. The progressive curves had a tendency to perform worse on attempts to match two dots on the screen than nonprogressives. Although this type of study does not address the cause of scoliosis it may be useful in predicting which curves are likely to be progressive.
EMG ANALYSIS BY MEANS OF ASYMMETRY INDEX OF SCOLIOTIC PATIENTS

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(*) Centro Don Gnocchi - Parma.

Introduction

Every stimulation procedure adopted so far involves stimulating the convex side of the scoliotic curve (1). By our procedure, 80% of scoliotic patient electrical stimulation, SFES, points to be stimulated were selected on the basis of each patient's electromyographic data. As known EMG activity is studied on both sides of the vertebral column, and it is symmetric in healthy subjects, whereas it is asymmetric in scoliotic patients (2). Stimulation was applied by us to the side showing low-level EMG activity.

Asymmetry was evaluated in scoliotic patients' EMG activity by a specially designed Distributed Microprocessor System (see Fig. 1) that receives signals from pairs of surface electrodes applied 2 cm apart on each side of the spine. The two signals are fed into an electromyograph in order to be monitored. The output from the electromyograph is sent to an analog tape recorder and the Distributed Microprocessor System (see Fig. 2).

The Distributed Microprocessor System adopted by us for this application permits fast, reliable, objective EMG activity measurements to be obtained, which is extremely useful in clinical practice.

Method

EMG signals coming from two couples of surface electrodes applied 2 cm apart on the patient's spine are fed into an electromyograph in order to be monitored. The output from the electromyograph is sent to an analog tape recorder and the Distributed Microprocessor System (see Fig. 2).

The asymmetric index (AI) was calculated using the formula:

\[
AI = \frac{R_{AI}}{P_{MAX}} \times 100
\]

where \( R_{AI} \) is the EMG activity of the right side and \( P_{MAX} \) is the EMG activity of the left side.

Results

Measurements of AI have been made on 39 curves from 30 scoliotic patients and have been repeatedly checked, with more than 4 times.

Experiments were performed to test AI variability by repeating measurements in the same scoliotic subject under the same conditions. Results are as follows:

<table>
<thead>
<tr>
<th>Sub</th>
<th>Time</th>
<th>n</th>
<th>Mean</th>
<th>Min</th>
<th>Max</th>
</tr>
</thead>
<tbody>
<tr>
<td>A</td>
<td>6</td>
<td>14</td>
<td>14</td>
<td>14</td>
<td>7</td>
</tr>
<tr>
<td></td>
<td>12</td>
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<td>7</td>
</tr>
<tr>
<td></td>
<td>18</td>
<td>14</td>
<td>14</td>
<td>14</td>
<td>7</td>
</tr>
</tbody>
</table>

The highest range of variability was observed in patient C who showed a difference of 29 between AI max and AI min. Generally it takes less than 2 min to repeat the test 5-10 times, with the difference AI is near 10.

AI was not significantly modified by slight postural changes, and body weight distribution on legs.

This was also confirmed by the occurrence of the inevitable postural changes in patients being due to the lengthy experimental procedure above reported.

Fig. 2 - Block-diagram of the equipments used.

The distributed microprocessor system software performs several tests on all critical devices and the communication protocol between the microprocessors to secure full-safety operation.

The user can select sampling frequency (fs), acquisition time (t) and start command.

The distributed microprocessor system computes partial sum while coupling incoming signals by a double-saturating technique. At the end of the acquisition time, the distributed microprocessor system computes the AI by the following formula:

\[
AI = \frac{ \left[ \int r(t) \, dt \right] }{ \left[ \int s(t) \, dt \right] } \times 100
\]

Fig. 3 - AI as a function of Cobb degrees sites where electromyographs should be applied to effect electric stimulation of scoliotic patients. SFES has given good results at the Centro Don Gnocchi (Parma) with 80% of success on 97 curves from 60 scoliotic patients, at the Instituto "G. Pini" (Milano) and Centro des Meseures (I.I.O.), the results of which can be found in (ref. 1 and 2).

The Distributed Microprocessor System adopted by us is also possible to perform other types of EMG analysis in real time, such as power spectrum, medium frequency, in order to monitor the effectiveness of the treatment being given to the patient, and also for the diagnosis of the disease.

References

A DISTRIBUTED MULTIPROCESSOR SYSTEM FOR REAL TIME MYOELECTRIC SIGNAL ANALYSIS

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(c) Fondazione Don Gnocchi, Parma

Introduction

With the recent improvements of integrated circuit technology there has been an increasing interest in specialized laboratory computers for biological data analysis. The importance of this type of equipment in the EMG signal processing area is growing very rapidly. The architecture of specialized laboratory computers is task oriented, that is these computers are tailored to perform a specific set of well-defined tasks. Task-oriented architectures offer the following advantages over general-purpose computers:

a. better cost-effectiveness because they are task specific;

b. favourable price/performance ratio;

c. much lower specialized hardware costs;

d. computation potential similar to or even higher than that general-purpose computers;

e. reduced space requirements and weight which spell the opportunity to make investigations at a patient's bedside without any difficulties.

A signal processor for EMG analysis clearly belongs to this class of task oriented devices and it should (4):

1. sample EMG signals via several channels;

2. perform basic signal processing in real time, and

3. exchange data with an external device (tape recorder, general-purpose computer) to allow off-line EMG analysis.

In this paper a distributed microprocessor system developed at the Dipartimento di Elettronica di Politecnico di Milano is described.

The System

To perform basic EMG signal processing in real time, EMG signal acquisition and sampling must be overlapped with EMG signal processing. This is obtained in our system by distributing the above two tasks between two subunits operating in a pipeline mode.

EMG signal coming from surface electrodes require a sampling rate up to 5 MHz. This requirement is easily met by only one subunit consisting of basically an A/D converter and a controller device. The same subunit can be adopted when necessary are used instead of surface electrodes to obtain EMG signals.

Difficulties arise with algorithm implementation because the processing unit may possibly be unable to perform the entire operation before next sample is received. To overcome this difficulties two alternative approaches can be adopted:

1. use faster processing unit, or

2. including a higher number of subunit acting in a pipelined or parallel mode in the system.

In this system the acquisition subunit and the processing subunit share a dual-port RAM memory. Data coming from the A/D converter are fed to the dual-ported RAM memory and are processed by the processing subunit by using a double-buffing technique. The number of two-port RAM memory and processing subunit can be increased as such as required to execute highly-demanding algorithms (e.g. Discrete Fourier Transform, Recursive Least Squares Estimation).

The acquisition subunit consists of a signal conditioning module, a sample and hold device for each channel, an analog multiplexer, and an A/D converter connected to a parallel I/O port controlled by a Z80 microprocessor (fig. 1).

The dual-ported RAM memory is shared for the acquisition and the processing subunits. It consists of 4K bytes of static memory that can be accessed in an exclusively way by one of the two CPUs.

The processing subunit consists of a small microcomputer system with a Z80 CPU running at 2.5 MHz, a total 8K of ROM memory and 8K of RAM memory, a serial I/O interface, an arithmetic processing unit (AMD911A), and an IEEE-488 interface. The processing subunit is responsible for output results on a CRT, and/or the transfer of samples or partial results to an external device via its IEEE-488 interface (see fig. 2).

Two different programs are implemented separately for each of the subunits. Tasks to be performed on the data acquisition subunit are:

1. testing every subunit device, channel selection, channel gain control, A/D control, and communication the processing subunit via the dual-ported RAM memory.

The entire system is managed by the processing

Conclusions

The system developed by us is cheap, self contained and can perform simple, real-time EMG analysis in a fail-safe mode, as discussed in (2).

More sophisticated EMG analysis can be easily performed by upgrading this system with additional dual-ported RAM memory and processing subunits without altering its critical software structure as the communication protocol between subunits is always the same.

Bibliography


Fig. 1 - The acquisition subunit block diagram

Fig. 2 - The processing subunit block diagram
LOCOMOTOR DISABILITY IN SPINA BIFIDA

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2. Faculty of Arts and Science, Concordia University, Montreal, CANADA

I. INTRODUCTION

The purpose of this study was to collect information concerning children born alive with Spina Bifida (sacral meningocoele) in the city of Montreal. While a number of European countries (1, 2, 4) maintain registration of congenital deformities, this is not generally the case in Canada except for British Columbia (6). There is, thus, an overall lack of information concerning such individuals. In addition, there is little information concerning older patients who have survived beyond infancy. It was for this reason that the present study was designed as a retrospective study to cover a twenty-year period. Specifically, the subject of the present work is the effect of spina bifida on locomotor ability.

II. METHODOLOGY

1. Case Finding: Information on children born alive with spina bifida, with or without hydrocephalus, in Montreal from 1953 to 1973 was taken from hospital records. Hospital records also provided ongoing medical histories for these patients for the twenty year period and where possible until the end of the study.

2. Data Classification: The level at which the myelomeningocele occurred was recorded, usually in descriptive terms. These have been classified as follows:

- 1. Cervical, cervical-dorsal/thoracic, thoracic thoaraco-lumbar,
- 2. Lumbar, and
- 3. Lumbo-sacral and sacral.

The patient's ability to walk (locomotor ability) was rated on a post hoc basis either at the last entry of information in hospital records (1953-73) or subsequently at the end of the study. These categories are described as follows:

- 1. Not Able to Walk, other than due to post-surgical state or other debilitating condition; or because of age - young children who could only sit, crawl etc.
- 2. Wheelchair dependent for mobility;
- 3. Partial ability to walk; requires assistance at all times, braces, crutches, etc.; and
- 4. Walks independently, can manage stairs but may also use orthoses.

III. FINDINGS

Of a total of 521 live-born patients with spina bifida located in the search of hospital records, only 266 survived beyond infancy. Due to missing data on one or more of the above factors (ranging from 836 cases available for the study of locomotor disability).

Preliminary analysis of these cases showed there were no significant differences in the rated locomotor ability between males and females, or between patients who also had hydrocephalus and those who did not. These data are presented in Table 1.

IV. DISCUSSION AND SUMMARY

The material presented has indicated that locomotor ability in patients with spina bifida is more closely associated with the age of the patient than with other factors investigated. Further, while there appears to be some improvement for those in late childhood and early teen-age years, subsequent determination and increasing dependence on the wheelchair is evident for older individuals. It must be remembered that such observations based on a retrospective study could change with respect to the younger patients as they grow and mature, since advances in treatment as well as improved availability of physiotherapy and occupational therapy could have a beneficial effect on their outcomes.

V. REFERENCES


PREVENTION OF DOWNWARD DISLOCATION OF THE SHOULDER JOINT

Masafumi Kobayashi and Atsushi Taguchi, Division of Orthopaedic Surgery, The Japan Red Cross Society, Nagasaki Atomic Bomb Hospital, and Byodo Suzuki, Nobufumi Ito, Shiro Tazoe, and Masayuki Kondo, Department of Orthopaedic Surgery, Nagasaki University, School of Medicine, Nagasaki, Japan.

INTRODUCTION

Downward dislocation of the shoulder joint does not occur in normal shoulders, but occurs in so-called normal shoulders or shoulders of some of the patients with hemiplegia. We thought that some muscles of the shoulder girdle might help to prevent downward dislocation of the shoulder joint. We examined in electromyographic study which muscles of the shoulder girdle played an important role in preventing downward dislocation.

EXPERIMENT

The subjects were eighteen normal shoulders and eleven loose shoulders, which had the possibility of functioning poorly in joint stability. After due consideration, six muscles were selected and examined electromyographically. These muscles were the upper fibers of the trapezius, the supraspinatus, the middle fibers of the deltoid, the long head of the biceps brachii, the clavicular portion of the pectoralis major, and the rhomboids. The subjects were examined in standing position, relaxed to the best of their ability, with their arms hanging. The electromyographic examination was made with static unloaded arm and static loaded arm. Load was increased from nil to five kilograms, adding five tenth kilograms at a time. The action potentials of the examined muscles were derived, using seventy-micron fine wire electrodes, then amplified, rectified, and recorded with a seven-channel data recorder simultaneously. Next the linear envelopes of the recorded muscle action potentials were compared to facilitate studies involving quantitative measurement of the average amplitudes of the action potentials. Furthermore EMG patterns were analyzed by computer with cluster analysis so that the mutual relations between the increase of load and the muscle activities could be examined.

DATA ANALYSIS

In the nil and light load of about one kilogram, the action potentials of all examined muscles were varied and did not show consistent changes. In the normal shoulders, the action potentials of the supraspinatus were higher and increased by degrees according to the increase of load, and in the upper fibers of the trapezius and the clavicular portion of the pectoralis major, their activities also increased. With the load of over two to three kilograms on these three muscles, the tendency of increasing muscle activities was obvious. While the electromyographic activities of the middle fibers of the deltoid, the long head of the biceps brachii, and the rhomboids were low, and did not tend to increase gradually in proportion to the added load. In the loose shoulders, electromyographic activities of the examined muscles did not usually show the increase compared to the normal shoulders. And at three kilograms of load, all of the loose shoulders had been already dislocated downward (Fig. 1).

The quantities of muscle activities which derived from their linear envelopes were expressed by percentage, and the mutual relations between the increase of load and the quantities of muscle activities were investigated. According to the investigations, in the normal shoulders there were high mutual relations between the activities of the supraspinatus and the load. Also the upper fibers of the trapezius and the clavicular portion of the pectoralis major had higher correlations than the other three muscles; that is, the long head of the biceps brachii, the rhomboids, and the middle fibers of the deltoid. In the loose shoulders, the supraspinatus had lower correlations compared with those of the normal shoulders (Fig. 2). The dendrogram of cluster analysis showed that in the normal shoulders, the supraspinatus were most closely related to the increase of load, among all examined muscles, but on the contrary, the middle fibers of the deltoid were least related. In the loose shoulders, all of the examined muscles were not so closely related as in the normal shoulders (Fig. 3).

RESULTS

An electromyographic study of six muscles of
The shoulder girdle in eighteen normal shoulders and eleven loose shoulders with increasing load added to the area showed that in the normal shoulders, the activities of the supraspinatus were most closely related to the load, but the middle fibers of the deltoid, the long head of the biceps brachii, and the rhomboideus were less related. The study also showed that in the loose shoulders, the activities of the six muscles were low and had little correlations when compared with those of the normal shoulders. In summary, the supraspinatus was most greatly influenced by the load. We concluded from the results of this study that in consideration of the origins and insertions of the examined muscles, downward dislocation of the shoulder joint was prevented chiefly by the joint capsule and the ligaments when the load was less than about two to three kilograms, but it was prevented by the supraspinatus when the load was more than about two to three kilograms.

In other words, the humerus was not pulled up but was drawn toward the glenoid fossa so that downward dislocation of the shoulder joint was prevented.

**Fig. 1** EMG and linear envelope

**Fig. 2** Relations between load and muscle activities

**Fig. 3** Dendrogram of cluster analysis

---

**Table 1**

<table>
<thead>
<tr>
<th>Examined cases</th>
<th>Normal persons</th>
<th>Patients with lumbar intervertebral disc herniations</th>
<th>Patients with tuberous spinous synostosis</th>
<th>Patients with spondylolisthesis</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>56</td>
<td>56</td>
<td>56</td>
<td>28</td>
</tr>
<tr>
<td>Total</td>
<td>210</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Electromyograms were taken with concentric needle electrodes from the gluteus maximus, the iliobial anterior and the gastrocnemius muscles during easy standing position. The consecutive discharges of each motor unit were recorded on the data recorder at once, and all the motor units with amplitude higher than 300 μv were then picked up and each motor unit consisted of 100 units were calculated by one anomaly from the oscilloscope screen regarding to the amplitudes and duration, as shown on Fig. 1.

**Fig. 1** Single action potential

**Fig. 2** Dendrogram of cluster analysis

And then fed to the computer these results. The motor unit discharge was finally displayed as a function of amplitude in the form of correlograms. And next, the same motor unit discharges were displayed again as functions of amplitude and interval variations in the form of diagram.

We had the variable peak to peak amplitudes, such as X1, X2, X3... and forth to Xn-1 and Xn. Secondly, we computed
K Ishiike EMG Discharge Patterns in Muscle of Lower Extremity

The correlation coefficient of X2 and X3, X3 and X4, ..., so forth to Xn-2 and Xn. We did no forth to gain the correlation coefficient (Fig. 3).

Equation of the Correlogram

\[ Y = \sum_{i=1}^{n} X_i + b \]

\[ X = \frac{1}{\sum_{i=1}^{n} X_i} \]

\[ Y = \frac{1}{X} \]

\[ \sum_{i=1}^{n} X_i = \text{mean} \]

\[ n \]

**Fig. 3**

Then we had the correlogram of amplitude, which were represented the time lag on X-axis and serial correlation coefficients on Y-axis (Fig. 4).

**Correlogram of Amplitude**

**Fig. 4**

The diagrams of amplitude variations and interval variations were represented as the discharge numbers on X-axis and peak to peak amplitudes or intervals between each peak on Y-axis (Fig. 5, 6).

**Amplitude Variation**

**Fig. 5**

**Fig. 6**

In general, the correlograms were classified into four types, as shown on fig. 7.

**Normal correlograms of amplitude (Gastrocnemius)**

**Fig. 7**

Type A represents a complete periodic changes regarding to muscle action potentials. Type B and C represents an correlation between the former and the later action potentials and type C represents a more periodic changes with some decrement in action potentials in the consecutive discharges.

**Results**

Much shown in normal persons showed type C in amplitude correlograms. No type A were obtained in our series. It was resulted in the amplitude correlograms in normal persons that type C were obtained in 66.7% at the gluteus maximus, 66.8% at the tibialis anterior and 17.1% of type D and 14.9% of type B were obtained at the tibialis anterior. As the gastrocnemius, type C were shown in 51.0% and 25.6% of type B and 25.0% of type B were obtained. As to the correlograms collectively, type C were shown in 60.9%, type 9 were in 22.1% and type B were in 15.1% (Table 3).

**Correlogram patterns in normal persons**

<table>
<thead>
<tr>
<th>Correlogram</th>
<th>Type A</th>
<th>Type C</th>
<th>Type D</th>
<th>Total</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gluteus maximus</td>
<td>1</td>
<td>16</td>
<td>7</td>
<td>24</td>
</tr>
<tr>
<td>Tibialis anterior</td>
<td>5</td>
<td>24</td>
<td>6</td>
<td>35</td>
</tr>
<tr>
<td>Gastrocnemius</td>
<td>7</td>
<td>14</td>
<td>6</td>
<td>27</td>
</tr>
<tr>
<td><strong>Total</strong></td>
<td>12</td>
<td>54</td>
<td>19</td>
<td>86</td>
</tr>
</tbody>
</table>

**Table 2**

Regarding to the analysis of the amplitude variations of the consecutive discharges, each cases showed small variations of the amplitude in the diagrams in normal persons. The range of the variations of the amplitude were classified into three classes, that is, (-) variation was below 100 µv of the amplitude, (0) was between 100 and 200 µv, and (++) was over 200 µv. The results were shown as follows: (-) variations were shown in 66.8%, (0) were in 19.6% and (++) were 14.6%. As to the diagrams of the amplitude variations correlation, the gluteus maximus showed (-) in 54.2%, (0) in 29% and (++) in 18.8%. The tibialis anterior showed both (-) and (0) in 91.2% and the gastrocnemius showed both (-) and (0) in 85.1% (Fig. 8).

**Amplitude variations Normal persons**

**Fig. 7**

The diagrams of the interval variations were classified into two types, such as (-) variations, which were within 100 msec changes between each discharges, and (0) which were over 100 msec. Almost cases in normal persons were shown (-) variations of the interval variations belong to type C correlogram (Fig. 10).

**Correlogram patterns and Interval variations Normal persons**

**Fig. 10**

On the other hand, as to the types of the amplitude correlograms in patients with intervaradislic disc herniations of 1.4% level, collectively, type C were shown in 17.9%, type 9 were in 66.6% and type B were in 17.5% (Table 3).
A Georgievski Electrokinesiological Testing in Lumbar-sacral Spinal Disorders

Neuromuscular pattern. Electrical silence was not observed during maximal trunk flexion, but was appeared during maximal backward extension. Mean value of amplitudes of EMG activity were significantly greater during forward bending and lower during extension movement in the patients' group than by normal subjects (p<0.05). There was significant decrease in the range of movement in both directions, comparing to the control group (p<0.1). Mean value for flexion movement was 50\% (15-100\%) and for extension 14\% (0-24\%) degrees. The mean time performance was significantly prolonged in patients' group (p<0.1).}

**Data Analysis**

Multifactorial analysis of Variance was undertaken to test the interaction effect of radiologic findings, clinical syndrome, sex and occupation on the electrokinesiological parameters. The F and P test were performed to determine the level of statistical significance of difference between electrokinesiological parameters from the patients and the control groups.

**Results**

1. Electrokinesiological finding in normal subjects (Fig. 1)

Two phases EMG activity was registered during flexion and extension. Electrical silence was registered in positions of maximal trunk flexion and extension. The mean value of the time for flexion movement was 5 seconds and 1.6 second for extension. The mean range of forward flexion was 110 (5-150) degrees and 26 (10-90) degrees for backward extension. No difference was noted between male and female subjects in control group.

2. Electrokinesiological finding in patients with low back pain and sil...
J Yamaguchi et al Study of P Wave in Patients with Intervertebral Disc Herniations

There were no significant differences in the mean value of these shortest latency of P wave between normal and patients with I.D.H.

We investigated the FWCV with the shortest latency and resulted in no statistical differences between normal and patients with I.D.H., as shown on the table 2.

<table>
<thead>
<tr>
<th></th>
<th>Tibial nerve (msec/sec)</th>
<th>Peroneal nerve (meters/sec)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Normal cases</td>
<td>53.14 ± 4.47</td>
<td>50.87 ± 3.54</td>
</tr>
<tr>
<td>I.D.H. Patients</td>
<td>53.87 ± 4.93</td>
<td>N.S.</td>
</tr>
<tr>
<td>P</td>
<td>50.51 ± 6.14</td>
<td>N.S.</td>
</tr>
</tbody>
</table>

Table 2 Mean of the FWCV with the shortest latency F-waves

Fig. 5 Items of the examination

P wave conduction velocity (FWCV) was calculated by Kimura's method (Fig. 5).

FWCV (meters/sec) =

\[
\frac{\text{Distance from stimulus point to Td} \times 2}{(P \text{ latency} - \text{M latency}) - 1} \text{ (msec)}
\]

Fig 3 (From kimura Arch Phys. Med. Rehabil, Vol.56, Nov. 1975)

Results:

In normal subjects and patients with I.D.H., the mean value of P wave with the shortest latency by stimulating the tibial nerve and peroneal nerves are shown on the table 1.

<table>
<thead>
<tr>
<th></th>
<th>Tibial nerve (msec/sec)</th>
<th>Peroneal nerve (msec/sec)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Normal cases</td>
<td>36.54 ± 2.27</td>
<td>39.11 ± 3.19</td>
</tr>
<tr>
<td>I.D.H. Patients</td>
<td>36.22 ± 3.00</td>
<td>39.12 ± 4.08</td>
</tr>
<tr>
<td>P</td>
<td>N.S.</td>
<td>N.S.</td>
</tr>
</tbody>
</table>

Table 1 Mean of the shortest latency of F-waves

![Fig. 4C FWCV of the shortest latency F-wave (Stimulation on the Tibial N: at the knee)](image)

Even though the involved legs of I.D.H., the FWCV were within normal range which were shown by the dotted area on the graph. Some investigations were done concerning the conduction velocity of P wave by stimulating the peroneal nerve in comparison with normal legs and involved legs in the case patients with I.D.H., even though two cases showed out of the normal range in the involved legs, we concluded in no significant difference between them (Fig 5).

...
Comparing with the healthy and involved legs in the case patients with l.s.h., as to the conduction velocity of F wave with the longest latency by stimulating the tibial nerve at popliteal area, the conduction velocity of the involved legs were strong tendency to out of the normal dotted area, as shown on the Fig.

Table 3 Mean of the longest latency of F-waves

<table>
<thead>
<tr>
<th></th>
<th>Tibial nerve (msec)</th>
<th>Peroneal nerve (msec)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Normal cases</td>
<td>39.68 ± 2.52</td>
<td>42.08 ± 2.46</td>
</tr>
<tr>
<td>I.D.H. Patients P</td>
<td>41.05 ± 2.95</td>
<td>&lt;0.05</td>
</tr>
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</table>

Table 4 Mean of the FWCW with the longest latency F-waves

<table>
<thead>
<tr>
<th></th>
<th>Tibial nerve (m/sec)</th>
<th>Peroneal nerve (m/sec)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Normal cases</td>
<td>47.32 ± 4.38</td>
<td>45.74 ± 2.94</td>
</tr>
<tr>
<td>I.D.H. Patients P</td>
<td>44.94 ± 3.58</td>
<td>40.58 ± 5.67</td>
</tr>
</tbody>
</table>

Table 5 Mean of the difference in latency between the shortest latency and longest one

<table>
<thead>
<tr>
<th></th>
<th>Tibial nerve (msec)</th>
<th>Peroneal nerve (msec)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Normal cases</td>
<td>3.16 ± 1.02</td>
<td>2.97 ± 1.44</td>
</tr>
<tr>
<td>I.D.H. Patients P</td>
<td>4.82 ± 1.65</td>
<td>6.80 ± 3.18</td>
</tr>
</tbody>
</table>

Table 6 Mean of the duration and M. C. V. of normal cases and I.D.H. patients

<table>
<thead>
<tr>
<th></th>
<th>Normal cases</th>
<th>I.D.H. Patients P</th>
</tr>
</thead>
<tbody>
<tr>
<td>Duration (msec)</td>
<td>7.53 ± 2.10</td>
<td>8.01 ± 2.21</td>
</tr>
<tr>
<td>P</td>
<td>N.S.</td>
<td>7.54 ± 1.18</td>
</tr>
<tr>
<td>M.C.V. (meters/sec)</td>
<td>52.79 ± 4.02</td>
<td>51.93 ± 3.49</td>
</tr>
<tr>
<td>I.D.H. Patients P</td>
<td>51.82 ± 5.39</td>
<td>50.34 ± 5.42</td>
</tr>
</tbody>
</table>

Discussion

In 1955, Magladery and McGeough described a potential which was recorded at a latency greater than that of the direct motor response in small hand or foot muscles in man when the right peripheral nerve to the muscle was electrically stimulated. This response was called the F wave. While many investigators, such as Casell and Wiesendanger, Mecolead and Vaug, Mayer and Feldman, and Himsa, have reported the P wave in man. 

P wave in man is due to the recurrent discharges of a few motor neurons in the anti-crotonic volley. Recently, some investigators have reported that the conduction velocity of P wave with the shortest latency were delayed in the patients with neurogenic damage, however, the delayed conduction velocity of P wave with the shortest latency have not been demonstrated in our investigations. In our investigations as to the conduction velocity of F wave to compare with normal and the patients with i.d.h., we resulted in not the delay of the conduction velocity of the shortest latency but also resulted in the delay of the conduction velocity of the longest latency with statistical difference on the involved legs in the same patients with i.d.h. 

Fig. 8 shows P wave of both normal and involved legs which were obtained from the extensor digitorum brevis by stimulating the peroneal nerve at the knee in the patients with i.d.h. It is marked fluctuation of the latency of F wave in the involved legs.

The sciatic nerve consist of the 1.4, 15 and 01 nerve roots, that is, so called multiganneral innervated fibers. At the damaged site of the sciatic nerve, the histopathological changes of the nerve fibers should be axonotmesis or axonocodium as proposed by Ranvom. Some stimuli would pass through the normal fibers which would demonstrate a normal F wave with shorter latency and others would pass through the damaged fibers in the sciatic nerve which would demonstrate a deteriorated F wave with somewhat longer latency than normal. As far as the maximum conduction velocity was concerned, there was no difference between the patients with i.d.h. and normal subjects. On the other hand, the minimum conduction velocity showed a marked prolongation in the patients. From these results it is inferred that the damage occurred in the smaller rather than larger motor nerve fibers.
LOG NONLINEARITY IN PROPORTIONAL MYOELECTRIC CONTROL

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* University of New Brunswick, Bio-Engineering Institute
** N.B. Telephone Company, Fredericton, N.B., Canada

INTRODUCTION

A myoelectric stochastic process, E(t), is zero mean which rules out the use of linear processes in the prosthetic control application. Typically a rectifier is used such that a signal moment proportional to contraction level may be obtained. The variance of the moment estimate is a limiting factor in the performance of such control systems. For on-off systems this is usually not too severe. However for proportional prosthetic control the information or control signal, c(t), is continuous over some range and the variance in the estimate of c(t) is of more concern.

LOG NONLINEARITY

For proportional myoelectric control it is useful to use the model for E(t) suggested by Kreifeldt and Yao [1]. It contains the signal c(t) explicitly in the form

\[ E(t) = c(t)n(t) \]

where n(t) is a stationary unit variance myoelectric process. This product form for E(t) is transformed to an additive form by a rectifier followed by a log nonlinearity, i.e.

\[ \log |E(t)| = \log |c(t)| + \log |n(t)| \]

\[ = s(t) + w(t) \]

where s(t) = log |c(t)| is the transformed control signal and w(t) = log |n(t)| is an additive noise process. It is now possible to apply optimal linear estimators such as the Kalman or Wiener filter to estimate s(t). These estimators were investigated in this application by Evans et al. [2]. This paper further considers the estimation errors in this estimate \( s(t) \) of \( s(t) \) and in the estimate \( E(t) \) of \( c(t) \).

ESTIMATION ERROR

Given \( \log |E(t)| \) where the statistics of the stochastic processes \( s(t) \) and \( w(t) \) are known then the Kalman estimator is obtained, [2], such that

\[ \hat{s}(t) = s(t) + \epsilon(t) \]

where the variance, \( \sigma^2 \), of the error \( \epsilon(t) \) is determined by the variance, \( \sigma^2 \), of \( w(t) \) and the Kalman filter parameters. Thus the variance \( \sigma^2 \) is constant and independent of the instantaneous value of \( s(t) \) while the percentage variance is a function of \( s(t) \). This characteristic is evident in Figure 1 which shows \( s(t) \) and its estimate \( \hat{s}(t) \) for \( \sigma^2 = 0.03 \; V^2 \). An interesting property of the log nonlinearity is that \( \sigma^2 \) is independent of the value of \( \sigma^2 \) and depends on the form of the probability density of \( s(t) \).

Using the antilog an estimate \( \hat{c}(t) \) of \( c(t) \) is obtained, i.e.

\[ \hat{c}(t) = \text{Alog} \; \hat{s}(t) = \text{Alog} \{ s(t) + \epsilon(t) \} \]

and the estimate mean square error is

\[ \beta^2 = E[(c(t) - \hat{c}(t))^2] \]

\[ = E[c^2(t)] \left[ 1 - 2 \exp(2.3c(t)) + \exp(4.6c(t)) \right] \]

Since \( \sigma^2 \) is a constant the second expected value will also be a constant, say \( k_0 \). To evaluate \( k_0 \) one must know the probability density function for the antilog of \( c(t) \). Using a Gaussian approximation for this density it is straightforward to show that \( k_0 = 0.2 \) and hence

\[ \beta^2 = 0.2 \; E[c^2(t)] \]

From this result is can be seen that the error in \( E(t) \) is proportional to \( c(t) \) and that

REFERENCES


ACKNOWLEDGEMENT

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REFERENCES


ACKNOWLEDGEMENT

This work was supported in part by National Science and Engineering Research Council Grant A4486.
Functional electrostimulation of muscles is ever widely used in treating infantile cerebral paralysis.

For example, successful correction of abnormal leg movements has been achieved by muscle electrical stimulation, which to a great extent facilitates the walking function.

But movement disorders in cases of infantile cerebral paralysis are manifested by discoordination of leg movements as well as by pathological body swinging on the sagittal and frontal planes in the process of walking. Normalization of sagittal body swinging is possible as a result of treatment, including surgical treatment. It is by more difficult to correct frontal swinging which is the result of complex interaction of the muscles of body and pelvic girdle under conditions of pathological muscle tonus.

In case of normal walking body swinging on the frontal plane does not exceed \( \pm 5^\circ \), while in many patients with infantile cerebral paralysis it is increased reaching in difficult cases \(-30^\circ\). These movements are conditioned by muscle distorsity in the body and pelvis girdle.

Intentional correction of the swinging by the patient is impossible because of the motor pathology.

We have developed an electric orthopaedic appliance for correcting pathological frontal body movements in the process of walking. The set of the orthopaedic appliance consists of: portable 2-channel electrical stimulator, two pairs of electrodes, a sensor of frontal body swinging, brochings and wires.

The electrical stimulator with the battery weighs 250 g, has the size of 33x76x150 mm and produces series of electric stimulating impulses with the amplitude of 25 - 70 V with the maximal instant power of impulses up to 30 VA, the period and duration of impulses are equal to 10 - 25 ms and 30 - 300 ms correspondingly. The latent period and duration of stimulation are equal to 0 - 0,5 and 0,2 - 2,0 s correspondingly.

The electrodes have the size from 20x20 mm to 40x250 mm, they are installed in clinically conditioned combinations on muscles: quadriceps, middle gluteal, latissimus muscle of back. The sensor of frontal body swinging contains two adjustable contacts that react to swinging relative to the horizontal line.

The sensor is placed on the patien's back and in case of body swinging that exceeds a certain prearranged value it provokes the switching of the corresponding channel of electrical stimulator and thus it excites muscles that reduce frontal body swinging.

2 orthopaedic appliances have been tested on 22 patients during 2 hours of walking daily for a month.

The clinical results have shown reduction of pathological amplitude of body swinging on the frontal plane during walking, increase of stability, unelioration of biomechanical parameters of walking, facilitation of walking and subjective feeling of motor comfort in patients.

The model of the man's lower limb skeleton

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Introduction

Gait analyses of the biped is an interesting problem for the kinesiologist as well as for the robotics. The computer-aided methods are based on the multi-link rigid body dynamical model of the antropomorphic mechanism \( \mathcal{M} = \{1,2,3\} \). The segment number depends on the main goal of the investigation and desired accuracy. The model of each lower limb is usually adopted as a three-link rigid body system connected with the ball and pin joints (Fig. 1). The accurate analysis is possible for the determination of external forces (ground reaction) and kinematics determination, but not for the internal forces and torques as well as for the real mechanism of motion of the living system.

The biomechanical model of the man's leg

The five-link model with deformable segments has been adopted as the model of the single leg (Fig. 2).

Fig. 1 The dynamical model of the man's leg (rigid body)

Fig. 2 The biomechanical model of the man's model

Such an mechanical structure is assumed because of the area ground contact (not just the point foot-ground contact); the movement in the ankle belongs to two joints (articularis talaris and articularis subtalaris et articularis talocalcaneonavicularis); and the knee polycentricity with the mass time distribution of the muscles results in the changeable inertia tensor and segment length.

The corresponding mathematical model, based on the principles of mechanics could be expressed in the form:

\[
\begin{align*}
\mathbf{\tau} &= \mathbf{\tau}_d + \mathbf{\tau}_c + \mathbf{\tau}_g \\
&= m_g \mathbf{\ddot{r}}_g + \mathbf{\tau}_c + \mathbf{\tau}_g
\end{align*}
\]
D. Popović: Biomechanical Model of Man's Lower Limb Skeleton

The normal gait dynamics was chosen for the presentation. The detailed Selspot and photo-camera measurements were taken with instantaneous application of the teno-scanner and on a line computer /3/. The data acquisition is prepared (lengths changes, inertial tensor, the knee joint moment pole of rotation position, the foot deformations, etc.), and computer-aided analyses of dynamics realized.

The assumption for the right-thigh ground reaction distribution during a double support phase was adopted due to experience and experimentation /2/.

The comparative analysis of the experimental and numerical results obtained by the application of the rigid body dynamical model and the model introduced in this paper has been made. The space limits of this paper imply that just the GR and displacement CP-ZMP are presented (Fig. 3).

It is easy to point out that through implementation of the biomechanical model better agreement with the experimental results exists. Secondly, it is interesting to show that ZMP during gait activities does not belong to the support area in the single support phase of the gait cycle. That means that during bipedal locomotion activities the body is in unstable position, and by a muscle action the desired one of the body moves in order to bring the bipedal structure in stable state. Just by slow motion the ZMP belongs always to the support area. Normal gait, turning around the vertical axis, sitting down, standing up, slope walking, running and plenty of other normal activities belong to the above mentioned class of nonstable-stable transitions.

References
M Wichers Computerized EMG Analysis in Evaluation of Below-Knee Orthotics

We are planning to do similar experiments, with hemiplegic subjects, since the clinical interest of our project is apart from getting acquainted with computer processing of EMG primarily in orthotics as a part of patient care in rehabilitation.

Conclusions: Thus far we have not drawn final conclusions.
Even if no significant influences of the different orthotic settings can be demonstrated computer processing of the EMG-signal is in our opinion a useful tool in gait or movement analysis.

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EFFECTS OF JOINT IMMOBILIZATION ON MUSCLE ELECTRICAL ACTIVITY DURING WALKING

A.S. VITTSIONOV, K.A. PETROUSHKINSKAYA, USSR, GIP

By means of a specially designed orthopaedic device we have studied effects of immobilisation of ankle and knee joints in sound subjects on electrical activity of shank and thigh muscle under conditions of various walking speeds. It has been found that joint immobilisation leads to vivid decrease and redistribution of electrical activity during the locomotor cycle. This activity decrease is more vivid in case of one-joint muscles rather than two-joint muscles and in case of slow and moderate walking speed, rather than fast one. When the walking speed is increased the stated activity changes are smoothed. We suppose that the decrease of muscle activity during walking under conditions of joint immobilisation results from a weaker afferent current from muscle spindles to the central nervous system.

The obtained data stress the necessity of special muscle training when using orthopaedic appliances that evoke locking of leg joint in the process of walking. Such training may be performed by means of muscle electrical stimulation during certain phases of the locomotor cycle.

INTRODUCTION

Electromyography has become a well-established technique in the assessment of neuropathies and myopathies. In many gait disorders such as cerebral palsy and hemiplegia the EMG patterns during the walking stride are a prime indicator of an abnormal motor pattern. Such patterns are closely related to the cause of the pathological walking pattern we observe or measure. Figure 1 demonstrates the complexity of the diagnostic problem. During stance, as illustrated, the moments of force at the ankle, knee and hip all control the knee angle, and therefore many combinations of moment patterns could result in the same knee angle pattern. Similarly, an equally large number of muscle force patterns could be responsible for the net moment of force at each joint. Thus it is extremely important to have a good direct measure of the motor pattern of each muscle group, and this best estimate is given by the EMG.

METHODS

Eight normal subjects were assessed using surface electromyography on the five muscles during treadmill walking at natural cadence plus slow speeds (1.5 km/hr, 1.25 miles/hr). The EMG and footswitch signals were transmitted via a multi-channel biotelemetry system and their linear envelope signals (Fc = 3 Hz) were analysed over a 32 second walking period. The ensemble averages over the stride period was calculated for each subject, muscle, and speed, and the coefficient of variation = (average standard deviation over stride / mean EMG level over stride) was calculated.

Normalization techniques to average these patterns across the subject population resulted in a considerably reduced variance when each subject’s average EMG level over the stride was set to 100%. The ensemble averages of the patterns for the 8 subjects was then calculated for each muscle and speed.

RESULTS AND DISCUSSION

Figure 2 in the ensemble average of the five muscles for the eight subjects walking at their natural cadence. The level of activity for each muscle had an average over the stride period of 100%, thus for some periods of time the activity may have been at 300%, indicating three times its average level. The coefficient of variation (CV) is listed under each muscle’s name.

Table 1 gives the coefficient of variation of the ensemble average for each muscle and speed and Figure 3 presents the average ensemble pattern at the three speeds.

From Figure 3 we see that the soleus
decreases its activity by about 10% as the subjects slowed down but the pattern remains essentially unchanged. The tibialis anterior also underwent a similar reduction at weight acceptance presumably because the slower velocity required less dorsiflexor force to lower the foot to the ground after heel contact. However, during swing the tibialis anterior had the same level indicating that the forces required to dorsiflex the foot against gravity to clear the ground are not speed dependent. The biceps femoris had about a 50% reduction pointing to the need of lower hip extensor forces at weight acceptance (that assist in reducing knee flexion) and also a similar reduction at the end of swing to decelerate the swinging leg and foot. Vastus lateralis and rectus femoris both reduced their activity by about 50% in both stance and swing indicating a marked reduction in knee extensor forces during weight acceptance and during initial swing.

The coefficient of variation (Table 1) changes with speed are indicative of the flexibility (adjustability) of that muscle during walking. The soleus and tibialis anterior are quite consistent as we slow indicating that the decrease in mean level is accompanied by a similar decrease in variability. The muscles acting across the knee and hip however increase their variability proportionately to the mean. Such variability is indicative of the tremendous flexibility that the hip and knee muscles that was alluded to in the opening comments relating to Figure 1.

Acknowledgement

The authors wish to acknowledge financial support of the Medical Research Council (Grant MA 4343) and the engineering assistance of Mr. Paul Day.

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PREDICTION OF ANKLE MOMENT IN NORMAL WALKING FROM EMG AND KINEMATIC VARIABLES

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INTRODUCTION

Joint moments generated by the muscles during walking have most commonly been calculated using a link segment model and Newtonian equations. Attempts have been made to determine individual muscle or muscle group contributions to the net moments. One such approach involves the use of a mathematical model and a selection of basic criteria in order to choose from a large number of possible solutions (1). A second approach which has been used on simple movements has involved the development of a deterministic model using processed EMG as an input variable (2), and commonly includes functions accounting for muscle length and speed of contraction of the muscles involved.

The purpose of the study was to develop and validate a deterministic model for calculation of ankle moments from EMG and kinematic data.

METHODOLOGY

The full model (Model 4) evolved was:

\[ M(t) + k_1 E(t) + k_2 E(t - T) + k_3 \mu(t) \]

when M was instantaneous moment about the ankle; E(t) and E(t - T) were instantaneous amplitudes of processed EMG from tibialis anterior and soleus muscles; \( \theta \) and \( \theta_0 \) were angles at which static calibrations were conducted to derive static force/EMG relationships for each muscle group; \( \theta_0 \) was instantaneous ankle angle; \( \mu(t) \) instantaneous ankle velocity; \( k_1, k_2 \), and \( k_3 \) were constants relating EMG to force; \( k_4 \) and \( k_5 \) were constants accounting for force/angle and force/velocity relationships.

In Model 1, the static model, \( k_1 \) and \( k_2 \) were set to zero. Model 2 included the effects of difference in angle when compared to the angle of calibration in addition to the static force/EMG relationship; \( k_1 \) was therefore set to zero. Model 3 included the effects of angular velocity in addition to the static force/EMG relationship; \( k_2 \) was therefore set to zero.

Calibration contractions were performed by each of the four subjects while seated on a high chair with the foot firmly held. Without eliciting co-contraction the subject activated the muscle group repetitively, simulating the EMG amplitude and frequency of repetition that was characteristic of his/her normal walking. The EMG and force exerted on the apparatus were sampled, A/D converted and stored on magnetic disc. Following rectification the EMG signal was filtered using a simple pass of a second order Butterworth low-pass digital filter varying the cut-off frequency until a correlation with the force signal yielded optimum values. This frequency was derived for each muscle of each subject and subsequently used to process the EMG signals obtained during the walking trials. Linear regression of force on processed EMG yielded constants \( k_1 \) and \( k_2 \) for each subject.

Walking trials included simultaneous collection of 16mm film cine, force plate and EMG data. The film and force plate data were processed and analyzed to yield instantaneous angles, angular velocities and moments of the ankle. The EMG was processed using the optimal cut-off frequency obtained from the calibration contractions.

Inserting \( k_1 \) and \( k_2 \) into each model, \( k_1 \) and \( k_2 \) were iterated to yield minimal root mean square (RMS) errors between predicted resultant moments and moments obtained from the link segment analysis.

RESULTS

For all subjects the optimal cut-off frequency for the digital low-pass filter ranged between 1 Hz and 1.4 Hz for the soleus muscle. For tibialis anterior, the lowest optimal cut-off was 1.4 Hz and the highest 1.0 Hz. Optimal force/EMG relationships derived from the isometric calibration contractions ranged between 0.2 and 0.6 with a mean of 0.37.

Plots of the input variables, namely the ankle angle, ankle angular velocity and processed EMG of soleus and tibialis anterior obtained during the stance phase for one subject...
8 Olney Prediction of Ankle Moment in Normal Walking From EMG and Kinematic Variables

are shown in Figure 1. The effectiveness in predicting the resultant moment when each model is shown in Figure 2. The full model yielded a RMS error in prediction of 7.9 N.m. in most trials the use of the angle constant with the basic force/EMG relationship reduced the error to near the values achieved for Model 4. Enough exceptions occurred, however, to prevent generalization. A table of constants and RMS errors for the full model in each of the trials appears in Table 1.

![Figure 1. Ankle angle, angular velocity and processed EMG obtained during stance phase of gait and used in models, Figure 2.](image)

**DISCUSSION**

The algebraic sum of predicted moments follows closely the true resultant. Small variations are probably due to non-representativeness of the selected muscle. Model I demonstrates the errors if EMG alone is used; constants are underestimated when the muscle is lengthening and overestimated for the shortening muscle. Although the variability of constants precludes the use of an average value at this time the method provides considerable potential for further development.

![Figure 2. Predicted moments and true resultant moments obtained from each model.](image)

**Table 1. Optimal constants and RMS errors between predicted and true moments at the ankle using full model.**

<table>
<thead>
<tr>
<th>Subject</th>
<th>Trial</th>
<th>$x_2$</th>
<th>$x_3$</th>
<th>RMS Error, N.m</th>
</tr>
</thead>
<tbody>
<tr>
<td>W999</td>
<td>B</td>
<td>0.641</td>
<td>0.0022</td>
<td>7.5</td>
</tr>
<tr>
<td>W999</td>
<td>C</td>
<td>0.033</td>
<td>0.0022</td>
<td>7.9</td>
</tr>
<tr>
<td>W979</td>
<td>F</td>
<td>0.062</td>
<td>0.0023</td>
<td>3.2</td>
</tr>
<tr>
<td>W979</td>
<td>H</td>
<td>0.038</td>
<td>0.0018</td>
<td>9.3</td>
</tr>
<tr>
<td>W901</td>
<td>N</td>
<td>0.020</td>
<td>0.0010</td>
<td>6.2</td>
</tr>
<tr>
<td>W901</td>
<td>O</td>
<td>0.014</td>
<td>0.0017</td>
<td>8.1</td>
</tr>
<tr>
<td>W902</td>
<td>A</td>
<td>0.006</td>
<td>0.0026</td>
<td>5.6</td>
</tr>
<tr>
<td>W902</td>
<td>B</td>
<td>0.023</td>
<td>0.0010</td>
<td>7.0</td>
</tr>
</tbody>
</table>

**REFERENCES**

MOMENT AND WORK OF THE CALF MUSCLES IN WALKING ON AN INCLINE, DETERMINED WITH EMG PROCESSING.


INTRODUCTION

In level walking the calf muscles, M.M. soleus and gastrocnemius, exert a moment during almost the whole stance phase. In the first (and usually greater) part of stance the ankle is dorsiflexed and negative work is done on the muscles. This is followed by a quick plantarflexion, the push-off, in which the calf muscles perform positive work (1). According to many authors the main function of the calf muscles in the dorsiflexion phase is to slow down the forward velocity of the body (4). The function of the push-off is less clear. In the view of A. L. Cavagna and Margaria (3) it serves to push the body forward and upward.

When walking uphill, positive work has to be done in each step to raise the weight of the body. In downhill walking the moving body has to be slowed down. Walking uphill should therefore result in increased muscle activity in push-off, while in downhill walking the activity in the dorsiflexion phase should become more prominent.

METHODS

A pilot experiment was done with two male subjects, A and B (age 32/37 yr, ease 91/69 kg, stature 1.91/1.77 m, leg length 1.00/1.04 m, respectively). They walked on a treadmill with a speed of 1.00 m/s at gradients of 0, +5, +10, +15 and +20%. The step length was left free. It turned out not to vary much with slope: sub A: 0.70-0.76 m, sub B: 0.56-0.63 m. EMG was recorded by means of surface electrodes, ankle position by means of an electromagnometer. Moment and work of the calf muscles were determined by EMG to force processing (2). For normalisation the work W has been given per kg of body mass, W/m, and the moment M as a fraction of the reference moment M_ref, the moment in quiet standing on one leg with the heel just off the floor.

RESULTS

Part of the effects in question can be observed by a mere inspection of the raw EMG in relation to the ankle angle φ (Fig. 1). In downhill walking there is a continual EMG activity during the whole dorsiflexion phase. Walking uphill there is virtually no EMG in the first part of dorsiflexion but a pronounced burst before push-off. 

EMG to force processing yields quantitative results. The moment M in downhill walking is about equal to the reference moment (dashed line in Fig. 1) during the first phase of stance. Combined with an extensive dorsiflexion, this results in an appreciable amount of negative work (W/m = 0.4 J/kg at +20%). In level and uphill walking the moment during dorsiflexion is smaller and lasts shorter, W/m therefore decreases gradually with increasing slope down to 0.1 J/kg at +20%.

The moment has a maximum, M_max, at the onset of push-off, which increases from 0.018 m at ref. (no peak) at steeply downward slope to a pronounced peak with W/m = 0.015 to 2.0 at a slope of +20%. The walking level walking is halfway in-between. The positive work W/m is low (0.1 J/kg) downhill, but increases sharply for uphill walking, to 0.5 J/kg at +20%.

The effects described are also reflected in the registrations of the muscle power P. At negative slope there is a prolonged negative P, due to the slowing down action of the muscle. The push-off corresponds with a positive peak in P. It is curious that this peak increases only twofold in height, from 150-300 W, while W increases from 3 to 44 J (at 0A and +20% respectively), so the greater work is largely due to a longer duration of the effective push-off.

DISCUSSION

In spite of our modest size these experiments provide a few quantitative data which were not available so far. The results are in accordance with the assumed functions of the calf muscles in both phases of stance, as put forward in the introduction. Following the indicated line of thought may eventually lead to more direct evidence.

Changes in Electromyographic Pattern of Gastrocnemius Muscle With Gait Walking Speed

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INTRODUCTION

The population variability of electromyographic (EMG) gait patterns and changes affected by walking speed are important normative data, and has been reported recently [1]. This report presents the results of a quantitative procedure for analyzing these various patterns and for associating them with different walking speeds.

METHODOLOGY

Measurement

The foot-contact sequences and the surface electromyographic patterns of lower extremity muscles were measured previously on twenty-five (25) normal individuals over a range of self-selected walking speeds [1]. Ninety-five (95) patterns from the gastrocnemius muscle were analyzed.

Pattern Analysis

The patterns were analyzed using factor and cluster analyses. The gait cycle is divided into sixteen (16) equal time segments and each EMG pattern is depicted by a sixteen element vector. The value of the elemental variable is equal to the ratio of time that the muscle is active during that segment. The dimensionality was reduced using factor analysis on the time covariance matrix (Karhunen-Loeve expansion). A four dimensional feature vector was required to account for more than 75% of the variance and produce clustering [2].

The cluster analysis of the patterns was performed by applying a K-means algorithm on the feature vectors [3]. A decision about the significant number of groups is based upon the R-ratio. It approximates the F-ratio and is a measure of the reduction of the within-cluster variance by using K1 instead of K partitions (clusters). The ratio is

\[ R = \frac{\text{SSE}(K)}{\text{SSE}(K+1)} = \left( \frac{\text{M}}{\text{M} - \text{K} + 1} \right) \text{FM} \]

where \( N = \# \) of patterns, \( M = \# \) dimension of feature vector, and \( \text{SSE}(K) = \) is the total within-cluster variance of \( k \) patterns and \( k \) clusters. The F distribution cannot be used because the partitions are formed to minimize \( \text{SSE}(K+1) \). As a guideline, values of \( R \) which are greater than 10 justify expanding the partition from \( K \) to \( K+1 \) clusters.

Analysis of Variance

After the appropriate partition was found, an analysis of variance of the speeds and relative speeds (speed/body height) associated with the patterns in each cluster was performed [4]. Three sets of calculations were undertaken to test for:

1) equality of group variances using the Bartlett test and chi-square statistic.
2) equality of group means using the ratio of between-group (BGMS) to within-group (WGMS) mean square and an F statistic.
3) inequality of pairs of group means using a Studentized range test.

RESULTS

The number of clusters chosen as significant was based upon a plot of the R-ratio vs. the number of clusters. It was found that correct partitioning with the fewest clusters was formed for the value of \( K \) for which the R-ratio had a second peak over 10. The second peak occurred at \( K = 7 \), which means that a partition of 7 clusters was suitable.

Figure 1 presents the results of the pattern analysis. The group number, the average (AVE) and variance (VAR) of speed and relative speed, the average group pattern and its percentage of the total population are tabulated. The patterns are plotted in the order of increasing speed. The group numbers were assigned randomly by the clustering algorithm. Listed also are the analysis of variance statistics.

DISCUSSION

A general trend with speed exists within a majority (75%) of patterns. Activity tends to commence earlier in the gait cycle as speed increases. Stance phase reduces from 74% to 40% as speed increases. In fact, at faster speeds the activity commences during late swing. Examine the patterns of groups 4, 8, 3, B, and 7. Notice that the main phase of activity always ceases before the end of stance and at the end of stance (50%) within the free and faster speed ranges. The pattern occurring at the fastest speed (Group 7) has a second phase of activity during the stance-to-swing transition. All these groups have significantly different speeds.

CONCLUSION

The EMG pattern analysis of the gastrocnemius shows that there are several distinguishable types. Comparison of the group speeds reveals that the majority of types occur within different speed ranges.

REFERENCES

Introduce

Kinesiologic gait studies have progressed remarkably, but performing gait analysis with patients is not practical because patients cannot walk very many times because of pain and other physical problems. On the other hand, if many parameters are recorded, it will require considerable effort and much time to analyze the data.

The purpose of this study is to develop a simple technique especially in relation to reproducibility of walking.

To realize this purpose there are several points which must be considered:
1. How many deviations are there in the walking cycles of the same subject?
2. How many frames are needed to analyze the walking.
3. How many walking cycles are needed for the purpose of clinical treatment.

Materials and Methods.

1. Fifteen walking cycles were recorded for the same subject by Nagasaki University gait analyzing system (Time duration 1/60 sec., 1/30 sec., 1/24 sec.)
2. Angular changes of body rotation, shoulder joint, elbow joint, knee joint, ankle joint and MP joint of the foot were measured from serial photographs taken with stick picture camera.
3. To calculate the standard deviation from different walking cycle, Cramer's rule was applied and a pc-8000 computer was used.
4. The problem that how many frames and walking cycles are needed for exact gait analysis was studied by the mean pattern and the simulation.

Result

There were almost no noticeable large deviations in shoulder girdle, shoulder joint, hip joint, pelvis, pelvic tilt and ankle joint with respect to their angular changes and pattern.

In the elbow joint, long time duration (1/36, 1/24) exhibited smaller angular changes compared with a short time duration (1/60 second) but the pattern was exactly the same.

The knee joint exhibited the same pattern among the three traces but each peak was different.

The metacarpophalangeal joint of the foot is very important joint, but it is difficult to analyze its angular changes exactly. The standard deviation was larger than for other joints and long duration especially 1/24 sec. trace was situated for outside one standard deviation.

Summarizing these various results, there were almost no abnormal patterns in human walking, and therefore, 1/24 sec. time duration is evidently sufficient to analyze human walking and also these results were proved by the simulation (Fig. 2, 3).

The last problem, how many walking cycles are needed for clinical treatment, was studied by the method as shown in Fig. 4.

The mean pattern of the angular changes of fifteen walking cycles are shown by solid line with the standard error indicated by the solid line. The dotted traces are trial trace of walking cycles. The most important thing is whether these traces are situated for the standard error of fifteen walking cycles. In this study, we proved that five walking cycles were sufficient to analyze human walking.
RELATIONSHIP BETWEEN THE SUBTALAR JOINT AND THE CONTROL OF MEDIAL-LATERAL BALANCE IN WALKING

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Introduction
The human gait is characterized by the smoothness of the displacement of the body center of gravity along the progression path. In this erect bipedal walking, it is very important to control the medial-lateral balance so as to perform the smooth forward movement. In this paper, we mention the control of the medial-lateral balance in walking from viewpoints of the ground reaction forces, the EMG activity, and the motion of the subtalar joint.

Methods
The ground reaction forces, the EMGs, the angular changes of the subtalar joint, and the foot switch signal were recorded simultaneously by data recorders and a pen-writing recorder. Recording of the ground reaction forces (GRF) in three directions (vertical, forward-backward and medial-lateral) were made using Watabe's force plate.

The EMG activities of various muscles were noted with the use of surface electrodes placed over the motor points of the paravertebbral muscle (PVW), glutus medius (G. med.) and peroneus longus (PL), respectively. Indwelling fine-wire electrodes were inserted into the muscle bellies of the tibialis posterior (TP), extensor hallucis longus (EHL), extensor digitorum longus (EDL), flexor hallucis longus (FHL) and flexor digitorum longus (FDL). These EMG signals were transmitted by telemeters and the raw analogue EMG recordings were integrated (time constant: 90 msec.).

Results
In the present experiment, the medial-lateral component of the body appears to be the early period of the stance phase and TP makes its activity weaker after heel contact. In contrast, when D is shorter, the medial-lateral component is smaller, and TP, FDL and EHL continue their activities throughout the period of T3 and T4 so as to keep the body balance. Above findings show how the subtalar joint plays an important role to control the medial-lateral balance in walking.

Discussion
In the human gait, the center of gravity displaces to the susceptible side, and is confined to the inside of the dynamic point of the stance foot (C). With the altering of the distance (D) from the locus of the center of gravity of C, the control of the medial-lateral balance in walking should be converted. When D is longer, in order to...
This paper aims to report about the significance of arm swing and body rotation in the total pattern of gait by measuring angle changes of the upper extremity joints, shoulder girdle and pelvis, and by simultaneously recording EMG of the corresponding muscles and floor reaction force.

Materials and Methods
Ten normal adults and fifteen patients with surgically fused lower extremity joints were examined. On these examinees the following studies were simultaneously performed during walking on the floor without footwear:

1) Serial pictures of the angles of the shoulder, elbow and hip joints as well as rotation moments of the shoulder girdle and pelvis in the horizontal plane were taken with a stick picture camera.

2) Electromyographically examined muscles were the biceps brachii, the brachialis, the triceps brachii, the tibialis anterior, the extensor digitorum longus and the flexor hallucis longus.

3) Three vectors of floor reaction force were measured and recorded using Matade's plate.

The results of the study will be mentioned in this announcement.

Introduction
The function of the metatarsophalangeal joints (MP joints) of the foot is considered to play an important role in walking.

We have reported on the motion and role of the MP joints during customary walking at the VIIIth International Congress of Biomechanics in 1981. In this paper it is proposed that the role of the toes was only accessory and their function was to act as roll off agents maintaining floor contact during the stance phase. The present study was performed to investigate the function of the MP joints in various types of level walking which were composed of customary, slow, fast and sliine gait styles of walking.

Materials and Methods
Five normal adult volunteers, whose ages ranged from 20 to 40 years were used for the study. Each subject was asked to walk 10 times with bare feet in each of the walking styles on a flat table, in which the foot was fixed on a force plate. By stepping on this force plate, the forward-backward and lateral components of the floor reaction force were recorded respectively by a pen-writing oscillograph and a tape recorder.

Simultaneously recording EMG of the extensor hallucis longus (EHL), extensor digitorum longus (EDL), flexor hallucis longus (FHL) and flexor digitorum longus (FDL) muscles were recorded by intramuscular wire-electrodes. Angular changes of the MP joint were measured by our new jig consisting of a piece of stainless steel about 10 cm long and 6 cm wide with a strain gauge attached on both the ends, and was bridged across the MP joint by binding the both ends onto the dorsum of the foot and phalanx. Foot switches were attached to evaluate the floor contact of the foot.

Experimental analysis
Fig. 2 shows simultaneous records of EMG, MP angle and forward-backward component of floor reaction forces. The 1st MP joint begins to flex rapidly as the heel strikes the ground and maintains an extended position of about 10 degrees during the foot flat phase. From the start of the heel off it begins to extend and shows a peak just before the toe off.

FHL and FDL are not active at the moment of the toe off, and the peak of the backward component of the floor reaction force appears prior to that of the peak of extension of the 1st MP joint.

From the above it would seem that the toes do not participate in the act of push off in customary walking. During 10% of the stance phase the central part of the foot is already in contact with the ground, and toes would not be required as propulsion force. Moreover we aimed to study the function of the MP joint in slow, fast and sliine gait styles of walking. Main points were measured to compare with customary walking and other styles of walking (Fig. 3).

Ts and TBP are the ratio of the floor contact time of forefoot in stance phase. T-FHL and T-FDL show the ratio of the time duration of muscle activities of FHL and FDL in the stance phase. MP is the peak of the angular change of the 1st MP joint. B-B is the peak of the backward component of the floor reaction forces.

If each of the parameters was increased, we suggested that the toes would be required as propulsion force. Ts, 76-7, B-B of each walking style were increased significantly according to velocity with the exception...
M Fujita et al. Motion and Role of MP Joints of the foot in Level Walking

of T5 in the longer stride style of walking, T-PLL and T-PDL were increased significantly compared with the custom-
mary walking style with the exception of T-PLL in the fast walking style.
In the fast walking styles, T5, T6-7, T-PLL, MP and F-B were increased significantly. It was supposed that the toes
were to be required to exert propulsion forces during the end of the stance phase.
In the slow walking styles T-PLL and T-PDL were increased but T5, T6-7, MP and F-B were decreased significantly.
The above data suggested that the toes would be required as propulsion forces but stabilizing forces to maintain
floor contact.
In the longer stride walking styles T6-7, T-PLL, T-PLL, MP and F-B were increased significantly. It was sup-
poused that the toes were required as propulsion forces.

Conclusion

The motion and role of the MP joint were studied.
In customary walking, they are only accessory and act as roll off to maintain floor contact.
In fast and longer stride walking styles it was suggested that the toes are required as propulsion forces.
In slow walking styles the toes are not required as propulsion forces but as stabilizing forces.

References


Fig. 1. Flow Chart of the Methods

Fig. 2. Simultaneous Records of EMG, MP angle & Floor R.F.

Fig. 3. Points of Evaluation in EMG, F-DL, MP angle and Floor R.F.

Fig. 4. Comparison with Customary Walking and Other Walking Styles

Introduction

Walking on tiptoe is used for training of the limb muscles in orthopedics and rehabilitation medicine. However, the gait is influenced by plantar flexion angle of the ankle joint.
This paper aims to analyze the gait with slight plantar flexion of the ankle (low tiptoe gait) and the gait with maximal plantar flexion of the ankle (high tiptoe gait) comparing with customary walking.

Materials and Methods

1. Normal adult men without shoes were asked to walk with customary gait (Type A), low tiptoe gait (Type B) and high tiptoe gait (Type C) on a platform for walking examination.
2. Angle changes of the hip and knee joints were measured with a 3D picture camera which can photograph 150 pictures per second in maximum. Angles of the ankle and first MP joints were measured with electrogoniometers consisting of a steel spring and a strain gauge.
3. Changes of the length from calcaneal tuber to the first metatarsal head (longitudinal arch of the foot) and the distance between the first and fifth metatarsal heads (Transverse arch of the foot) during a walk were measured with special equipments consisting of steel springs and strain gauges.
4. Vertical, forward-backward and lateral components of the ground reaction force during the stance phase were measured using Matakö's force plate which was embedded in the platform for walking examination.
5. Action potentials of the 16 muscles were recorded on a pen writing oscillograph using surface and/or fine wire electrodes and linear envelopes were made.
1) M. tibialis anterior
2) M. gluteus medius
3) M. gluteus minimus
4) M. rectus femoris
5) M. soleus
6) M. biceps femoris
7) M. gastrocnemius
8) M. peroneus longus
9) M. tibialis posterior
10) M. flexor digitorum longus
11) M. flexor hallucis longus
12) M. extensor digitorum longus
13) M. extensor hallucis longus
14) M. flexor digitorum brevis
15) M. abductor hallucis

Results

1. Range of motion of the ankle joint during the stance phase was as follows:
Type A: 10° of dorsiflexion.
20° of plant. flexion.
Type B: 0° - 25° of plant. flexion.
Type C: 25° - 40° of plant. flexion.
Among three types of gait, remarkable differences of hip and knee motions during walking were not found.
2. Breath of the foot fair increase in stance phase in every type of gait. However, in the type A, the maximum peak existed in the stance phase, while in the remaining two types, peak of the reception phase was higher than that of the stance phase.
Foot length increased only slightly in the stance phase in every type.
3. Vertical component of ground reaction force had two peaks corresponding to reception and thrust phases.
In the type A, two peaks were almost equal in height, but in the types B and C the breaking force was stronger than the driving force. Curves of forward-backward component showed...
almost the same pattern in each type of gait. Many variations were found in the curves of the lateral component even in the same type of gait.

4. Muscular activities of the rectus abdominis, sartorius, glut max., biceps fem., and add. halluc., were found to have almost the same pattern respectively in every type of gait. Activity of the glut. med. had a peak at the reception phase in the types A and B, while it maintained plateau level throughout the stance phase in the type C. The rectus fem. also maintained moderate and strong activities in the types B and C respectively during the stance phase.

The gastrocnemius had double peak of activity in the type A and B, but it maintained strong continuous activity throughout the stance phase in the type C.

The tib. ant. was activated chiefly in the swing phase in every type of gait, but it showed moderate and strong activities in the types B and C respectively in the stance phase.

In the types B and C the psoas, long., tib. post., flex. dig. brev. and long toe muscles maintained continuous activities during the stance phase, which were different from those in the type A.

Conclusion

In tiptoe gait various muscles were seen to be activated to stabilize the lower extremity joints, especially the ankle and the foot, and to maintain body equilibrium in the stance phase. These activities were more exaggerated in high tiptoe gait.

Ground reaction force in the stance phase was strongly influenced by the muscular activities. However, the activity of the other muscles was scarcely influenced by walking types in normal adults owing to rigidity of the arches.

The results of the above mentioned experiments are considered to be useful for evaluation of equilibrium of the foot and planning of muscular training.

INTRODUCTION

The human gait has been investigated by many authors with different techniques. Many have been the goals of these analyses: to set up appropriate instrumentation, to investigate and differentiate normal and pathological patterns, to know the mechanism of motor coordination. An important field of study is the gait of children as the knowledge of the biomechanical modification produced by the growth, play an important rule for the prevention of pathologies of the musculo-skeletal apparatus. The aim of this study is to furnish a contribution in this area through a particular method based on the on-line elaboration of the forces exerted by the foot on the ground during walking.

METHOD

Twenty two children, eleven males and eleven females, whose age ranged from 12 to 13 years, were enlisted for the study. After a visual inspection all them appeared to walk without remarkable defects despite one subject was affected by mental limitation without physical disorders and other three declared light impairments to the lower limbs. The children were requested to walk at the fixed cadence of 100 st/ min. The forces exerted during the stance phase were measured with a piezoelectric force plate and the relative electrical signals, proportional to the force components, were computed with the technique proposed by Pedotti (1977). The results, named vectograms, are obtained directly on-line in the sagittal or frontal plane and show the evolution of the ground reaction force with its amplitude, inclination and point of application, in discretized form. Five vectograms were recorded in the sagittal plane for each limb of the subjects; the considered steps were those obtained when the steady state and the cadence were respected.

All the vectograms of the children were compared with those of healthy adults, largely described in previous works (Roccardi et al., 1977; Cova et al., 1980). The main characteristics of normality are as defined: a) high repeatability of the patterns of each subject; b) two equilibrated maxima located during the impact and the pushing phases, and a minimum between them; c) a continuous variation of the vectors' inclination from the backward to the forward direction (excluding the first impact); d) the absence of reversal in the displacement of the point of application from the heel to the toe (excluding the first and last vectors); e) the symmetry of gait due to the high resemblance of the patterns obtained from both the legs.

RESULTS

The results showed that all the children verified the high repeatability of the vectograms. By considering separately the groups of vectograms obtained by each limb twenty-two of these groups were associated to a normal shape. Fig. 1a reports two examples of normal, where the before cited characteristics are well illustrated. All the vectograms of the children with impairments were recognized although the analysis was conducted without considering the physical conditions. Fig. 1b shows those of a boy affected from bilateral talalges; the incorrect concentration of vectors and the corresponding reduction of the maxima, during the impact indicate the painful contact. At least sixteen patterns were defined 'border-line' as their shape presented some differences in comparison to the normal one and, at the same time, lacked of any evident pathological feature. The children with normal vectograms in both the legs were eight and two of them showed an appreciable symmetry of gait while six were characterized by various asymmetries due to the
R. Rodano: Analysis of Gait in Boys and Girls with Vectogram Technique. First Results.

Demonstrated useful instrument for a mass analysis of human gait. The method moulds in short time results easy to interpret and with an high contents of information; moreover it permits to well distinguish the pathological patterns even if the impairments are light.

The geometry of gait typical of normal adults was verified by only two of the eight children with normal vectograms in both the legs. This suggests that the bodily growth do not permit the complete assimilation of optimal mechanism of control in relation to the continuous change of the musculo-skeletal apparatus.

By observing the relatively low percentage of vectograms sketched as normal (50%) and the consistency of the groups defined border-line (34%) raises a question: are the border-line indicating unknown dynamics of normal gait or early advances of latent pathologies? The importance of this question require a precise answer for an optimal use of the method in diagnostic.

In order to answer the question a statistical approach, able to transform the visual information of the vectograms in terms of numerical values, was conducted without significant results. Three are the possible explanation: the low number of patterns, the little amount of total variability described by the only equation utilized (44%), the exclusion from the initial variables of the point of application made. In this phase, since the point of application is a function easy to recognize by looking at the vectograms but very difficult to transform in a simple mathematical form.

REFERENCES


CONCLUSIONS

The vectogram's techniques has been...
BALANCE TRAINING BY MEANS OF BIOFEEDBACK

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INTRODUCTION

Functional electrical stimulation (FES) of paralyzed extremities enables secure standing to completely paraplegic patients. The candidates for such a rehabilitation process are the patients with central nerve lesions having spastic paraplegia. Two-channel stimulator was built providing contraction of both knee extensors and thus locking of both knee joints. The next step in the rehabilitation process with FES is synthesis of a stable walking pattern. First experiments of paraplegic patients walking were already performed, most of them in the parallel-bear. To enable effective walking also with walking frame or even crutches the biofeedback learning of standing was introduced.

The standing balance is influenced by the feedback signals from the vestibular organs, the eyes, the muscle proprioceptors and tendon receptors. The paraplegic patient has vision and activity of the vestibular organs preserved, while no sensory information arrives from muscle and tendon receptors. Because of the paralyzed muscles of lower extremities paraplegic patient can control his standing position only by the movements of trunk, head and arms.

INSTRUMENTATION

The most important part of the biofeedback instrumentation for standing balance training is the precision pendulum. A small pendulum is built into a special housing filled by a viscous fluid. A potentiometer is placed in the axis of the pendulum rotation. The properties of the precision pendulum are: resistance 2 kΩ, range of motion ± 45°, accuracy ± 1° and frequency ± 2Hz. In Fig.1 precision pendulum attached to the sacral part of a paraplegic patient is shown.

The rest of instrumentation consists of electronic circuits and two loudspeakers. The amplified signal from the precision pendulum is split into positive and negative part, representing forward and backward leaning of the body. The voltage proportional to the angle of the leaning is transformed into the sinusoidal signals of different frequencies which are led to the two loudspeakers. The electronic part of the biofeedback instrumentation is presented in Fig.2. The zero position, dead zone (when both loud-speakers are silent) and volume of the tone can be adjusted on the front panel. The instrument is showing the deviation from the central position. Two light emitting diodes are depicting either leaning forward or backward.

RESULTS

Four paraplegic patients participated in the program of the balance training by the biofeedback: patient R.S. with complete paraplegia at T-9 level, patient P.D. having complete T-9,10 spinal cord lesion, patient M.C. with incomplete paraplegia at T-10,11, and completely paraplegic patient H.L. with the level of injury at T-12, L1. All the patients suffered from the accident which had occurred in the second half of 1981. While standing their knees were locked by mechanical braces. The training lasted for half an hour each day. The success of the training was estimated by measuring the time when patient could stand with the support of only one arm or even without any arm support while both loud-speakers were silent.

The patient R.S. suffered from severe spasms. At the beginning he was able to stand with one arm support for 7s. After 14 days of training this time was increased to 1 min 3 s. This patient didn’t succeed to stand without the arm support. The second patient P.D. started with 13 s with one arm support and ended with 6 min 35 s after twenty days of training. He was also able to stand for 3 s without the arm support. The training did not help to improve this result. The incompletely paraplegic patient M.C. was already after 3 days of training standing for 3 min 15 s with one arm support. With no arm support he started with 4 s and finished after 25 days of training at 4 min 49 s. The last patient’s standing time with one arm support was 8 s in the beginning and 6 min 10 s after 17 days of the biofeedback training. Without the arm support he was not able to stand for more than 3 s.

CONCLUSIONS

The biofeedback balance training was added to the other physiotherapeutic methods improving standing in spinal cord injury patients. The patients tested liked the procedure described. The method is particularly efficient with the patients who have chance to stand unassisted after the exercise procedures. The results obtained are well correlated to the status of the trunk muscles.
The force arm for a push off is the part of the foot reaching from the ankle joint complex to the metatarsophalangeal joints. In the human foot the row of metatarsophalangeal joints is divided by the protruding head of the second metatarsal bone so that two axes exist at this level, a transverse axis through the heads of the first and second metatarsals, and an oblique through the heads of the second to the fifth. The distance from the ankle joint to the oblique axis is about 20% shorter than the distance to the transverse axis (Bojesen-Møller, 1978). The oblique axis will therefore be preferred unless the foot is forced by muscles to use the longer arm.

Movements about the oblique axis is combined with inversion of the foot, while movements about the transverse axis is carried out in eversion. It is thus the balance between invertors and evertors which selects the axis. In order to determine the activity of the peroneus longus and the flexor hallucis longus as two important muscles for this timing of the foot, a combination of chronophotography of the push off and electromyography of the muscles was applied. The two techniques were combined so that the events could be displayed and correlated with an accuracy of 5-10 msec (fig. 1).

Methods
A signal generator was constructed which could emit impulses at 5 and 50 Hz. The low frequency was used to trigger a stroboflash for photography while the high frequency was fed into a chain of four light-emitting diodes, which, attached to the shank and foot, could form four light tracks on a still photo. From the high frequency signal one impulse was removed each second and with the signals recorded on an oscillograph the missing signal could be reconstructed in the series of light tracks on the photograph as well as on the oscillograph. Simultaneously, electromyographic signals were recorded from the peroneus longus and the flexor hallucis longus muscles. Periods of activity in the two muscles could thus by way of the missing signal be correlated to the light tracks and the phases of the push off.

The light-emitting diodes were placed at the tip of the great toe, at the metatarsophalangeal joint, on the heel below the malleolus medialis and on the middle of the shank. The surface electrodes were placed laterally on the shank at the middle of the peroneus longus muscle, and behind the malleolus medialis, where a flashy part of the flexor hallucis longus muscle is superficial. A third electrode was placed on the tibialis anterior. The proper location of the electrodes was checked by stimulating the muscles and controlling the peripheral effect of the contraction. The other leg was covered by a black stocking.

Results
With the signal generator set at 50 and 5 Hz, the diodes will form light tracks in which each light and dark period lasts 10 msec, while the strobe light will show the position of the leg and the synchronism between the four light tracks each 200 msec.

During stance phase the foot is arrested by its contact with the ground while the shank moves on continuously. The diodes on the shank forms therefore a light track in which a full time scale can be marked, while the diodes on the foot shows the phases of the foot and the centers of movement. This is illustrated in figure 1 in which time zero is set at the first deflection caused by the heel strike and in which the phases are followed to lift-off, in this case 820 msec later.

Fig. 1. Still photo with four light tracks and corresponding electromyogram showing the stance phase of a left foot. The light tracks are formed by diodes which are supplied by the 50 Hz signal shown on channel 4 on the EMG. Time 0 is set at the first deflection of the light tracks caused by heel strike. The five Hz stroboflash shows the position of the leg and foot at 20, 220, 420, 620 and 820 msec. The missing signal which in this case is found at 640 msec is used to correlate the timescale formed by the diode on the shank with that on the EMG. Elevation of the heel and metatarsophalangeal joint occurs at 480 and 740 msec, respectively. The EMG shows the rectified and averaged signals from the tibialis anterior (t.a.), the peroneus longus (p.l.) and the flexor hallucis longus (f.h.) muscles.

Push off in eversion (transverse axis).

Fig. 2. Diagram showing the coordination of the flexor hallucis longus and the peroneus longus muscles during ten push offs in eversion (transverse axis) and ten push offs in inversion (oblique axis). Same individual. The contact phase is normalized into 100 time units.
Since four years automatic analysis of ANODE computer has been used in our department. The analysis is given in a form of 5 histograms describing a complete EMG examination. Three parameters evaluate single motor unit potential (MUP) and two other maximal effort pattern. Using different friction of a subject's maximal effort the diagnostic yield of this method has been checked. Single MUP parameter covers from 60 to 90% of diagnostic power whereas the maximal effort pattern covers only 10 to 30%. The most diagnostic value has such friction of maximal effort at which single MUP are still to be identified. This depends on the type of neuromuscular disorders and the course of disease and corresponds to 10 to 30% of maximal force.

In Rehabilitation a major role is represented by the prevention of sequelae in hemiplegic patients. Among them one of the most frequent and severe is the hypertonic evolution of the upper arm in patients with cerebrovascular diseases (CVD) featuring a so called drop shoulder (Colombo et al., 1970). Investigations have been already carried out by several authors in patients with CVD to spot the presence of "sick motoneurones" (McComas et al., 1973). Especially American literature suggests a peripheral involvement (Moskowitz and Porter, 1963), (d'Ors and Lebret, 1967; Bhatia, 1969), (Kruiger and Maylonis, 1973; Zalis and al., 1976). On the other hand we can find in literature data suggesting a not complete agreement upon the presence of lower motoneurone diseases (Ishii, 1968; Alpert, 1971, 1977). A certain diastomogenic criteria in the patient's choice has been noted too. Both in different authors and from patients of the same group, time from the hemiplegic onset ranges from 2-3 weeks to several years, and more important, no authors specify the kind of hemiplegic evolution. Only Cazalet al. (1976) gave notice about this but without taking any correlation with the electrophysiological data. Zalis and al. (1976), too, gave an indirect sign of tonic evolution pointing out a lower presence of fibrillation potentials in hemiplegic with a beginning of voluntary activity than in patients without any voluntary activity (Severe Hypotonia). The aim of this work was to study, using conventional techniques, the presence of peripheral disease in patients with two different tonic evolutions following acute CVD.

METHOD

Subjects. The study involved 20 patients, male and female, aged from 60 to 72 (mean value: 64), who had been rendered hemiparetic after an acute CVD, timing from the onset two to six months. Patients were then divided in two groups of ten each according to the tonic evolution (Goriani, 1972): I° Group - Hypotonic evolution; II° Group - Hypertonic evolution. Patients with known associated diseases which might cause a neuropathy such as diabetes mellitus, chronic alcoholism and renal failure, were not included.

Electrophysiological study was carried out on deltoid muscle and on axillary nerve of every patient on both side (healthy for control). Recording techniques and data evaluation were used according to the methods developed in the Center of Clinical Neurophysiology of Pavia University (Buolthal and Pinei, 1971, 1973; Buchthal 1977; Cazale, 1984; Arico, 1982). Axillary nerve and its supplied deltoid muscle were selected for their anatomical peculiarities.

RESULTS
Statistical analysis. We computed statistical analysis of the distal latencies of the axillary nerve. The mean values are shown with their standard errors. The significance of difference between two means was estimated by Student's *t*-test. Within the population of the first group a mean value of 6.18±1.16 (hypotonic side) against 4.23±0.56 (healthy side) with *t*-test=-7.71 and p<0.0005 was carried out. We had also data as much significant as the previous in the hypotonic patients (II* Group) with a mean value of 4.97±0.5 (Plegic side) against mean values of 4.26±0.45 (Healthy side) with *t*-test=5.31 and p<0.0006. Student "t"-test was also calculated between the mean values of the affected sides of both groups with "t"-test=2.19 and p<0.05. Significant differences between the two healthy sides were not found "t"-test=0.5 and p>0.3.

**DISCUSSION**

In both groups, in the impaired sides, there is presence of signs of motoneuron syringing but more pronounced in the hypotonic group. This fact is emphasized by the evidence of fibrillation spontaneous activity in 70% of the hypotonic patients against 50% of the hypotonic ones. No evidence of controllateral healthy side of both groups was noted. The presence of pathologic data on the affected sides of both groups is according to data and theories present in literature as trophic factors, alteration in blood supply, transsynaptic changes. But these theories did not explain why alterations are more evident in the hypotonic evolution and more, there is evidence of Wallerian degeneration in some of such cases. A possibility is that drop shoulder in the hypotonic evolution might cause a stretch of the axillary nerve (Cassale et al. 1982). So EMG screening after acite CVT could be of great importance in order to evaluate and prevent this kind of complications, with early application of orthosis and appropriate choice of PPT strategy.

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Some peripheral nerves contain more sympathetic fibres than others. This is particularly so for those nerves which supply the hand: the median and to a lesser degree the ulnar nerve.

After section of a peripheral nerve the vessels in the denervated area are dilated and the correspondent part of the skin becomes warmer because of vasodilator paralysis. "Warmer phase" is lasting for some weeks. In incomplete irritable lesions "warm phase" can be prolonged for some months and may be combined with causalgia. Later, "warm phase" is replaced by a delayed or "cold phase" in which the affected part becomes abnormally susceptible to cold. Affected tissues are usually colder than normal ones. The sudomotor fibres are also involved, sweating is reduced and skin becomes dry.

In order to record loss of vasmotor function the temperature was measured by DSA Biological thermometer with skin and intramuscular thermoelectors.

In order to record the function of sudomotor sympathetic fibres the skin resistance was measured. This parameter is significantly increased after denervation. The measuring instrument Bio-In-Meter and recording devices were constructed by Ing. M. Kolaj.

A total of 50 patients had skin temperature and skin resistance measurements.

6 patients had median nerve, 9 had ulnar nerve and 4 patients had plexus brachialis lesion. Two patients had reflexory sympathetic dystrophy. Skin temperature and skin resistance values were compared with contralateral normal ones and correlated with EMG, neurographic and clinical findings.

I - Patients with median nerve lesion

In affected hand on the thenar, skin and muscle temperature is lower for 1.5-2.5°C than on contralateral normal side. Only one patient had 2.7°C more. The skin resistance is higher on affected side. In case with motor fibres damage with nerve potential preserved, differences in temperature and skin resistance were not significant.

Representative case: Four months ago trauma in fossa cubiti, now paresthesia in thenar and first three fingers. Skin of thenar is livid, cold and dry. EMG: nerve injury in m. opponens pollicis, nerve potential absent. Skin temperature of thenar and intramuscular temperature in m. opponens is for 2°C lower than on normal contralateral side. Skin resistance for 20 kΩ higher.

II - Patients with ulnar nerve lesion

The findings of temperature and skin resistance are different in ulnar...
II. V. Radić et al. Cutaneous Temperature and Skin Resistance in Differential Diagnosis.

Case 1: Three months ago third finger lesions. Nowday paraesthesia and pain at hypotenar, extending to axilla. Skin of hypotenar is cold and wet. EMG and nerve conduction normal. Skin temperature of thenar for 1,5°C higher, of hypotenar for 0,5°C higher on affected side. Temperature of m. abductor digiti V for 2°C increased. Skin resistance over the hypotenar for 9 kΩ lower, on thenar for 17 kΩ lower, than on opposite side.

Case 2: Cubital tunnel syndrome, operated twice, second time with post-operative supputation. The whole right hand oedematous, hyperesthetic especially on its ulnar half. The skin of hand cold, thin and dry. EMG: abductor digiti V slight chronic nerve lesion signs. Skin temperature on thenar for 2,5°C, on hypotenar for 1,5°C decreased against the healthy side. Skin resistance in thenar and hypotenar for 20-30 kΩ higher than on the healthy side.

Conclusion

In median nerve lesion, sympathetic affection signs and symptoms are much more developed than in ulnar nerve lesion. The "warm phase" is lasting in the second longer, especially if partial nerve lesion is present. In median nerve lesion the sympathetic dysfunction signs are more pronounced if additional sensory fibres lesion is found. In sympathetic dystrophy nerve potentials and conduction velocities can be normal and sympathetic fibres affection well pronounced. In low temperature cases the prolonged bleeding was observed during the EMG analysis.

III. Patients with plexus brachialis lesions

Case: Weakness of hand after Ca manmoe operation and radiotherapy. EMG: complete denervations in hand muscles. Median nerve potential very small, ulnar nerve potential absent. Skin temperature of thenar for 3,2°C lower on affected side, hypotenar temperature for 2,6°C lower. Skin resistance for 16 kΩ higher. Skin of the hand was dry, livid and atrophic.

IV. Reflex sympathetic dystrophy

Cutaneous temperature and skin resistance measurement are useful in differential diagnosis of organic causality against psychogenic symptoms.

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EVOKED ELECTROMYOGRAPHIC OBSERVATION OF INTERFASCICULAR NERVE GRAFTING

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Specific object of study

To elucidate nerve, especially motor nerve, regeneration and nerve recovery after interfascicular grafting, 1 have performed evoked electromyographic studies in clinical cases and in experimental interfascicular nerve grafting using rabbit nerves.

Methods

As a test of motor function, electromyographic examination is a useful method, but evoked motor response studies (N wave) yield more useful information on the regeneration of the motor nerves. The M waves picked up by concentric bipolar needle electrodes enable more precise wave analysis.

Three infants who received interfascicular nerve grafting of the sural nerve several months after the initial injuries were the clinical cases for this study. After the operation clinical and electromyographical observations on the regenerating nerve were carried out. The motor conduction velocity (MCV) was measured using the wave pattern of the M wave picked up from the coaxial bipolar needle electrode. Evoked sensory nerve potentials and somatosensory potentials were also recorded.

In experimental study, the right peripheral nerves of 40 adult rabbits were exposed aseptically under anesthesia, a section of peripheral nerve 1.0 cm in length was excised from the popliteal area, and to this defect 4 strands of the sural nerve were grafted under an operating microscope while avoiding any tension.

At the intervals of 4, 8, 12 and 24 months after operation, the peripheral nerve was again exposed, and by applying bipolar electrodes at two points 4-5 cm apart on the regenerating nerves below the nerve graft portion, a supramaximal electrical stimulus was given, and with the concentric bipolar needle electrode, the M wave was picked up from the anterior tibial muscle.

The peripheral nerves were taken out, and stained with nyelin stain. The axon site, the thickness of myelin as well as the number of regenerating nerve fibers were compared in the graft site and distal portion of the peripheral nerve.

Results

Clinical cases

By the sixth postoperative month some initial recovery of muscle function was observed in all three cases. Muscle power in postoperative year 5 ranged from 4 to 5 in each case. By electromyography with maximum voluntary contraction, individual potentials of motor units are observed. In one and one-half years after grafting, N waves show mostly a polyphasic tendency with an MCV of less than 30 m/sec. In the case where a single polyphasic wave and a complex polyphasic wave exist simultaneously, the complex polyphasic wave with low amplitude has a longer latency and a slower conduction velocity.

By postoperative year 5 the averaged MCV was approximately 40 m/sec which was a little less than 70% of the normal side. The MCV in the simple waves of high amplitude is higher than that in the low amplitude or polyphasic waves. The MCV of the nerve graft portion including the nerve suture sites was slower than that of the distal nerve segment.

Sensory nerve conduction velocity (SCV) at 5 postoperative years was 65% of the SCV on the normal side.

Somatosensory evoked potentials (SEP) at 5 postoperative years showed clearly each component, but there was a delay in the latency.

--- Experimental study

Wave pattern of the M wave: By the fourth postoperative month the wave
Y. Yamashiro Evoked Electromyographic Observation of Interfascicular Nerve Grafting

pattern was mostly polyphasic and in general of a low amplitude, and distal motor latency was prolonged. The often appeared low amplitude waves of short duration of the so-called nascent units. By 3 postoperative months, numerous polyphasic waves appeared, and by 12 postoperative months the polyphasic tendency had decreased and waves of high amplitude with 3-5 phases appeared. By 24 months after operation, simple waves of 3 phases could be observed more frequently.

MCV: The MCV below the nerve graft site in 12 cases, 4 months after operation, was 28.1 ± 13.8 m/sec in average. By 12 months the average MCV in the 12 cases increased to 44.3 ± 14.4 m/sec. The MCV of 8 cases at 24 postoperative months was 51.8 ± 4.1 m/sec in average. The MCV in the simple waves of high amplitude is higher than that of the low amplitude or polyphasic waves. The average MCV of normal peroneal nerves measured by this method was 74.1 ± 44.5 m/sec.

When polyphasic waves of high amplitude and low amplitude exist simultaneously, the high amplitude wave has a higher conduction velocity and a shorter latency. The threshold of these M waves was calculated by changing the stimulus voltage. The conduction velocity amplitude wave has a lower threshold level, indicating that this wave seems to belong to large and more mature nerve fibers. By 4 postoperative months myelinated fibers in the distal peroneal site can be observed, but the diameter of these fibers is less than several microns or myelin sheath is also thin. Histological findings of the graft site are similar to the above findings, but the nerve fibers are somewhat finer than the fibers of the distal peroneal site, perhaps because of the grafted nerves are sensory nerves which are more slender than motor nerves. Nonetheless, the myelination is fair.

At the proximal and distal suture sites of the graft, there are axons 3-4 mm in length, that have migrated and strayed into interfascicular tissue. Such strayed axons are more numerous at the distal suture site.

Conclusion

The M wave recorded in this manner is of short duration, low amplitude, and a so-called nascent unit like wave in the initial stage of nerve regeneration. With recovery it undergoes change to a polyphasic wave, and then to a still simpler polyphasic wave, and when the nerve matures to a certain extent, a simpler wave of high amplitude, similar to normal waves appears.

The MCV of high amplitude waves is faster than that of polyphasic waves and so regenerating fibers showing high amplitude waves would seem to be more mature nerve fibers. Recovery of MCV after nerve grafting was 65-70% of normal.

In the present study, the number of large fibers in the normal peroneal nerve is 35.9% and the percentage of large fibers in the graft site at 24 postoperative months is 24.6%, which corresponds to 69.6% of normal.

The MCV of the graft site was delayed. This suggests that the regenerating nerve fibers are smaller in diameter and that the maturation in the graft site and suture line is restricted. The diameter of the fibers in the graft site is less than that of the fibers in the distal portion of peroneal nerve, indicating less maturation in the graft site.

I = \frac{X_2 - X_1}{D}

Where: I = Amount of recovery;
X_2 = Final global score;
X_1 = Initial global score;
D = Days of treatment.

The value of (I) was related to and compared with the ENG data to obtain useful information on the maturation attained during a certain span on time. The time was quantified in terms of number of daily therapeutic sessions. The mean score thus obtained can be used to assess neural recovery attained in the follow-up of various pathologic condition affecting the NCS.

In this study, on the ground clinical and ENG data, the Authors have tried to develop a score of amelioration. This value is meant to help the clinician in the organization of the rehabilitation training in various working RMS lesions.

The patients were examined at the beginning and at the end of the pharmacological treatment and evaluated both clinically and with the aid of ENG testing. In the clinical examination the parameters of strength, sensitivity and functionally wave issued a score. This was given setting score by the gravity of the affected muscular segment. The global numeric score resulted from the total of the partial values. The initial and final global score were statistically processed, to assess the daily recovery rate. This parameter was obtained with the following formula:

I = \frac{X_2 - X_1}{D}

Where: I = Amount of recovery;
X_2 = Final global score;
X_1 = Initial global score;
D = Days of treatment.

The value of (I) was related to and compared with the ENG data to obtain useful information on the amelioration attained during a certain span on time. The time was quantified in terms of number of daily therapeutic sessions. The mean score thus obtained can be used to assess neural recovery attained in the follow-up of various pathologic condition affecting the NCS.

For superficial peroneal nerve the initial position was that on peroneus profundus, typical for evoking extensor digitorum brevis direct (80) muscle answer. 4 cm proximally to the middle of the bimalleolar line. The stimulation electrode was moved many times to obtain the potential with lowest stimulation intensities. The distal needle of bipolar electrodes (not isolated) was than placed laterally at 2 cm interelectrode distance as indifferent electrode, and nerve stimulated.

As tibial nerve recording point, the distal stimulation point for evoked fleasor hallucis brevis muscle M-potential was used without position change. The not isolated bipolar needle electrodes were left in a position along the nerve.

Above quoted placements of recording electrodes were declared optimal after series of systematic needle rearrangements along the nerves or perpendicular to it, with indifferent electrode placed on either nerve side.

The sural nerve neurography was done with placement of stimulating strip electrodes on the lateral surface of the dorsum of the foot more posteriorly than Behse and Buchthal (1971) did. The recording points was the same, only with both electrodes along the nerve at the height of the root of lateral maleolus, nearer to the tendo Achilles.

The stimulation and recording points were controlled by biological thermometer DISA and actual skin temperature recorded. The room temperature was 22-26 °C.

For superficial peroneal the patient was lying supine with knee extended. For tibial nerve neurography the patients position was prone with the foot bent over the end of the examination desk. In sural nerve analysis the patient is lying halfway on the side opposite to the analysed, with the knees slightly bent, more on the side of analyses.

The nerve potentials parameters were analysed in the following way. The latency was read by the first deflection from the isoelectric line. The duration was determined from this very first deflection until the restoration of the isoelectric line. The amplitude was determined from "peak to peak" for biphasic or polyphasic potentials. The number and direction of potential phases was described also. The millimeter distance between the stimulation and recording side was read from the middle of the space in between the two strip electrodes to distal recording electrode.

By the way the normal peroneus profundus terminal latency quotient (distance from the distal stimulating electrode to recording needle electrode point, through latency) was determined and M potential amplitude and shape variations as well.

The machine used were DISA, three channels and digital averager DISA.
### Table 1
**SUPERFICIAL PERONEAL NERVE conduction velocities and nerve potentials features**

<table>
<thead>
<tr>
<th>No</th>
<th>Mean</th>
<th>SD</th>
<th>CV</th>
</tr>
</thead>
<tbody>
<tr>
<td>36</td>
<td>41.06 m/s</td>
<td>7.18 m/s</td>
<td>17.49%</td>
</tr>
<tr>
<td></td>
<td>3.28 μV</td>
<td>1.72 μV</td>
<td>52.44%</td>
</tr>
<tr>
<td></td>
<td>0.60 ms</td>
<td>0.00 ms</td>
<td>40.00%</td>
</tr>
<tr>
<td></td>
<td>29.55°C</td>
<td>1.28°C</td>
<td>4.33%</td>
</tr>
<tr>
<td></td>
<td>30.05°C</td>
<td>1.50°C</td>
<td>4.98%</td>
</tr>
</tbody>
</table>

### Table 2
**TIBIAL NERVE conduction velocities and nerve potentials features**

<table>
<thead>
<tr>
<th>No</th>
<th>Mean</th>
<th>SD</th>
<th>CV</th>
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</thead>
<tbody>
<tr>
<td>36</td>
<td>43.28 m/s</td>
<td>7.26 m/s</td>
<td>16.78%</td>
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<tr>
<td></td>
<td>4.55 μV</td>
<td>4.14 μV</td>
<td>90.98%</td>
</tr>
<tr>
<td></td>
<td>1.53 ms</td>
<td>0.49 ms</td>
<td>32.03%</td>
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<tr>
<td></td>
<td>28.33°C</td>
<td>1.23°C</td>
<td>4.36%</td>
</tr>
<tr>
<td></td>
<td>29.86°C</td>
<td>0.99°C</td>
<td>3.13%</td>
</tr>
</tbody>
</table>

### Table 3
**SURAL NERVE conduction velocities and nerve potentials features**

<table>
<thead>
<tr>
<th>No</th>
<th>Mean</th>
<th>SD</th>
<th>CV</th>
</tr>
</thead>
<tbody>
<tr>
<td>30</td>
<td>40.62 m/s</td>
<td>6.15 m/s</td>
<td>15.15%</td>
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<tr>
<td></td>
<td>2.83 μV</td>
<td>1.24 μV</td>
<td>43.83%</td>
</tr>
<tr>
<td></td>
<td>0.4 ms</td>
<td>0.0 ms</td>
<td>42.24%</td>
</tr>
<tr>
<td></td>
<td>29.17°C</td>
<td>1.13°C</td>
<td>3.88%</td>
</tr>
<tr>
<td></td>
<td>29.37°C</td>
<td>2.06°C</td>
<td>7.02%</td>
</tr>
</tbody>
</table>

### Table 4
**PERONEAL PROFUNDUS terminal latency quotient**

<table>
<thead>
<tr>
<th>No</th>
<th>Mean</th>
<th>SD</th>
<th>CV</th>
</tr>
</thead>
<tbody>
<tr>
<td>41</td>
<td>1.58</td>
<td>0.12</td>
<td>20.25%</td>
</tr>
</tbody>
</table>

---

A study was conducted to assess the changes in the median frequency of the myoelectric signal in the quadriceps and biceps brachii of children affected with Duchenne or Becker muscular dystrophy. Each child was motivated to perform and maintain his possible maximal isometric contraction of the required muscle using the Cybex II dynamometer. Simultaneously, EMG signals of the contracted muscle were recorded and subsequently analyzed using a muscle fatigue monitor which calculates and tracks the median frequency of the myoelectric signal.

Preliminary results show that the median frequency of the myoelectric signal did not demonstrate the consistent monotonic reduction during a sustained isometric contraction in the muscle as reported previously in normal subjects.
The variety of names given to the syndrome in question reflects the disagreement about the nature of the abnormal muscular activity, its cause, and its pathophysiology. Our report aims to present some data which might bring some light into the problem.

**CASE HISTORY**

48 years old man, handworker. Three years before admission increasing stiffness of hands and feet with generalized twichings appeared.

**Neurological findings.** Generalized fasciculations and myokymias over trunk and extremities sparing the head. The permanent muscular contraction with consecutive stiffness involved hands, forearms, feet and calves with dorsiflexion contracture of the toes, and hyperextension and subluxation in interphalangeal thumb joints. No percussion myotonia. The moderate reduction of muscle strength in arms and legs and mild generalized muscle wasting were found also. Muscle stretch reflexes, Babinski, triceps surae, present, with quadriceps femoris reflex even lively. Plantar response absent, extensor as well. Cutaneous sensibility and vibration sense decreased on the toes and fingers.

**ELECTROPHYSIOLOGICAL ANALYSES**

**Electromyographical findings.** In all muscles analysed (opponens pollicis, abductor dig. min., extensor dig. communis, deltoïd, quadriceps femoris, anterior tibial, gastrocnemius, flexor hallucis muscles) during the rest spontaneous sequences of potentials with high unloading frequency. Most of them were medium or small motor units potentials. Wide, polyphasic or high potentials were exceptional. The unloading frequency was most often of so called "bizarre high frequency discharges" or "pseudomyotonic bursts". Now and then short bursts appeared made of "multiplets". Volitional activity recruited potentials until the full interference pattern, the potentials being in average smaller and more polyphasic than those typical normal for the muscle analysed.

**Ischemic ENG test** (done with the cuff on the upper arm and suprasystolic pressure of 19.5 kPa) during the first minute of ischemia already the full interference pattern of motor unit potentials was recruited. The dense discharges lasted until the third minute getting afterwards always more sparse. During the fifth minute the single potentials unloading was recorded only. After ten minutes the ischemic compression was released. During the second half of the first minute of the normal circulation already, a sudden recruitment of interference pattern with single potentials of extremely high unloading frequency was recorded. The frequent potentials had the amplitude of 2 mV, the others were much smaller and more polyphasic. After seven minutes many potentials dropped out.

**Hyperventilation ENG test**. During the first minute already, on the background of small polyphasic potentials the high frequency potentials of 2 mV appeared. They appeared, disappeared and reappeared for ten or twenty seconds. Three minutes after the hyperventilation was stopped, new bursts reappeared. During ischaemia and hyperventilation the thumb got into the strong opposition and adduction, meanwhile the other fingers were abducted and extended strongly.

**Neurographic results.** N. peroneus profundus: 42,3 m/s. Terminal latency quotient 1,20 (norm. 1,6 ± 0,3). M-potential: more polyphasic; P potential covered by numerous repetitive discharges. N. peroneus superficialis: no nerve potential. Tibial nerve: 35 m/s. In gastrocnemius and in flexor hallucis brevis muscle secondary potential of P potential type. Sural nerve: no nerve potential. Median nerve: terminal latency quotient 1,2 (norm. 1,9 ± 0,3). M-potential in his first part 8 mV, biphasic; its second part was extremely polyphasic and wide with additional repetitive activity. Sensory conduction (root of the II finger to wrist) 38,2 m/s, nerve action potential amplitude 20 µV.

The nerve conduction block results. Median nerve over the elbow was infiltrated with 8 cm of 2% Xylocain. The canule was used as different stimulating electrode and displaced carefully in order to obtain the M-potential with an low stimulus intensity as possible. Inspite of extreme M-potential reduction, and deep hyposthesia no reduction of spontaneous activity was observed. With the nerve stimulation at the conduction block area new potentials series and single multiplets appeared.

Median nerve in the wrist on the left and on the right hand side were infiltrated with 4 cm of 2% Xylocain. The results were identical both sides. Total repetitive activity joining the M-potential disappeared first. M-potential disappeared with thumb opposition activity. Direct muscle infiltration (gastrocnemius muscle) in the myokinia region resulted with only slight reduction of spontaneous activity, reduction which could be due also to the change of distance between the recording electrode point and electric source, caused by infiltration.

**OTHER LABORATORY FINDINGS**

Computerized cerebral tomography: diffuse atrophic changes. Cerebrospinal fluid: total proteins (Kieder) and gamma globulins percentage increased. IgG increased. Phosphates in urine increased (4 times normal values). Calcium in urine normal. Calcium, phosphates, and alkaline phosphates in serum normal. Parathormone normal.

**DISCUSSION**

In our case besides the signs and symptoms overtly due to peripheral nerve...
Should our case represent a combination of two syndromes stiff-man and Isaacs-Mertens syndrome?

Our electromyographical and electro-neurographic findings are very similar to those recorded in our three cases of untreated tetanus parathyreoidralia Mertens and Ischcke 1965, already pointed to electromyographical similarity with spasmodilia.

In hypoparathyreoidismus cuff ischemia and hyperventilation provoke the abundant discharging of multiplets. We had the opportunity to observe in four such cases spontaneous generalised appearance of multiplets and serial repetitive frequency discharges with other electromyographical laboratory and clinical signs and symptoms pointing to neuromuscular disease not described as yet. The representative case should be described with more details.

CASE REPORT

1. V. Dž., 50 years old, male worker. Anamnesis: Since the age of thirty sight difficulties, with 14 years cataract was operated on the left side and with 39 on the other.

Some years ago he started to have cramps in the hands and feet. During the last month before the admission he walked around with always more trouble, feeling unsteady and dizzy. He was admitted because of the epileptic, grand mal fit.

Neurological findings: The general appearance was of skinema and bradikinesia with unstable gait and with some resistance on passive movements pointing to muscle hyper-tonia. There was no possibility to elicit the propriosensitive reflex movements by usual tendon tap but only by direct percussion of the muscle bulk. The same phenomenon was observed even with masseter reflex. The patient complained of fasciculations, cramps and paraesthesia.

Electrographical finding (before starting the Calciferol treatment): Quadriiceps femoris muscle: during relaxation abundant multiplets and serial high frequency discharges of motor unit type potentials. On voluntary contraction sudden recruitment of almost interference pattern made of extremely polyphasic potentials of 500 to 1,000 μV. Extensor digitorum brevis muscle both sides: During the relaxation abundant multiplets and serial high frequency discharges of motor unit type potentials. On voluntary contraction good intermediate pattern with some more polyphasic potentials of 2 mV. Anterior tibial muscle: the same finding as in extensor digitorum brevis. Opponens and adductor pollicis muscles: during relaxation abundant multiplets. On voluntary activation good intermediate pattern of low voltage, polyphasic potentials.

Neuographic results: Median nerve: motor conduction 50 m/s distal latency quotient 1,2 (5,8 cm/4,6 ms). Sensory conduction (1 finger - wrist) 38,0 m/s. N. peroneus profundus; 55 m/s. Distal latency quotient 1,3 (8,5 cm/6,3 ms). N. peroneus superficialis: no nerve potential elicitable.
Ischemic electromyographic test: immediately after stopping ischemia, abundant multiplets and triplets appear discharging for more than 11 minutes. Some serial high frequency discharges of fibrillatory type appear also.

Hyperventilation electromyographic test: Two minutes after stopping hyperventilation abundant multiplets and triplets discharging for five minutes.

Electromyoneurographic conclusion:
Most prominent feature is (on upper and lower limbs) generalized spontaneous appearance of multiplets and high frequency serial discharges, the quantity of which increase on ischemia and hyperventilation.

In quadriiceps muscle electromyographic findings points to myopathy.

Significantly increased action potential polyphasia, distal latency quotient reduction, some slowing of sensory conduction velocities on the hand and no nerve potential elicitable on the foot indicate some distal motor and sensory nerve involvement also.

On reexamination after a week of Calciferol treatment spontaneous activity was significantly reduced.

Relevant laboratory findings:

CT of the brain: At 2A and 2B abundant califications projecting into the capsul nuclei caudati. Abundant califications projecting into the posterior thalamus nuclei. Abundant paraventricular califications also. Even at both cerebellar hemispheres significant region of increased absorption coefficient.

Cerebrospinal fluid: Cells 28/3, Ly 10, polymorphonuclear leucocytes 3, proteins 0.45 g/l, sugar 4.05 mmol/l, Parathyroid hormone (PTH) 2.5 mol/l. Calcium in serum: 0.45-1.15 mmol/l, Calcium at urine: 0.2-0.64 mmol/l, Phosphates at serum: 1.94-2.90 mmol/l, Phosphates at urine: 3.30-11.6 mmol/l, Alkaline phosphatase: 51-119 U/l, Magnesium at serum: 0.3-0.5.

In part I it is described how surface EMG can be analyzed in a statistical way. In this paper the application of the procedure in a clinical situation (a rehabilitation centre) will be discussed. First it will be shown that the EMG registrations can reveal characteristic patterns with certain kinds of disorders.

Secondly it will be shown that the spectrum, derived from the time registration, can show marked changes with specific disorders.

Finally it will be shown that surface EMG can be used, during the rehabilitation process, as a tool for quantification.

**METHOD**

The equipment used for these investigations is described in part I. The subject is first asked to exert maximum force three times. Hereafter EMG registrations are made with four different forces in the range of 20-90% of the mean maximum force (for instance 20-30-50-90% force). With every next investigation these same forces are used. At every investigation also the maximum force is determined but after the EMG registrations in order to avoid fatigue.

**RESULTS**

Changes in the time registrations

Especially with disorders that result in a relative large loss of motor units deviations of the normal pattern are found. In Figure 1a a registration is shown of a subject with polynymetias. The individual action potentials are clearly visible.

In Figure 1b a single motor unit pattern is shown of a subject with a plexus lenticulitis. These kind of registrations cause alterations in the amplitude distribution function (usually Gaussian). Especially changes are found resulting in a high value of the kurtosis (between 2 and 5, normal value is 1) and a different value of the kurtosis. Furthermore the number of peaks per second decreases, even to frequencies that coincide with f1f2.
H. Hermens et al Use of Surface EMG to Evaluate Process of Rehabilitation

Changes in the parameters during a follow-up investigation

A number of hemiplegic subjects who were treated in our centre participated in a follow-up study; once a week an EMG investigation was made. A summary of the results is given below:

- The relative spread in the exerted forces; the value of this parameter usually decreases if an improvement of motoric function is achieved (e.g., from 65 to 25).
- The parameter RTG; the value of this parameter is often significantly increased. An example of the course of this parameter is given in Figure 3.

![Graph showing changes in parameters](image)

Figure 3. The EMG Flight Correlation and the Maximum Force vs. Time of a hemiplegic subject.

- The parameter RIC; this parameter can show marked changes with hemiplegic subjects especially during the first weeks after the stroke. An example of the course is shown in Figure 4. In this case the value of this parameter increased to 11. This means that the subject was unable to exert a force without contracting his antagonist in a similar level.

![Graph showing changes in parameters](image)

Figure 4. The reciprocal innervation coefficient vs. Time with a hemiplegic subject.

- The parameter effect; in contrast to healthy subjects this parameter is rather irreproducible with hemiplegic subjects and there is also no obvious relation with the exerted force.

Discussion

From our study it appears that surface EMG can be used as a tool for quantification. The RIC is interesting to quantify the amount of dysbalance between two muscles. Therefore deviations can be expected with disorders like hemiplegia (Vissel and Zilvold, 1978; Feys, 1979). The RTG is useful to quantify the 'efficiency' of a muscle. Lang (1981) found in half of his group of hemiparetic subjects an increase of the slope between exerted forces and integrated EMG which is according to our findings. Also with peripheral nerve lesions an increased value of this parameter can be found. Surface EMG is especially suited as a tool for quantification in rehabilitation centers. An example of the application of surface EMG is the quantitative evaluation of phenol block (Zilvold et al., in this proceedings).

Additional advantages of the use of surface EMG are its noninvasive character and the possibility to delegate the investigation to e.g., physical therapists.

References

INTRODUCTION

The quantitative estimation of the maximal voluntary muscle contractions is one of the basic elements in the diagnostic and follow up of neurogenic lesions in routine electromyography. The normal finding is a full interference pattern. In neurogenic affections, we have a rarified pattern due to a smaller number of activated motor units. The reduction or absence of interference pattern is proportional to the degree of the impairment of the neurogenic affection. The estimation of the reduction of the interference pattern is usually subjectively done by the myographer himself in the routine clinical electromyography. The reduction is usually quantified in the measurement of the full interference pattern, or by description as: full interferential trace, reduced interferential trace, intermediate etc.

The computer analysis has already been introduced in clinical practice. In last 15 years qualitative and quantitative analyses of the electromyography findings were applied mainly in investigations. The digital frequencies analysis /1/, pattern recognition /2/, phasic electromyography /3/ or so-called digital filtering method could be used just if the computer is on-line connected. The use of the computer analysis is expensive, complicated, (especially in our conditions) for routine clinical electromyography. Unsatisfactory accuracy of the subjective estimation of the maximal voluntary contractions, especially in the follow-up of neurogenic affections brought us to the idea to develop a simple measuring device for quantitative objective estimation.

Two physical magnitudes of the electromyography signal were available: sound and video signal. We decided to use the video signal after experience and tests. Two reasons were dominant: the possibility of the analysis of the signal and accuracy of the quantitative discrimination.

METHOD AND EQUIPMENT

From the viewpoint of view the electromyography signal is a scene obtained as the sum of the motor unit action potentials. The single scene analysis method is convenient for the objective quantitative electromyography analysis /4/. The luminosity or brightness could be chosen as the classifier. The photo detector has to be the feature extractor.

The classification in several categories is of interest, as it will be discussed. Usually the main purpose of the scene analysis is the determination or recognition of the picture via computer. The purpose of the suggested method is to recognize the integral gray level of the oscilloscope screen. The used instrumentation (Fig. 1) consists of the concentric needle electrode, preamplifier, 565 dual beam Tektronix oscilloscope, optical adaptation system, sensor, photo cell, amplifier and LED display. The optical adapter is constructed in such a way that it is compatible with different oscilloscope screens. The whole screen is in "view" of the photo cell.

The choice of the photo transducer is made after tests comparing the results obtained by computer analysis. The sensor transducer gives the best discrimination of the brightness. The output voltage obtained from the photo cell has to be analyzed. The inertia of the registration system is of interest because of fast changes of the gray level. The application of the analog or digital display was not necessary because the classification is in the categories, i.e. the use of LED display is possible. The five light emitting diodes, connected in the line, show in previously adopted levels of motor unit activities.

The five assumed quantitative categories are 0%, 50%, 60%, 75% and 100% of the normal maximal voluntary contractions. The 0% and 100% level, have to be calibrated according to the zero and full activity of the corresponding non-affected muscle of the same subject. That is to be done by simple optical and trimmer fittings. Two effects of nonlinearity must be pointed out. One of these is due to instrumentation and the other one affects interference and nonlinear rise of activity in comparison with the brightness of the screen.

The testing of the system was carried out on six patients with the verified unilateral lesions of the peripheral nervous system. The quantitative deficit was tested:

1. Electromyography, by visual estimation (subjective).
2. Electromyography, by computer (frequencies analysis).
3. Electromyography, by photo detection of brightness.
4. Intensity/time curve, by all six patients.

DISCUSSION AND RESULTS

The synthesis of the simple, cheap device for quantitative analysis of the electromyography signal was the basic goal of our work. The comparative analysis of the above mentioned tests showed that the photo detecting system is simple for application and maintenance, adaptive to the different oscilloscope screens, and objective. Comparing the quantitative results obtained by computer analysis and photo detection of the system the difference rise up to 10%. Bigger difference exists in comparison of the subjective estimation and photo detection, especially by smaller reduction of interference patterns. The use of photo detecting device is instantaneous because the fitting is quick, adjusting of zero and full level is simple and the result instant. The whole procedure takes less than five minutes, which is important for routine clinical work.

The application is possible for unilateral neurogenic diseases (lesions of central and peripheral nervous system). We also think that this system could be applied with adjustments in some kinesiologic investigations.

The additional clinical evaluation of the suggested system is still necessary to make better the classification of the reduction.

CONCLUSION

The possibility of quantitative analysis of neurogenic affections in routine clinical electromyography photodetecting device is approved and described in this paper.

REFERENCES

The effectiveness of the electrotherapy of the denervated muscle in reinnervation processes has given rise to perplexity, so far. Actually, from the clinical point of view, conclusions often turn out to be contradictory (Osborne, 1951; Pinelli et al., 1979a). The results, which were the same both in treated animals and in controls, were obtained experimentally on the rabbit's Tibialis anterior muscle by electrostimulation with square wave 60 msec duration, 1/sec, twice a day, each time for 5 min; for 90 days after surgical section and subsequent injury of the sciatic nerve (Pinelli et al., 1979b; Moglia et al., 1983).

Our aim was to tackle the problem again, always experimentally on rabbits, by increasing both the number and duration of the daily electrostimulation sessions and comparing the results not only with those obtained in controls, but also in animals treated with kinesthesiotherapy techniques.

Materials and methods
Fifteen New Zealand 90-100 days old rabbits, weighing between 2.4-2.7 kg, were used. The sciatic nerve section and the subsequent peripheral neurophysiology were performed according to previously described techniques (Zamulino et al., 1978).

Five of the animals did not receive any treatment after the operation. In 5 animals the electrotherapy of the tibialis anterior muscle was applied with surface electrodes until the 90th day after surgery: 3 sessions lasting 15 minutes each, using triangular exponential stimuli of 300msec duration, 1/sec, in isometric conditions of the muscle. For the remaining animals very large cages were used, in order to allow maximum spontaneous motility. Furthermore, these animals underwent passive mobilization of the limb operated on, twice a day, with 20 "stretches" a session. Maximum Motor Conduction Velocity (MVCV) of the sciatic nerve was assessed in all animals before and 90 days after the surgical section of the nerve. The presence of fibrillation activity was also evaluated in the tibialis anterior muscle. The techniques were those previously described (Moglia et al., 1979).

On the 90th day all animals were sacrificed and specimens of the sciatic nerves and anterior tibialis muscles were processed for histological, enzyme-histochemical and ultrastructural studies.

Results
The mean value of basal MVC (15 animals) was 80 m/sec (SD 4); that of the maximum amplitude of muscular response (M Ampl.) was 15 mV (SD 5) when it was evoked both proximally (p) and distally (d). The mean values of MVC and M Ampl. obtained in the various groups of animals on the 90th day are shown in Table 1.

Table 1

<table>
<thead>
<tr>
<th>Group</th>
<th>MVC (m/sec)</th>
<th>M Ampl. (mV)</th>
</tr>
</thead>
<tbody>
<tr>
<td>C</td>
<td>56.39</td>
<td>10.06</td>
</tr>
<tr>
<td>P</td>
<td>4.91</td>
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<tr>
<td>D</td>
<td>2.74</td>
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<tr>
<td>SD</td>
<td>1.21</td>
<td>1.00</td>
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</tbody>
</table>

Table 1 supplies the mean values (m) and standard deviations (SD) of neuromuscular results obtained on the 90th day after surgery in control animals (C), treated with muscular electrostimulation (E) and kinesthesiotherapy (K).

The mean value of the fibrillation activity (i.e., counting from 0 to 5) that was checked on the 90th day, was 2 in the 2 animals treated with electrostimulation and in those undergoing kinesthesiotherapy. No significant differences were observed among the 3 groups of animals (test and analysis of variance) for any parameter under consideration.

No significant differences were observed in the 3 experimental groups when the histological examination was performed. Degenerative changes of the myelin sheath of 5% of large myelinated fibres and axonal degeneration were present in the proximal nerve stump. Some patterns of nerve regeneration were also found both in the distal and proximal nerve stump. They were characterized by the presence of clusters of enlarged regenerative axons with thin myelin sheaths, enclosed within Schwann cell basement membrane.

Conclusions
The objective, neurophysiological and histologic results obtained seem to exclude that reinnervation processes are "speeded up" by rehabilitation techniques such as those employed in this study. In particular, due to the electrostimulation of the denervated muscle, the results are the same as those that were observed with different kinds of stimuli and when the muscle was free to contract (Pinelli et al., 1979b; Moglia et al., 1981; Arrigo et al., 1983).

Finally, our experimental data seem to confirm what had been reported by Pinelli (1979): voluntary muscles, correlated with the valid muscle contraction seem to recover better in the denervated limb by physical activations which are easier and physiological.

References
Marletti et al. Effect of Handicaps and Sex on Cross-Section and X-Ray Density....

Introduction.

Data on cross-section and X-Ray density of human muscles can be obtained by means of isotropic equipment and may be valuable either for detecting and monitoring neuromuscular diseases or for evaluating the effectiveness of therapies [1,2]. Data on healthy subjects are obviously necessary to establish the range of normal values and the effect of side dominance, sex and age.

Materials and Methods.

In our study, 7 healthy males (mean age ranging from 25 to 30 years) and 7 healthy females (with age ranging from 20 to 40 years) were selected at random from a group of normal volunteers free of orthopaedic problems and free of activity in asymmetrical sports. All subjects had a right side dominance.

A Siemens Somatom total body scanner was used to take 2 scans at 125 kV and 250 mA. Both arms and legs were relaxed.

Area of the muscles and bones as well as area of each muscle volume (Muscles and Extremities) were computed with the Siemens E-solution System with also provided the average value and the standard deviation of the density values within the region of interest. Symmetry indices for cross-section and density were already defined as the ratio of the area to the dominant to the non-dominant side. Table 1 shows how the areas were defined.

Density values were defined according to the ISO standard: the least density area of all the muscle and bone area is defined as the lower limit of density. The difference between these 2 areas is defined as the standard deviation of the density values within the region of interest. These symmetry indices for cross-section and density were already defined as the ratio of the area to the dominant to the non-dominant side. Table 1 shows how these values were defined.

Table 1. Average and standard deviation of the area of muscle and bone area ratios. (Average + Standard Dev. Dm. %)

<table>
<thead>
<tr>
<th>Muscles</th>
<th>Females</th>
<th>Males</th>
<th>% of area</th>
</tr>
</thead>
<tbody>
<tr>
<td>Upper limb</td>
<td>0.964</td>
<td>0.995</td>
<td>1%</td>
</tr>
<tr>
<td>Lower limb</td>
<td>0.874</td>
<td>0.902</td>
<td>1%</td>
</tr>
</tbody>
</table>

Results.

When cross-section and density values were plotted against muscle volume, a cluster of points was obtained around the 1:1 point. The distribution of these points was not affected by the type of muscle, by the section of the body, or by sex; the points therefore fell on a single cluster. Under the assumption of gaussian distribution of the pooled points, statistical analysis shows that the data from 95% of the normal population lie within the ellipses of Fig. 2.

Fig. 2. Cross-section and density ratios of muscle volume (muscle/dominant limb) in subjects. The ellipses include 95% of the normal population.

Density of muscle lodes appears to be affected by section level, muscle type and sex in the lower limb and mainly by sex in the upper limb, as shown in Fig. 3. Standard deviation of density within a muscle lode is not affected by section level, side dominance, muscle type or sex and has an average value of 1.05% with a standard deviation of 3.1%.

Discussion.

The ratio of the total muscle and bone area (non-dominant limb) is slightly but significantly lower than one for the upper limb (p < 0.05), but not for the lower limb in either males or females (Table 2).

The single lode density ratios are lower than one at a regionally significant level (L5 > L6 < C6) while the area ratios are significantly lower than one for the upper limb but significantly greater than one for the lower limb (Fig. 2).

Density of the leg muscles is always significantly greater than females than males and at the distal section level lower than at the proximal one. Density of proximal femoral muscles of the foot is lower than that of the dorsal flavesce. Density values of the new lodes are markedly affected by sex, the males having lower density than the females (p < 0.05).

Only within the female group upper limb muscle density is lower at the proximal level.

While some of the results obtained were expected, the higher muscle lode density found in the non-dominant leg is not surprising and deserves further investigation and correlation to the individual characteristics. Concerning the emphasis of histological data needed to explain the variations observed between the anterior and posterior muscle groups of the leg and between the proximal and distal section levels.

Our data do not match with the observations of Israels et al. (1972) who found no significant density differences between section levels and sexes. However, the age groups considered in this study, they also found a significantly higher density in the lateral anterior with respect to the biceps surae. Our data provide a good reference point for the pathological values already available in the literature [1].

Sabbatini et al. (1971) and others researchers found significant differences in EMG spectral parameters in relation to hand dominance and sex. Such parameters are related to the motor unit action potential sizes and to the filtering transfer functions of the fibers in the muscle and the electromyogram. The latter parameter might well be influenced by the amount of fatty tissue within the muscle or by the thickness of the "perimysial adipose." Both factors may be well quantified on EMG stands and some variations of EMG spectral values may be related to them. Further research is under way to establish such relationships.

Fig. 3. Density of muscle lodes. SF: dorsal flaverce, PF: plantar flaverce, D: distal sections, P: proximal sections.

References:
3. Sabbatini M., Marletti R., Delucchi C., Rozzi D. Handicaps, sex and force level affect the median frequency of the electromyographic impulses.
Tetanus toxin is known to have effects at various sites (1,2,5,6,7,8). In experimental studies, increased central nervous system excitability has been observed (3,9). There is some evidence to implicate alpha neurons as well (10). Action at neuromuscular junction in terms of failure and blocking has also been demonstrated (5,7). Shahani, Baskur & Baskur (9) have shown that stimulation of peripheral motor and sensory axons, in the context of tetanus toxin patients with severe tetanus and coccidial tetanus, was studied during which there was evidence of tetanus, and/or span in some of the skeletal muscles, in an effort to understand the neurophysiology underlying this manifestation.

MATERIAL & METHODS: Fifteen patients with tetanus were studied between the third day and about 12 weeks after the onset of tetanus. Various skeletal muscles were stimulated: anterior biceps, biceps, abductor pollicis brevis, abductor digiti minimi, quadriceps, iliobursa major and minor, soleus and extensor digitus longus brevis, etc. were sampled. Concentric needle electrodes were introduced into the muscle and the electrophysiological activity generated was monitored on a pen recorder and photographed when necessary. The activity was observed for at least one hour (5 minutes) and changes if any, in the signal or the pattern were noted. In some patients, movement was applied to the muscle to see if there was any movement. Sometimes patients were asked to perform movements in the same direction and the opposite direction (demanding action from the antagonist muscle).

Occasionally, the nerve supplying the muscle was stimulated supramaximally, and change in spontaneous activity noted if any. Needle block with anesthetic agent was tried in few patients by an injection of procaine at the site of entry of the needle. The block was verified by proximal electrical stimulation failing to produce an EMG response in abductor pollicis brevis muscle. Spontaneous persistent activity if any was recorded from abductor pollicis brevis after the block. When the spontaneous activity was being recorded from the muscle, the patient was studied to detect any action of the affected. When the long loop reflex was demonstrated by the use of a needle, the excessive activity was extinguished. In some patients, the subject was asked to do the action of looking for presence and/or absence of long loop reflex. In some patients, when the reflex could not be elicited at rest (but with spontaneous activity present), the subject was asked to do the action of looking for oscilations (when the long loop reflex was demonstrated by the procedure of stimulating a median nerve).

RESULTS: During the presence of cholinergic medication in the patients of tetanus, invariably spontaneous activity is picked up on introduction of a concentric needle. This activity resembles single motor units producing a mixed or an interference pattern. After some interval of time, varying in different patients, the activity reduced and gradually discrete oscilations could be identified. The oscilations resembled those of single motor units. On careful study, it can be observed that these units are occurring almost at fixed time intervals, the frequencies being from 5 to 20/sec. in case of facial muscles to about 50/sec. in case of skeletal muscles.

If the muscle is subjected to mechanical stimulation, the spontaneous activity becomes more dense. In demanding voluntary action in the same muscle, it is possible to get recruit-
TOPICAL ANESTHESIA: KINEMATIC IMPROVEMENTS IN PATIENTS WITH STROKE

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Introduction

The improvement of motor control in patients with stroke after application of topical anesthetics on the skin has been demonstrated in our laboratory (Sabahi et al., 1981). In this report, the short and long term effects of this technique on active movement patterns and gait in patients with stroke will be described.

Methods

Seven patients (6 males and 1 female) with an age range of 21-83 (46 ± 16) were tested. Patients were afflicted by stroke due to either cerebral vascular accident or head trauma. All had sustained after participation in physical therapy programs. Four patients were affected with eight hemiparetic. Three exhibited mild reduction in skin sensation. Onset of ataxia had occurred 7 months to two years prior to this investigation.

Initially, time measurements of 10 repetitive movements (RMM) at the affected elbow and knee joints and gait measurements were taken in separate sessions. These measurements were made before and 45 min after application of topical anesthetics (202 benzocaine) to the affected upper and lower limbs. For the gait analysis, 3 foot switches (placed under the heel, head of the 5th metatarsal, and the 2nd metatarsal) and 2 electromyographs (placed on the lateral aspect of the ankle and knee joints) were used on each limb to record the temporal components of the walking cycle and the angular displacement of the ankle and knee joints. Other gait parameters such as angular speed (SR), temporal asymmetry index (TAI), symmetry factor (SF), and time for single limb support (SLS) (right or left) during normal walking were calculated as well.

After completion of the prelaboratory tests, patients were selected randomly to undergo a physical therapy program, and to receive either a topical anesthetics spray or a placebo spray three times a week for one month. After one month, all tests were repeated before and 45 min after application of a placebo spray. All patients received a second month of treatment identical to the first, except that the spray was changed from active to placebo or vice versa. To test the long term effect of the treatment program, the patients were tested at the end of the two-month period, without application of topical anesthetics or placebo spray.

Results

Topical anesthetics caused partial reduction in skin sensation beginning 5 min after its application. However, patients were able to feel deeper sensation of pin prick. Partial reduction in skin sensation continued throughout the test period and diminished gradually afterward. This effect was not noticed by the patient when sprayed with placebo.

The time for the completion of 10 RMM at the knee joint was measured in 6 patients. Figure 1 shows that in a typical subject, the time for 10 RMM substantially decreased 45 min after application of topical anesthetics to the skin of the lower limb (arrow). Similar reduction was recorded in 5 patients with a range of 24-72% (mean value 57 ± 24%). In one patient, the time was slightly increased.

In the elbow joint, however, only 3 patients were able to complete this task. In 2 patients, the time for 10 RMM decreased to 15.42 and 30.4 respectively, and increased in 1 patient after application of topical anesthesia to the upper limb. This was the same patient who showed an increase in the time for 10 RMM at the knee joint. Patients stated that their affected limb was "much looser" after application of topical anesthetic and that they could better control the speed, direction, and magnitude of their movements. No appreciable changes were recorded with the placebo spray.

The TAI (Shawe and Judge, 1980) compares the time duration of the swing and double support phases of the sound limb to that of the affected limb. In normal subjects, the value of the TAI is zero. Prior to the application of topical anesthesia, the value of the TAI for 7 patients tested ranged from 8.5 to 30.0 (average 16.8 ± 7.8). After application of topical anesthesia, the range of the TAI decreased to 7.0-18.5 (average 10.2 ± 6.0).
Topical anesthesia was applied to the skin of the lower limb. Correlation between the results of the past and the recent electrophysiological studies implies an overall change in the control mechanism of movement in patients with stroke after application of topical anesthesia. Desensitization of the skin by topical anesthesia could have a potential effect on the rehabilitation of patients with spasticity after stroke.

References


**METHOD**

A. The phenol blocking procedure

The phenol blocking procedure is described in a previous paper (Hermon et al., 1981).

First a bipolar surface electrode electrode is placed on the distal part of the muscle over the tendon. In practice it was found that the best position for the indifferent (stimulation) electrode is on the opposite side of the limb to the motor endplate zone. The best position for the grounding (ENG) electrode is between the motor endplate zone and the ENG electrode. This however is only possible with muscles with minimal muscle fibers (e.g., fibula). Therefore often the indifferent electrode is placed more proximally and the grounding electrode more distally. The localisation of the motor endplate zone is done in the traditional way. After this a special needle is used that serves both as a stimulation needle and as a needle to inject the phenol solution. After connecting the needle to the stimulator and a syringe that contains 5cc of the phenol solution (5% phenol in water), the needle is inserted. Now the needle is moved in order to find the point with the largest ENG response upon stimulation (50ma, 2ma). If such a point is found the stimulation current is lowered (5ma) and the needle is moved slightly around this point in order to find an optimal position. Both the amplitude of the ENG response and the shape of this ENG response are used to judge the suitability of the point. To obtain the ENG response a scope with a storage unit is used. Some examples of ENG responses are shown in figure 1. It was found that positions that cause clear biphasic responses are the most suited to inject phenol.

![Figure 1](image)

**Figure 1.** Typical example of ENG responses. The shape of the ENG response can be used to judge the position of the needle. When such an optimal point is found a little amount of phenol solution (0.2-0.3cc) is injected. The ENG response will disappear immediately but it can recover partially. In this case again a little amount of phenol solution needs to be injected. Hereafter the needle is moved in a new direction in order to find new points.

B. ENG investigation before and after the phenol blocking

In order to evaluate the effect of phenol blocking a separate ENG investigation is made one week before and two weeks after the blocking. For a pilot study we selected eight hemiparetic patients that were treated at the Movement Disorders Unit. The subject is seated in a device with which it is possible to register the isometric force during plantar and dorsiflexion of the plantar joint (figure 2).

Bipolar surface electrodes (Model, E121M) are placed on the muscle of interest and on an antagonist (Gastrocnemius, Tibialis). The subject is asked to exert several isometric forces (3 different low forces and maximum force) both with the plantar flaxion and with the dorsal flaxion. During these contractions and during passive dorsiflexion, registrations are made. The isometric force is measured with strain gauges and displayed so the patient can maintain the force on a constant level. The following parameters are recorded: the standard deviation of the exerted force (SD), the standard deviation of the ENG signal (SD), the coefficient of reciprocal stimulation (RCS) during dorsal flexion and the ENG force gradient during dorsal flexion and at plantar flexion. These parameters are
A Milvold & Perraiyen: A report of spinal dummies by means of fetal blocking. A study of these procedures.

Analysis described in the use of surface EMG, part 1 (in this proceeding).

- The H1 at dorsal flexion decreased with seven subjects: this means that the Tibialis is used in a more efficient way.
- The H1 at dorsal flexion decreased with seven subjects: this means an improved balance between thigh and calf muscles.

DISCUSSION

The method of phrenal blocking described in this paper is applied during one year at 95 patients. Compared to the 'old' method (without the use of EMG), the following remarks can be made:

First, it can be noticed that this method can be learned in much less time due to the feedback of the EMG response on the scope. Secondly, it can be noticed that also reflexes caused by stimulation of mixed nerves can be detected on the scope (as a second EMG response). This is of importance because blocking of the nerve can cause reflexes. Only two cases of neuralgia occurred within this group of 95 patients.

After the pilot investigation we can conclude that surface EMG can be used to study the effect of phrenal blocking. In the future attention will be paid to the parameters of the surface EMG and the duration of the blocking.

REFERENCES


Specific object of study:

The hand motility has been investigated in 20 a cute hemiplegic patients 2-11 months from the onset of the paralysis with a follow-up period of 3 months. Eleven patients presented a severe paresis with a loss of strength in the wrist extension of the order of 70-95%. Seven patients presented moderate or slight paresis; among these last 9 patients, 5 were improved from previous severe paresis, while 2 presented a moderate paresis from the onset with a strength loss of less than 30%.

Methods:

Clinical examination of the whole upper limb was performed with particular and specific concern with the evaluation of the extensor and flexor activities of the hand, a mechanical device was built for measuring and recording the passive (p) and active motion of the wrist: different weights were applied to a cable/pulley arrangement acting on the hand either as a stretching force which moved the hand passively or as a resistance opposing the voluntary contraction of the wrist muscles. EMG investigations were carried out at the same time on the extensor (E) and flexor (F) muscles of the wrist. All patients underwent a conventional physio-kinetic treatment in addition to it, some of them, received electrical stimulation of the E (20 Hz/sec).

Results:

1) Different kind of functional impairment in the E have been ascertained: in a few cases the loss of strength strictly corresponded to the deficiency in extensor voluntary innervation; the reflex tone of E was also decreased.
2) In the other cases a relatively high amount of E, active innervation was preserved and the decrease in voluntary strength depended upon flexor spasticity (2a) or flexor innervation (2b) or flexor reduced elasticity (2c).

The high degree of innervation is reflexly elicited in F (FpI) by the shortening of E; the FpI innervation can act as a mechanical negativeter force on E contraction and should also exert a secondary effect of antagonistic inhibition. This FpI reflex is only elicited in the isometric contraction (EEL-), with a resulting difference between the innervation of F in EEL- with respect to the isometric contraction (EEO). Fig. (1).

2b) Enhanced innervation of the antagonistic muscles without reflex spasticity. During voluntary innervation intended to selectively activate the extensor muscles (Ee) a simultaneous innervation of the flexor muscles was recorded with a corresponding reduction in the range of motion (E). The degree of flexor spasticity did not play a role in these cases. Patients with equal degree of paresis in E and with selective innervation of E, were able to develop a higher degree of motion (Tab. 1).

2c) Low degree of range motion with respect to the degree of selective innervation of E (EEL-). This EMO/reflex dissociation could be explained by the presence of increased myoarticular mechanical resistance, neither coinnervation nor flexor spasticity were present in significant degree.

II) Follow-up investigations showed the following results which deserve particular mention:
A) E, muscles: modality of recovery in voluntary innervation and force. By means of investigation repeated at intervals every few weeks we have found that E during the phase of the most severe paralysis shows very small multiple increments of innervation and strength. On the contrary, E, presenting or reaching slighter degree of paresis (with a decrease of innervation and strength less than 40% with respect to normal), shows a single much larger stepwise increase in both innervation and strength. The improvement rate becomes again relatively small at the lowest degree of paresis. The distribution of both EEO and force in the total population of patients seems to be in agreement.
A Villing Pimentel et al. A Analysis of wrist extension movements in hemiplegic with these findings. (Fig. 2)

II) F-muscles: ratio between increase in volun-

tary innervation and spasticity.

Voluntary innervation of F evaluated in isome-

tric (FL50) contraction showed an increase pro-

portional to the degree of spasticity in patients

with the moderate to highest degree of paresis.

After this point the force could further increase

while the spasticity was less marked (point of

reversal). An increase of spasticity was not

observed in E.

Conclusion:

The impairment in strength of wrist extension

(in hemiplegia) can be due to different factors:

I) lack in voluntary innervation, II) reduced

passive motion caused by myoarthropathic disor-

ders, III) antagonistic central conervation or

IV) increase in antagonistic spasticity.

A stepwise type of recovery specifically found

in E, could correspond to the reactivation of a

complex motoneuronal assembly able to induce an

"automatic" increase in extensor strength and to

diminish the flexor spasticity.

The electrical stimulation of F induced a substan-
tial improvement in E voluntary innervation, in

the few patients treated until now.

<table>
<thead>
<tr>
<th>Causes</th>
<th>Selective E Innervation</th>
<th>Conservation of E and F</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>a</td>
<td>b</td>
</tr>
<tr>
<td>PdEp</td>
<td>$\mu$V/sec</td>
<td>162.5</td>
</tr>
<tr>
<td>FpEp</td>
<td>$\mu$V/sec</td>
<td>20</td>
</tr>
<tr>
<td>Ba-Ep</td>
<td>$\mu$V/sec</td>
<td>8.1</td>
</tr>
<tr>
<td>Range (*) of motion</td>
<td>30°</td>
<td>30°</td>
</tr>
</tbody>
</table>

Tab. 1: Voluntary contraction of E, against a constant resistance (0.1kg) in 4 patients (a,b,c,d).

E Strength (Mayo Cl. scale)

Fig. 1: Difference between isotonic and isometric E innervation versus F. spasticity in 4 patients.

Fig. 2: EMG and strength of wrist extension voluntary contraction in all patients investigated (1st investigation).

INTRODUCTION

Kinetic alterations and the sensori-motor 
treatment of spasticity in hemiplegia constitute
areas of insufficient knowledge, which require
further fundamental and clinical studies. In uni-

lateral brain lesions, integral and edocable

neuromuscular balance can regain some coordi-
nate action by different therapeutic approaches.

In several stages of the spastic condition, the

movements are affected, in part, by an increase

of the muscle tone and by a prolonged response to

stimulation. The present EMG study was

performed to investigate the extent of the degree

of upper limb stability on the control of tonic

muscle activity and on the grasp and release

training. The method includes the use of the

investigator's hands and/or objects to stabilize

the spastic forearm and hand in order to treat

and clarify the pathologic neuromuscular activity.

METHOD

Four hemiplegic patients whose ages were re-

spectively 53, 58, 57, 70 years, were evaluated
twice within a three weeks autonomous program

of treatment. Three of these were classified as

being severely spastic and one was very severely

spastic. Two suffered from an aphasia of expres-

sion. Previously, they had all undergone an inten-

tive period of rehabilitation of 12 to 18 months

duration. At the time of this study, the post-

independent interval was 18, 30, 25 and 43 months.

A bipolar Beckman skin miniature electrode

was attached on the side of spasticity over the

skin of the belly of 1) the first and (fourth

dorsal interosseus, on line with the head of the

first metacarpal bone; 2) of the extensor digi-

torum, at the center of the superior one third

of the back of the forearm; 3) of the flexor

digitum profundus superficialis, on the ventral

forearm at the middle ulnar side.

Subjects were comfortably seated with their

back and feet stabilized so that their shoulders

were aligned and slightly depressed. Their feet

were spread on the floor. To facilitate the

inhibition of the spastic upper limb, the

abducted (30°) and flamed (30°) arm was first

immobilized by placing their elbow on a table,

the forearm in pronation and the hand hung from a roll of tissues (Figure 1). This func-
tional resting position was held until the fig-
gers straightened. A second relaxed functional

cutaneous position was then searched for. The elbow was supported and the forearm was held in pronation and flexion (30°) and the hand flattened on the table with the fingers apart. If necessary, a weight of 0.5 kg was added until the three first metacarpal segments to slowly relax.

Figure 1: The functional resting position in the treatment of a chronic spastic hand.

The forearm and hand were stabilized on the table in mid-supination, while the elbow was
drawn in full extension. The wrist was fully flexed, the dorsal metacarpal and phalangeal

segments were supported to encourage extension of the fingers. The patients were asked to
grip a cylinder of 2.5 cm diameter three times

with their fingers. After each grasp, rest was

facilitated by asking the subjects to check

their head, shoulders, contralateral upper limb

and lower limb for correct posture, to breathe

depthly and to think of an interesting idea.

EMG recording was made at the third trial. A

second test consisted of doing the same task,

but now at a 0° wrist extension. A sup-

dorsal metacarpal-phalangeal flexed position. When no

neuromuscular activity was seen on the oscilo-

scope, the patient was asked to grasp the same
Function of hand in post-accident patients

In the third trial, the MNK was recorded. In the second trial, the proximal phalangeal segments of the second to fifth fingers were placed in extension while the thumb was either moved out of the way or positioned in palm opposition. The wrist was held in 10-15° extension. While the grasp was made, the patients were asked to pay attention to the trace of the oscilloscope, and to activate their extensor digitorum muscle.

In the daily autonomous training program of three weeks, subjects were asked to do their best in performing the experimental tasks three times a day. They were asked to practice arm flexion and abduction against the wall in the largest range possible without bending their elbow.

Data Analysis

For each hand posture, the highest intensity of activity, varying from 0 to 5, during the effort of the grasp, was tabulated. The mean of the muscular activity was deducted. The finger movements from the thumb to the little fingers were carefully observed in respect to the range (outer, middle, inner) achieved. The speed of the complete trials of contraction was noted in observing the record. The mean of the gripping speed was tabulated.

While communicating with the patients, it was suggested that they increase the frequency of functional use of their hands, some tasks were proposed and they were required to report on their achievements.

Results

In the three postures studied during the maximal grasp effort, the first dorsal interosseus muscle was the most active (Xs: 3.0-3.4). It is this muscle which contractions tonically to a greater degree than usual in the postoperative period. In contrast the least intense activity was observed in the fourth dorsal interosseus muscle (Xs: 2.9-2.3). While the wrist was at 10° extension and the metacarpophalangeal joint flexed at 90°, the flexor digitorum superficialis muscle was less active than the first dorsal interosseus (Xs: 3.1-3.4). The extension of the proximal phalangeal joints in wrist extension slightly decreased the action of the first and fourth dorsal interosseus. The full flexed wrist affected significantly the cylindrical grip; in this position, sequential movement of third, fourth and fifth fingers was employed in three subjects (Figure 2). One patient moved the second finger in a different pattern at the three-week interval test. While the wrist was at 0-10° extension, and the metacarpophalangeal joint either flexed at 90° or extended at 0°.

Figure 2 - Example of a grasp while the wrist was fully flexed in a severe spastic patient.

Conclusion

The stability of the body permitted the improvement of coordinate activity, movement and function of the severe spastic hand or patients treated and evaluated in a brief period during the chronic stage of this disease. Further study on the application of sequential stability are needed to clarify the pathological mechanism of spasticity.

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ADHRENAID, FORCE-EMG RELATIONS IN PARESIC LIMBS OF HEMIPARETIC RLMA SUBJECTS.

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Experimental Objective

The weakness of hemiparetic spastic human subjects is usually attributed to interruption by the pathologic lesion of descending excitatory projections from forebrain to brainstem. However, recent reports describe significant disturbance in motor unit discharge rate in supraspinal lesions. It appears possible that disorganization of motor unit output, arising at the supraspinal level might also contribute to the weakness resulting from supraspinal lesions.

Our experimental objective was to assess the role of abnormalities of motor unit recruitment and discharge rate in the weakness of hemiparetic spastic human subjects. Rather than recording directly from motor units, we chose to evaluate the relationship between surface EMG and limb force in hemiparetic human subjects.

Methods

The relationship between surface EMG and force was recorded in normal and paretic arms of 17 hemiparetic, adult human subjects. The normal limb was used as a control. We also studied both arms of 11 normal subjects.

Subjects were seated with wrist strapped to a load cell attachment, and positioned with elbow flexed at 90 degrees, and forearm horizontal. The upper arm was immobilized in a splint. Pairs of pregelled, electrolyte impregnated surface electrodes, 2 cm in diameter, were placed over the biceps brachialis, brachioradialis, and the medial aspect of the triceps brachii muscles, in precisely measured location. EMG was amplified, full wave rectified and band-pass filtered at 10-100 Hz, sampled at 500 Hz by A-D converters, and stored in a 4 second epoch on computer disks. Tension measurements were also recorded and stored.

Subjects were required to generate a particular level of isometric force, by watching a display of actual wrist force against a computer controlled display window.

The mean rectified EMG level over a 1 second interval of stable tension was then calculated, and graphed against wrist tension for a range of tension, on normal and paretic sides of each subject.

Results

Relations between isometric force and surface EMG of elbow flexor muscles were derived in 17 hemiparetic and 11 normal human subjects. The hemiparetic subjects included 5 with stroke, and 5 with carotid occlusive trauma. The subjects were screened to exclude significant contralateral neurological disorder, muscle wasting, contracture or sensory loss. It was also necessary that the subjects be able to support a weight of at least 2 kg at the wrist. (Complete paralysis made the testing procedures impossible).

Over the force range studied (0-10 Kg), force-EMG relations were reliably linear, allowing quantification with linear regression analysis. Paretic and normal limbs were then compared by deriving the ratio of force-EMG slopes on paretic over that of the normal sides. Ratio values greater than 1 imply that force-EMG slopes were relatively increased in paretic elbow muscles. In 8/16 subjects, recordings of brachioradialis and in 7/16, biceps-brachialis recordings showed slope ratios significantly greater than 1 (t test, at 5% significance level). Values ranged from 0.6 to 5.3 in hemiparetic subjects (biceps brachialis BB mean 1.71 ± 1.37 S.D., brachioradialis BR mean 1.59 ± 0.84 S.D., p=17), and from 0.3 to 2.8 in normals (BR 1.19 ± 0.38 S.D., BB 0.98 ± 0.58 S.D., p=11). Subjects displaying abnormal slope ratios were not found to have significant contractions of antagonists, (such as triceps brachii). Such cocontraction would also have provided abnormal slope ratios.

Conclusions

Our results suggest that abnormal force-EMG relations arise in paretic muscles of approximately half of patients with supraspinally-mediated hemiparesis of moderate severity. Such abnormalities could arise if motor units were operating inefficiently — either because they were discharging at inappropriately low rates, because they were recruited out of normal sequence, or for both reasons. Such abnormal- ities of rate and recruitment might also contribute to the sense of weakness and of increased effort that is often reported by paretic subjects.

THE BIOFEEDBACK APPROACH TO THE REHABILITATION OF HIPKAPATROIC PATIENTS

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This work is concerned with the research activity carried out at the Centro di Ingegneria Biomedia (C.I.B.M) within the University of Naples and the "Centro di Ingegneria Biomedia" of the University of Vienna. The aim of the study is to investigate the effects of biofeedback on the rehabilitation of patients suffering from different injuries to the central nervous system.

The therapy of biofeedback comes down directly from the concept of "negative reaction" and "stability" which are basic in the physiological system controls. The application of these concepts, with the relative implications, to a biofeedback system, becomes particularly interesting if one keeps in mind the possible "learning function" which allows the organism to reconoscize longer or shorter periods of time the favorable effects of the "biofeedback" intended as a negative action when applied to the physiological system under study.

It has been attempted to bring into the rehabilitative approach this fundamental concept. In fact, the experience of several years of activity carried out at the C.I.B.M. in the treatment of centrally damaged (hipkapatroic) patients, has been taken, through treatment methods and relative "ad hoc" instruments, to realize such schemes of "physiological reaction".

The fundamental concept is that of eliminating different sensorial paths from which, through direct or indirect ways, to obtain information about an "error signal" to be made always smaller.

In the present paper the general considerations about the biofeedback approach in the neuro-rehabilitative activity in reported and the realized devices are described from a technical and functional point of view in a clinical environment.

INFLUENCE OF ELECTRICAL STIMULATION ON ANTAGONIST MUSCLE SPASTICITY

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INTRODUCTION

Therapeutic electrical stimulation of the neuromuscular system is in more and more rehabilitation centers applied to the patients suffering from different injuries to the central nervous system. The aim of this modern rehabilitation method is to reestablish the atrophied paralyzed muscles, facilitate the lost movements, improve blood flow in the stimulated extremity, prevent contractures or pressure sores. The purpose of this study was to evaluate the effects of such a cyclical neuromuscular stimulation program for the quadriceps on the spasticity of knee flexors spasticity.

METHOD

Spasticity of the knee flexor muscles was tested by placing the patient in the sitting position on a tilt table with both legs bent over the edge hanging free at the knee (Fig.1). The patient was asked to relax as much as possible. The examiner grasped the foot and brought one leg to the full flexion. The limb was then allowed to fall freely while recording knee angle with an electrogoniometer. To estimate the degree of spasticity from the knee goniogram, a relaxation index was defined as the ratio of the first minimum angle reached by the falling knee to the resting angle. By measuring the normal population it was found that the value of 1.6 of the relaxation index means no spasticity. Zero relaxation index belongs to extreme spasticity. The values between 0 and 1 (the lower leg does not reach the resting position during the first swing) correspond to considerable degree of the spasticity.

Fig.2 shows knee goniograms and knee flexors EMG when the patient was lying and sitting. As the hamstrings were during lying in their shortest position, almost no spasticity was assessed and the lower leg reached full extension during the first swing. High degree of spasticity was found immediately after this test when patient was sitting. In both cases the lower leg was bent to the full flexion and then dropped.

Three measurements were performed with 30 min of rest between measurements. Cyclic electrical stimulation was applied to the knee extensor muscles for 30 consecutive days. The cycle times of stimulation were 6 seconds of stimulation followed by 12 seconds of rest. Stimulation
A NONLINEAR ANKLE JOINT MATHEMATICAL MODEL

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INTRODUCTION

The analysis of the dynamical properties of muscle leads towards the need to represent available knowledge by means of a mathematical model that will describe muscles input-output behaviour "good enough" in a given dynamical range. In his work Wilkie (1) establishes the relationship between the muscle force and the muscle tension-length. Tendons at all (2), Kreberšek et al (3) developed a linear sixth and third order models of the ankle-joint system respectively.

Gottlieb and Agarwal (4) approximate input-output properties of the ankle-joint system by means of second order linear system.

Based on the experimental data we recognized the nonlinearity in the input-output behaviour of the ankle-joint. That is why we tried to get a better understanding of the nature of the ankle-joint muscle dynamics behaviour.

METHODOLOGY

The electromechanical servosystem (Fig. 1) using in the study of spaticity through measurements of a resistive torque appearing during the passive foot movement (3) was also applied in the study of biomechanical properties of ankle-joint of normal subject. During the measurements the subject was completely relaxed. Measured resistive torque, as a reaction of the system on the input sinusoidal displacement of the foot plate, slightly differed from a sinusoidal waveform. There were two possible reasons for that:

- the subject under consideration was not enough relaxed, this implies the additive negative input taking part (besides the passive properties of muscle) in producing the output
- there are nonlinearities in passive properties of the system.

Only the neurological input can be modified experimentally. That's why the measurement was repeated once again on the subject under a deep hypnosis. The EMG measured in such conditions was practically negligible (5 mV) and is at least the result of the noise (Fig. 2a,b).

This proved the idea about the nonlinear passive system structure being the reason for the nonsinusoidal system output. The Fourier analysis of the resistive torque (Fig. 2c) shows that both the second and the third component can not be neglected if compared with the first one. Thus the input-output relationship in the time space can be described by means of a differential equation having the coefficients depended on position and velocity of the ankle.

The order of system was determined with the following experiment. A fast change of the position was applied to the input of the system (Fig. 3). The width of the "delta" impulse was limited downward by the servosystem dynamics being equal to 240 ms. The amplitude of the delta impulse was 30°, so it was identical to the observed range of motion. The quotient of the Fourier transforms of measured output and input signals as a function of frequency was calculated. The obtained transfer
Fig. 2. a) EMG of M. Tibialis anterior or was measured with Ag-AgCl disc electrodes b) EMG of M. soleus was measured with simultaneous electrodes changes in the measurement cause the artefact.

Values of parameters in (3) cannot be measured. We obtained them by means of the parameter identification based on the reference model. The parameter values were calculated for two subjects. The results of this identification show practically equal values for parameters $k_0$ and $k_1$ widely used in models of other authors. Beside this the frequency dependence of $k_2$ and $k_3$ parameter mentioned by Agarwal (4) also can be also seen from our identification results.

CONCLUSIONS

Based on the experimental data we concluded that input-output behaviour of ankle-joint is nonlinear. This behaviour can be described by second-order nonlinear model, having coefficients dependent on input signal. The nonlinear part of the response is not negligible, so it must be taken into account in studies where neural input to the system is dominant such as myoelectric measurements and so on.

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The EMG signal was amplified by a Tekca EMG Electrometer (low freq. cut off 20 Hz; high freq, 10 KHz). Signals were stored on FM tape for computer analysis. An A/D conversion was performed and the digitized signal of each trial was submitted to IBM-370 time series program for a fast fourier transform. The parameters of the program were set to provide a frequency spectrum from 0-500 Hz. Subsequent computer software was developed to store the resultant spectrum for submission to statistical manipulation and/or three dimensional graphics.

DATA ANALYSIS

There has been a tendency in biological data acquisition to analogue average incoming signals in order to obtain definable parameters which can be more easily interpreted. In so doing much of the richness of the individual signal sample is lost and perhaps significant patterns. For these reasons we chose to create multiple individual samples from which we may search at random to build a 3D graphic. In addition the total popula-
tion of frequency patterns were analyzed by Q-factor analysis for similarity of pattern. Q-factor is a statistical procedure documented for broad pattern recognition (Boop and Billa 1981). Using a search technique, separate computer files were generated for selected biological parameters which were viewed for differences and then submitted to a program which generated a mean frequency pattern for each biological para-
eter. The Q-factor patterns and mean frequency patterns were then compared.

RESULTS

Q-factor for four factor retention is presented in Table 1. A perusal of the mean frequency pattern shows a similarity of Q-
factor to mean frequency as follows: Factor 1 -400 KMC; Factor 2 400 KMC; Factor 3 350 KMC; and Factor 4 500 KMC. The 3D graphic of the MRC (Fig.1) shows multiplot peaks spread broad based frequency. The number and frequency spectrum and the number
of peaks are reduced in the 80% MVC samples and are more distinct (Fig.2). As the force decreases to 80% the number of peaks increases and the distinctiveness of the peaks remain the same. At 40% MVC the number of peaks approximates the number of peaks present in the MVC (Fig.1) but the distinctiveness of the peaks remains much the same as the 80% and 60% MVC.

**DISCUSSION**

Inhibition is apparent in the suppression of the number of peaks as the voluntary contraction drops from maximum to lower levels. Our view is supported by Salih, Basajian and Vanderstrop (1976) who showed a suppression of neighboring motor units. The distribution of the peaks across the frequency spectrum is as a result of the variability of the duration of the separate motor unit action potentials and their summation. This variability combined with the rotation of individual motor units required of a skilled task (Lehr and Kaspar 1978 and Lehr and Hashiko 1979) account for the distinctiveness of the peaks across the various levels of contraction. The relationship of increased distinctiveness with the increased force revealed can be accounted for by the increased difficulty of combining skill and force.

**CONCLUSION**

It seems there is mounting neurophysiological evidence that the inhibitory activity in the agonist motoneuron pool is of paramount importance in smooth coordinated movements. We concur with the views of Kots (1977) that inhibitory activity of the supraspinal nature is reflected in the agonist motoneuron pool rather than exclusively in the antagonistic motoneuron pool of muscles providing for postural fixation. Using the methodology of this investigation (6-firing, mean-frequency patterns and 2D graphics) on a broad sample of muscles and movements we may approach a more comprehensive understanding of movement.

**REFERENCES**


**RESULTS AND DISCUSSION**

It has been shown that there is an EMG silent period just before a rapid voluntary movement (premotor silent period). The silence seemed to be related to a readiness of voluntary contraction, since this phenomenon could not be found in the slow contraction of the agonist. A slight stretch of the agonist can be advantage before strong force sudden movements. From videotape records and observations of jumping cats, there is always a brief but clear crouch just before the lift-off, so that all extensors are stretched (Walsley et al. 1987). The gravity might be adequatly to accomplish the passive brief stretch of the extensors. The present study was aimed to investigate the effects of passive stretch of the agonists on the appearance of premotor silent period with respect to the combinations of direction of movement and gravity.

The experiments were performed on five normal subjects, ranging in ages from 23 to 30 years old. The subject was asked to respond to a xenon lamp by performing a knee extension in the position of supine, prone and sitting (Fig. 1). Right knee extension was chosen for this experiment. The subject was asked to maintain his knee joint at 90° (0 degrees = fully extended). With the experiment's oral signal 'ready', the subject was instructed to make a slight voluntary contraction of both knee extensors (10-15% of maximum strength). Then the subject was trained to extend the right knee joint as quickly as possible when the lamp was presented 2-3 sec after the ready signal. A total of 90 trials was performed.

The action potentials of rectus femoris, vastus medialis and hamstrings of both legs were led off by bipolar surface electrodes (10 mm diameter) which were placed on the long axis of the muscle about 3 cm apart. EMG responses were sampled at the rate of 2,000 samples/sec and converted to 12-bit digital signals, which were stored on digital magnetic tape. The stored signals were lowpass filtered and digital differentiated, and displayed on the CRT. The entire data-taking, display, storage and analysis process were controlled by a programming system called RCMT residing on a NEC-3200 digital computer (32 k core memory).

**DISCUSSION**

Fig. 2 shows a characteristic pattern of EMG activity with right knee extension. The tonic discharge in the preparatory phase were observed continuously, and then abruptly disappeared on the agonist EMG activity. This abrupt or complete disappearance of EMG response preceding the phasic discharge was named the premotor silent period. Two types of premotor silence were found: one of them was the complete disappearance of EMG activity (A), and the other was decrease of excitation or an incomplete disappearance of electrical activity (B). These were labeled the "complete" and "incomplete" silences respectively.

The frequency of premotor silence in the various muscles was 60 ± 23.5% (mean ± SD) in the supine, 70 ± 26% in the sitting, and 62 ± 22% in the prone. No significant difference was found in the rate of occurrence of premotor silence in the three positions. The mean latency of silence for the five subjects was 115 ± 15 msec in the supine, 120 ± 17 msec in the sitting and 115 ± 11 msec in the prone. The mean duration of silence was 42 ± 17 msec in the supine, 46 ± 15 msec in the sitting, and 44 ± 14 msec in the prone. There were no significant differences in latency and duration of silence, and onset of phasic discharge with premotor silence in the three positions.

It was found that the premotor silence occurred not only in the agonist but rarely occurred in the antagonist simultaneously. The most consistent results were obtained in the supine (70 ± 17%). As shown in Fig. 4, no passive stretch of the knee extensor during silent period was obtained in the mean activity (ensemble average) of rectified EMG responses. These findings suggest that the appearance of premotor silent period was not due to the passive stretch of the agonist before a rapid voluntary movement.
THE STRETCH REFLEX OF PARASPINAL MUSCLES OF NORMAL MAN AND HEMIPLEGICS

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INTRODUCTION

The stretch reflex of paraspinous muscle (SRPM) was reported by Trontelj of Yugoslavia regarding to scoliosis and Carlsen of Sweden who examined SRPM of cats, both in 1973. In 1974, K. Schmidt and A. C. Y. M. McIlwraith reported SRPM of the patients with back pain, in 1974. Reading these articles, we were interested in SRPM and we thought it could be useful to investigate the effects of the postures to those back muscles in hemiplegics. Although there were some difficulties for getting SRPM, we reached to certain methods for getting SRPM and found some interesting evidences. This time we are going to talk about the effects of the positions of the neck in three different postures to the SRPM.

METHOD AND MATERIAL

The levels from which SRPM were taken, were L-2, L-1, L-3, Th-8 and Th-6. These levels were decided by counting spinous processes.

The surface electrodes were attached to the skin 6 cm. apart from posterior median line bilaterally. A couple of silver cup electrodes were used for each sides and levels. There were fixed with 2 cm. distance by adhesive tape. The ground electrode was attached to the Th-12 level at posterior median line. For getting triggering signals of tapping, EEG was taken from V-4 position.

The SRPM were elicited by the electromagnetic hammer. Tapping was given interface of spinous processes of L-1 and L-2. The force of tapping was controlled by a current transformer but acting force was changed considerably by the consistency of underlying tissue. For measurement of the force of the tapping, a load cell was attached to the skin where the tapping was given. (Fig. 1)

To avoid the effects of EEG, the R-wave of EEG was used for the triggering signal. Some delay was set between R-wave and tapping. So EEG was never superimposed on the SRPM.

SRPM was recorded in two ways. One was on the monitoring by EEG machine. One channel of EEG, 8-channels of SRPM and tapping, totally ten channels of records were taken simultaneously.

RESULTS

General impressions were followings:
1. With some exceptions, L-3 and L-1 were larger than the other levels.
2. The reflex waves were highest in prone position. The waves were lowest in sitting position. Standing is in between.
3. The reflex waves were largest in normal. There were no clear difference between right hemiplegics and left hemiplegics.
4. The reflex waves were changed by the position of neck.

Followings were the findings at each levels:
1. In the prone posture, there were some changes when the face was turned to the affected side. For instance, right L-1 lower and left Th-8 was higher in right hemiplegics in left hemiplegics when face was turned to left, right L-1 and left L-3 were higher.
2. In sitting position, right Th-8 were generally higher in left hemiplegics. In left hemiplegics, when face was turned to right, right L-1 and left L-3 were higher.
3. In standing position, left L-3 and left Th-6 got higher when face turned to right and left L-1 and left L-3 got higher in normal when face turned to left.
BLINK REFLEX INFLUENCED BY TONIC VIBRATION REFLEX (TVR) ---- SOME OBSERVATIONS UPON THE PATIENTS WITH NEUROLOGICAL DEFICIT ----

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Introduction

An unilateral electrical stimulation to the suprascapular nerve elicits the early response R1 of n. orbicularis oculi on the ipsilateral side to the shock and the later response R2 on both sides. In case of brain-stem lesion R1 is usually abolished or reduced, in case of lesion situated in the thalamus R2 is said to be reduced or abolished.

When vibration (vibratory stimuli) is given to the upper or lower extremity some influences might be observed in the amplitude and/or duration of the late response R2 in the normal situation.

This study deals with analysis of the blink reflex activity influenced by upper extremity vibration in the normal control, in case of brain-stem lesion and of thalamic lesion.

Methods

1. Electrical stimulation of 0.1 msec. duration square wave is given to the suprascapular nerve, and the blink reflex of R1 and R2 to the shock is recorded with two pairs of surface electrodes placed at the infraspinatus regions (right and left). Amplitude and duration of the late response R2 influenced by upper extremity vibration are observed.

The materials used in this study are 25 cases as a total, namely, 10 normals, 5 thalamic hemorrhage, 5 cases of tumor situated in the thalamus and 5 cases of pons glioma.

Results

1) Normal

Early response R1 on about 10msec. latency could be recognized ipsilaterally to the shock, and late response R2 on about 30msec. latency could be recognized on both sides. When 100 Hz. vibration (Hagbarth's apparatus was used) was given to the upper extremity ipsilaterally to the electrically stimulated side, the response R2 of both sides (right and left) showed dispersive change in amplitude and duration of electrical activities.

2) Thalamic lesion

(a) Hypertensive thalamic hemorrhage, 5 cases were examined.

Electrical stimulation to the suprascapular nerve contralateral to the thalamic lesion elicited R1 response on the stimulated side but no R1 on the other side, and elicited R2 response on the other side (contralateral to the stimulated side, ipsilateral side to the lesion).

Vibration given to the upper extremity contralateral to the thalamic lesion made neither augmentation nor dispersive change in amplitude and duration of R2 of contralateral side to the lesion, whereas on the ipsilateral side to the lesion some dispersive changes in amplitude and duration of R2 response could be recognized.

Electrical stimulation to the ipsilateral side to the lesion elicited R1 and R2 responses on the ipsilateral side, but no response on the other side (contralateral side to the lesion).

Vibration given to the upper extremity ipsilaterally to the lesion made dispersive changes in amplitude and duration of R2 response on the ipsilateral side but no influence on the other side.

3) Tumor

Situated in the thalamus, 5 cases were examined in this study.

Electrical stimulation to the contralateral side to the lesion elicited only R2 response on the stimulated side but no R2 response on the other side. Whereas the R2 response on the ipsilateral side could be recognized.

Vibration to the upper extremity contralateral to the thalamic lesion made neither augmentation nor dispersive change in amplitude of duration of R2
response on the stimulated side. Whereas the R, response on the ipsilateral side showed some dispersive change.

Electrical stimulation to the ipsilateral side to the thalamic lesion elicited the R, and R2 responses on the ipsilateral side. No response could be seen on the other side.

Vibration to the upper extremity ipsilateral to the lesion gave no dispersive change in the R, response on the ipsilateral side, but no change on the other side.

Discussion

In this study, the blink reflex influenced by upper extremity vibration could be observed. Whereas in the normal situation, vibration given to the upper extremity made some dispersive changes in both amplitude and duration of R, response.

Vibration given to the upper extremity contralateral to the lesion gave no dispersive change in both amplitude and duration of R, response on the contralateral side to the lesion.

INVESTIGATIONS AND THEORY

The shape of intra-muscular as well as surface-recorded signals is strongly influenced by volume conduction. When considering the rise of surface potentials the assumption of an unlimited medium must be replaced by the introduction of a boundary layer (muscle-skin-air). In the case of monopolar electrode-arrangement and little interference by potentials of adjacent muscles, signals structured like those schematically shown in Fig. 1 can be obtained from most superficial muscles. Numerical values are to indicate dimensions: latency is shortened. The inward current flow into the depolarized muscle cells of the fast motor unit is not opposed by a negative deflection from the baseline in signal A (electrode placed near endplate region). Electrodes situated distantly from the endplate region (B and C) do not record any negative deflection until depolarization progresses with muscle conduction velocity (NCV) has reached the electrodes (t2 in signal B). These electrodes register an initial positive phase (IPP) simultaneously beginning with motor-point depolarization at t1. Our experiments have shown that the IPP observed in some distance from the endplate region is generated by volume-conducted currents preceding the propagating depolarization. At a still greater distance from the depolarized area the volume conducted currents decrease and so does the initial steepness of IPP at t1 (signal C).
measurement in "belly-tendon-position" the different electrode is rarely posi-
tiond in direct proximity of the en-
plate region. Then latency is often me-
asured to the beginning of IFF, assum-
ing the first visible signal deflection to be caused by the fastest motor-units.
This is considered not possible near the
endplates, since early depolarization causes early volume-conducted currents. In
practice, however, the slow initial increase of IFF is often disturbed by
interference of potentials of adjacent muscles. In the case of a marked IFF as
it is usually recorded a few centimeters away from the endplate region, latency
of the fastest fibres cannot be measured to the beginning of IFF. Before de-
polasarization and preceding IFF have reached the near of the electrode a lot
of slower motor units are already excited and cause a strong volume-conducted
current determining the IFF's shape. Therefore, too, latency determination at
the begin of negative deflection after preceding IFF (signal h at c) only
measures the mean conduction time of many fibres.

CONSEQUENCES FOR LATENCY MEASUREMENT

The above considerations supported by experiments indicate that latency
of the fastest motor units can only be derived from signals without IFF recorded
in direct proximal endplate region. A device for fast, exact and reli-
able latency determination should be able to identify the end-
plate region with sufficient accuracy (maximum deviation about 6 cm for ob-
servation at small muscle). Greater deviations cause a decrease in steepness
of the negative deflection (along the transition from signal A to B). The
microcomputer-based system for non-invasive measurement of neuromuscular para-
eters realized in our laboratory (Fig. 2) utilizes a basic configuration con-
sisting of 5 - 9 longitudinally arranged electrodes. Electrode diameters are 4 mm,
distance from center to center is 5 mm. When the multi-electrode assembly has been easily
positioned on the muscle belly the electrode next to the endplate region is
determined by identification of the sign-
reversal between the bipolarly recorded signals (see Fig. 2). After the switching to
monopolar configuration latency is
automatically derived from the signal
recorded by the chosen electrode making use of an adaptive threshold algorithm.

With high reliability, accuracy and re-
producibility are the same as are
achievable with careful manual latency determination. Manual inspection of
Depolarization and measuring IFF is possible.

An uncomplicated electrode positioning and time-saving interactive computerized
measurement open a broader field of
clinical applications, while non-invasive
analysis of neuromuscular function.

Fig. 2: Improved system for non-invasive measurement of neuromuscular parameters

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INTRODUCTION

There are many reasons why it is of interest
to obtain information on the strain of tissue
in relation to muscular effort of freely moving
man. During recent years, the greater emphasis
on occupational medicine has called for im-
proved methods to assess working postures from
an ergonomic point of view. Electromyographic
signatures, particularly through the use of sign
spectrum analysis, have proved useful in
studies of muscular work (for a review, see [5]). The information obtained through elec-
 tromyographic methods concerns muscle load and
physiological consequences of the load as well.

However, clinical experience seems to indi-
cate that complications due to temporal or
long-term overload of muscles, in working life
or in sports, focus on the tendons rather than
on the muscle itself. Electromyographical
methods give only indirect information about the
physiological consequences for the tendons in
connection with loads on the musculotendinous
system in relation to, e.g., aging phenomena.
There is a lack of reproducible indices of the
effect of mechanical strain on tendons direc-
tly. The present preliminary report concerns
the possible use of acoustic waves to detect,
in quantitative terms, the level of strain
applied to tendinous tissue.

TEORETICAL BACKGROUND

The most obvious way for studying the physical
properties of tendons is the use of physical
resonator by investigating how the state of the
tissue affects wave propagation. Electromagnetic
field and mechanical waves would be as relevant
priori candidates for methodological develop-
ment.

From a mechanical point of view the opti-
mal information to be determined would be
a complete stress picture, which, provided the
inverse problem has unique solution, would
provide all mechanical properties of the
tendon.

It is presently unknown whether biological
tissue has a "bundle" in the microwaves - infra-
red domain, and, if a window exists, whether
or not in this domain stress would affect the
refractive index of the tissue. The subsequent
discussion will therefore be concerned with
acoustical (mechanical) waves only. Although
little is found in literature on this partic-
ular subject extensive research has been
carried out on the mechanical properties of
tendons (e.g., [2]).

Disregarding for a moment the surrounding
tissue, a tendon should be able to support
wave motion in three qualitatively different
"modes" i.e., transverse and longitudinal elas-
tic waves and pressure waves. Due to inhom-
genities, non-linearity and possible non-linear
properties of the tendon itself as well as
boundary conditions, involving effects of
shape and influence from the embedding tissue,
these "modes" will interfere and scatter in
a complicated manner. However, it is tenta-
tively assumed that the mixing is a second-
order effect, at least in the absence of rup-
ture-resistant studies of the tendon. Enough
light is cast upon the identification of the
modes and the influence of stress on their speeds of propaga-
tion. These factors could be measured with-
out a too sophisticated apparatus.

Having identified the modes the continued
investigation will start with analysing the
response to mono- and bi-frequency wave-
packets and determining any presence of struc-
tures of strong nonlinearities. Correlation of
the mode signals to the reflective informations regar-
ding the second-order effects. On the basis
of such results it should be possible to make an
electroacoustomechanical model of the tendon, and
evaluate measurements to mechanical properties.

There are of course several technical prob-
lems encountered in pursuing an investigation of
this kind. Since the signals are to be in-
jected and recorded noninvasively through sur-
rounding tissue, the most easily detectable
mode should be the transversal one. A pre-
liminary study regarding the qualitative effect of
stress on the propagation speed of this mode is reported below.

A NEW INVASIVE MEASUREMENT OF TENDON STRAIN IN VIVO

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METHOD AND REPORTS

The experimental set-up employed in the investigation is illustrated in Figure 1. The principle of the measurement is to generate a low frequency mechanical wave in the Achilles tendon by means of a vibrator. The wave, which is transmitted by the tendon is picked up at a distance a few centimeters away from the point of excitation, along the tendon (see Figure 2). The excited and detected signals are transmitted to signal processors (amplifiers, filters and detectors), and are then fed into a phase-shifting measuring instrument. The output signal from the phase meter is a measure of the phase shift between the two signals. A graphic recorder is used for documentation of the measurements. Instruments, settings etc. are given in Table 1.

The vibration generator was a microphone connected to a sinus oscillator. The phase meter was connected in such a way that the generator signal was taken as a reference, and the transmitted signal phase shift was adjusted to zero deflection on the phase meter at zero mechanical force acting on the tendon. A positive deflection on the meter indicates that the phase of the transmitting wave lags the reference. This means that the velocity of the transmitted wave is decreased.

Figure 3 shows a graphic record obtained during a series of plantar flexions applied at the ankle-joint. It is seen that a higher force was reflected as a progressive phase shift in such a way that the transmitted signal phase lag showed a continuous decrease; this indicates a higher velocity of the transmitted wave as the force on the tendon is increased. This observation was valid in all experiments carried out. The effect is always the same: signs of an increased velocity, and whenever large force changes are at hand, the effect is first-order and easily detectable. In the example given in Figure 1, the phase shift is of the order of +1.0°/s. Similar recordings carried out on the tendon of the hibectas brachii muscle have shown average phase shifts of +1.7°/s, where the force was applied at the wrist.

DISCUSSION

The set-up described above is effective in order to show the feasibility of using acoustic wave measurements to detect changes in the properties of body tissues as a result of applied forces. The results are unequivocal in this respect. Observations made during these preliminary experiments indicate, however, that there is a considerable non-linear behavior inherent in the signal transmission.

The use of sinusoidal waves and linear theory must therefore be questioned, and the quantitative measure arrived at must be taken with considerable caution at this point. It is necessary to proceed with the development of measurement techniques so as to be able to detect separately the properties of waves propagating through different tissues, and in longitudinal and the transverse modes. Such methodological development is in progress.

ACKNOWLEDGMENT

This work was supported by the Swedish Medical Research Council, grant no. 882-043-05926-02.

REFERENCES


![Figure 1](image1.png)

![Figure 2](image2.png)

![Figure 3](image3.png)

CANNON DRIVE: CONCEPT OF MOTOR UNIT FIRINGS

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Introduction

During the past two decades, considerable concern has been expressed over the question of whether the Central Nervous System controls the motor units of a muscle individually or collectively. The few investigations which have been performed to determine the firing characteristics of concurrently active motor units have supplied confusing and conflicting results. This state of affairs has been mainly due to the lack of a reliable technique for accurately identifying the occurrence of motor unit action potentials. Recent developments in our laboratory have removed this impasse.

Signal Recording and Decomposition

A technique has been developed which enables the decomposition (separation) of a myoelectric (ME) signal into its constituent motor unit action potential trains (MUAPs); (Leifer et al., 1981; Leifer et al., 1982). It consists of a multichannel (6 to 12-electrode) ME signal recording procedure, a data compression algorithm, and a hybrid visual-computer decomposition scheme. The algorithms have been implemented on a PDP 11/44 computer. Of the four major segments of the technique, the decomposition scheme is by far the most involved. The decomposition algorithm uses a multiparameter template matching routine and details of the firing statistics of the motor units to identify MUAPs in the ME signal even when they are superimposed with other MUAPs. In general, the algorithms of the decomposition scheme do not run automatically. They require input from the human operator to maintain reliability and accuracy during decomposition. A skilled operator may decompose an ME signal containing 6 MUAPs with an accuracy of 99.8%. Up to 8 MUAPs have been decomposed from one ME signal.

Experiment

Voluntary contractions were performed by thirteen normal subjects. They were all right-handed males whose age ranged from 17 to 52 years. The deltoid and first dorsal interosseous (DII) muscles of the right upper limbs were studied. The force output of each muscle was assessed by a force gauge and was displayed to the subject as a horizontal line on an oscilloscope, along with a target force line which the subject attempted to match. The subjects were asked to track increasing and decreasing force as well as constant force isometric contractions. The force-varying contractions ranged from zero to 10% of maximal force (MVC) level at these force rates: 10%, 20%, and 40% MVC/sec. The constant force isometric contractions were performed at 30% and 60% MVC. The three channels of ME signals were recorded with a specially constructed bipolar needle electrode and were stored on magnetic tape.

Data Analysis

A total of 225 individual MUAPs were examined in the two muscles. The inter-pulse interval between adjacent MUAPs belonging to the same motor unit was used to calculate an unbiased estimate of the time-varying mean firing rate of each motor unit. For both types of contractions, the temporal relationships between motor unit firing rates and the measured forces were extracted using a fast-implemented cross-correlation routines after the data had been de-filtered.

Results

The firing rates of the motor units during both the constant force and the force-varying contractions were found to fluctuate about an average value of approximately 1.5 Hz. Furthermore, the firing rates of all concurrently active motor units fluctuated in a similar fashion. Figure 1 provides a sample observation obtained from a constant-force contraction of the DII muscle. Figure 2 represents the cross-correlation among the firing rates of the motor units. Note the remarkably high values of the cross-correlation and the
virtual lack of time shift between the cross-correlations. The average cross-correlation value of 0.68 was found among the firing rates of all the concurrently active motor units in both muscles. A similar behavior was noted in all the force-varying contractions. For details, refer to De Luca et al. (1982).

It must be strongly emphasized that this dramatic presence of cross-correlation among the firing rates of motor units does not imply that the motor unit discharges are in any way synchronized.

Discussion

The commonality of the behavior of the firing rates of the concurrently active motor units in all contractions of both muscles in all subjects points to the existence of a common source which regulates the firing characteristics of the motor units. This common drive is schematically illustrated in Figure 1. One possible explanation for the observed behavior of the firing rates may be obtained by considering that the net excitatory and inhibitory input (supraspinal, segmental and intraspinal) to the motor neuron pool to the anterior horn has a simultaneous and common effect on all the motor neurons which are excited at any given time. The recruitment properties of a motoroneuron are determined by the Drive Principle, but once a motorneuron becomes active, its firing rate behavior is regulated by a common drive. It appears that when the level of excitatory inputs in the anterior horn increases, the firing rates of all active motor neurons increase in like fashion. A corresponding parallel behavior is noted for increased inhibitory inputs (or decreased excitatory inputs).

References

and/or tecto-spinal pathways (11,19). Similar influence, though not so pronounced, has been observed on the firing rate of motor units when other inputs like Qr.14, and/or cutaneous are reduced or eliminated (19). However, at all times for each individual for a particular programme of motor activity (field or plastic, inherited or learnt) there is a certain minimum rate of firing below which a programme cannot be carried out. It can be stated that sensory input to the excitatory response of motor neurones so as to provide a greater safety margin for any possible failure or need for change of programme. This establishes a certain redundancy in the central nervous system which subsumes a definite biological function (14).

The changes seen in IP1 waves on elimination of visual input are extremely significant. The reduction in peak to peak amplitude may be due to failure of as many units to respond to antidromic volley as seen under normal circumstances. It has already been demonstrated that visual input has an excitatory influence on the alpha (A) neuronal pool. The deactivation of IP1 wave indicates that some failure of Renshaw inhibition, thus some motor units are involved later are able to get depolarized (19). This shows that visual input has a general excitatory influence on neurons in the anterior horn, including Renshaw cells.

The reduction in silent period after V1 response picked up from elbow muscles in standing suggests many possible mechanism. Either the subjects become apprehensive on bilateral and are now driving the motor neuronal pool with intenser, resulting in the reduction of silent period (A,11). Or more likely in view of the above evidence of failure of Renshaw inhibition observed when subtract IP1 waves, it is the same mechanism operative in this experimental situation as well.

REFERENCES

QUANTITATIVE ANALYSIS OF JAW MUSCLE ENG

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An interactive program package was developed, written in FORTRAN, for use on a DEC PDP11/34 computer with ABR1-K A/D converter. The program allows simultaneous digitisation and storage of eight ENG-signals and a synchronisation signal. Parts of the signals can be selected for further treatment from the screen of a graphic terminal. The limits of the chosen block may be indicated by data from motion analysis (e.g. maximal gape during a masticatory cycle) or by a previously selected value of a transducer signal. The operator may then subdivide the block into up to 32 compartments of equal length and calculate a value of the rectified and integrated ENG for each compartment. The values may be expressed as percentages of the maximal value found for a particular muscle and a particular pair of electrodes within an experiment; an indication is thus obtained of relative activity of the muscle during any chosen phase of body motion. In combination with data on physiological cross sections and working lines of the muscles a biomechanical model can be constructed, which shows the direction and relative size of muscle forces acting upon a skeletal element during consecutive phases of a movement.

The program is particularly suitable for analysis of cyclical phenomena such as chewing, but it may also be adapted for analysis of relative muscle forces during other movements also. An advantage is the possibility of simultaneous analysis in multi-channel ENG at fairly high sampling frequency.

STUDIES ON PHYSIOLOGICAL TREMOR

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ABSTRACT

In this work physiological tremor in human subjects was detected by means of a very small piezoelectric accelerometer. Recorded data were then processed in order to compute the velocity spectral density (VSD) graphs of the tremor movements for further analysis and interpretation of the results.

INTRODUCTION

Physiological tremor is usually specified in terms of the amplitude and its frequency spectrum. The amplitude is invariably very small and varies from one subject to another, and to a lesser extent from time to time in the same subject. Physiological tremor also undergoes changes under the influence of some physical, physiological and psychological effects. In the so-called "normal" or "physiological" tremor pathological states are excluded.

METHODOLOGY

In this work physiological tremor movements in human subjects were detected by means of a very small and light piezoelectric accelerometer. Its output was then amplified and filtered by using precision electronic technique and apparatus which were specially designed for this work in our laboratory. Data were recorded on a magnetic tape which were then sampled and fed to a digital computer in order to compute the velocity spectral density (VSD) graphs of the tremor movements. In the spectral analysis, auto-correlation methods were used.

HYPOTHESIS

In the present work physiological tremor has been studied by attempting to cause changes in the level of each of the possible "electrical inputs" to the motor-neurone pool (Fig 1), and observing the resulting changes in the shape and the magnitude of the VSD graphs of tremor. Results are summarized below.

RESULTS

Measurements performed on nine subjects indicate that:

a) The frequency spectrum of normal tremor extends from a few Hz to about 40 Hz. (Note: This statement excludes the ocular tremor, where the spectrum may extend up to about 150 Hz).

b) In the frequency spectrum of normal tremor there is always at least one peak.

c) Tremor is different in different parts of the body.

d) Tremor is different in different planes of motion.

e) Both the amplitude and the frequency spectrum of normal tremor vary from one subject to another.

f) Both the amplitude and the frequency spectrum of normal tremor vary from time to time in the same subject.

g) In all cases, the spectrum attains a well-defined shape for long enough time durations.

h) In the case of hand and finger tremor (particularly the little finger tremor) following factors have been found to influence tremor at different levels:

- The position of the head; fixation of vision; the extent of eye rotations; mental calculations; "intention" to perform a certain task; fatigue of the finger muscles; extent of stretch of the relevant muscles; voluntary bending of the finger or hand; pulmonary ventilation;
H Bengi Studies on physiological tremor

The influence of the retinal illumination on finger tremor is negligibly small.

CONCLUSION

Results of measurements indicate that physiological tremor should not be attributed to tonic impulses reaching a muscle from a single origin. Rather, many sources of tonic innervation, such as, the higher centers, the proprioceptors, the labyrinth, the stretch reflex arc, etc appear to contribute tremor movements at different levels.


REFERENCES

Other
PROPRIOCEPTORS

MECHANICAL, Visceral, Eerie, Painful

MUSCLE CENTRAL LABYRINTH

SKELETAL MUSCLE

MECHANICAL, Visceral, Eerie, Painful

FIG. 1. Block diagram showing the origins of tonic innervation of muscle.

Other propagation

Mechanical, Visceral, Eerie, Painful

Mechanical, Visceral, Eerie, Painful

Electric, Visceral, Eerie, Painful

Figure 1. Diagrammatic representation of possible source of order energy.

Figure 2. Diagrammatic representation of possible source of order energy.

Other propagation

Mechanical, Visceral, Eerie, Painful

Mechanical, Visceral, Eerie, Painful

Electric, Visceral, Eerie, Painful

Figure 3. Diagrammatic representation of possible source of order energy.

Figure 4. Diagrammatic representation of possible source of order energy.

SARCOCHEM LENGTH AND ENHANCED ACTIVITY OF THE RABBIT MUSCULAR MUSCLES

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Sarcoceme length of the masseter, temporal, medial and lateral pterygoid and digastric muscles was determined in anesthetized rabbits, with their mandibles fixed in seven different positions, representative for the normal jaw excursions during feeding. From EMG of the same muscles recorded during feeding the distribution of integrated EMG within average chewing cycles was determined. There was a close correspondence between timing of EMG and sarcoceme shortening in both jaw openers and closers. The jaw closers started to fire at sarcoceme lengths of 2.6 - 3.1 mm and, during closure and occlusal, shorten till 2.1 - 2.3 mm. The jaw openers started firing at the same length, but did not shorten (at maximum jaw opening) beyond 2.6 - 2.7 mm. Consequently, the openers fire about the plateau of the length tension diagram, the closers at the ascending slope. During occlusion, the latter group is active at suboptimal length. During initial (first) jaw closing the speed of shortening of the jaw closers is estimated to approach 50 % of the presumed maximum speed. The rest of jaw closure and food crushing takes place under near-isometric conditions. The jaw openers act only at low shortening velocities.
Introduction.

In daily practice of rehabilitation medicine, muscle force and joint motion are generally used as a means for quantification of muscle function. In this way the quantification is performed rather subjectively, but satisfying for most practical situations. Usually muscle force is rated in a grading system, numbered from 0 (normal, healthy muscle) to 4 (no perceivable contraction in a totally paralyzed muscle).

In case of neurological disorders some electrical activity can remain present in the paralyzed muscles. It is of utmost importance that these activities can be detected in a very early stage of the prevailing process as early muscle re-education, e.g. by myoelectric feedback techniques benefits the patient by:

a. motilization of the muscle control system;

b. reducing muscle spasticity;

c. motivating the always laborious training sessions.

As there is apparently a need for assessing paralyzed muscles (graded 0 to 1), we developed a multichannel EMG recording and analyzing system with surface electrodes. This equipment should meet the following conditions:

1. the recording and analyzing procedures should be carried out easily without difficult precautions as working sterile, long lasting skin preparations with regard to the electrodes, and so on;

2. the equipment should be simple enough to be operated by medical and para-medical personnel without the assistance of a technologist;

3. the results of the measurements should be objective, reproducible and not too difficult to be interpreted.

As the surface EMG in this case is not the objective, but just a means to reach that objective, we do not use the term EMG and instead named this method of assessing (myoelectric) function myoelectric function analysis or MFA, differentiating it from the neurologists' conception of the (neural) EMG.

The Myoelectric Function Analysis.

The MFA provides a method to analyse muscle function. In this case of 12 muscles simultaneously. Although certainly applicable and useful for healthy muscles, its virtue is the application in cases of paralyzed muscles, which are graded between 0 and 1, as the only way the examiner can judge the condition of these muscles is by recording the remaining electrical activity.

By a certain method of processing the myo-signal electronically, we can quantify this signal and thus get an objective value of its. This value represents the product of the mean rectified amplitude of the signal and the time during which the measuring took place and is expressed in microvolts.

By recording the myo-signals simultaneously the examiner gets a view of the (electrical) function of the muscles being examined. From this recording he can see:

a. which muscles are still active and which are totally paralyzed;

b. whether there is or is not a disorder of the normal activity pattern or sequence of action of the muscles concerned.

From the recordings of the 12 EMG channels simultaneously one can set up a strategy for muscle re-education, especially by means of myoelectric feedback techniques.

By repeating the MFA recordings after each three months, a good check on the healing process can be followed and the treatment can be adjusted accordingly.

The equipment.

The equipment consists of a 12-channel EMG differential amplifier with a bandwidth of 00 to 1200 Hz. This amplifier is designed and built by the Institute of Medical Physics ThD, Utrecht, Netherlands. The electrodes are also manufactured there and consist of flexible Pt-Ir wires with a thickness of 0.8 mm and on each of which two discs of guilded brass are mounted. The diameter of the discs measures 7 mm and the distance between their centres is 30 mm.

Their sizes were chosen according to the commercially available double-stick discs to attach them to the skin (Scotchi nr. 2181 or Hellige nr. 217,123,01).

The EMG amplifier also contains the EMG integrator which measures the smoothed rectified EMG signal and shows the result of the processed signal as a number on a digital display.

This number represents the value of the smoothed EMG signal in microvolts and is also printed out by a small printer.

The 12 EMG signals can be recorded on paper by a Honeywell 1850 ultraviolet Visonorder. Simultaneously, the signals can be stored on magnetic tape by a 16-channel Honeywell 101 instrumentation tape recorder.

In this setting the processing can be performed on-line and off-line as well.

The EMG signals are monitored on a knot SG 4100 large-screen display.

The EMG amplifier unit has also a monitoring feature, consisting of two vertical led-arrays, green and red respectively, each of which can be switched to any channel, thus enabling the examiner to monitor any pair of muscles to be judged with regard to their interactivity. This is particularly important if one would test the possibility of muscle re-education by biofeedback.

Signal processing.

For the manner by which the signal is processed we refer to an article of the designers of the MFA equipment: "A microcomputer based instrument for electromyogram quantification" by J. Penger and B. de WIJN in Progress Report 7 (1989), Inst. Med. Phys. ThD, Utrecht, Netherlands.

In short, the signal processing takes place as follows: the raw EMG is passed through the differential preamplifier and attenuator, followed by an output amplifier. Then, a full wave rectifying and smoothing circuit delivers the smoothed rectified EMG to an integrator/display unit.

A control module determines the maximum integration period and the integrator sensitivity for all of the 16 channels so that, for a full scale EMG input signal the integrator display reaches its maximum value at the end of the selected integration period.

The actual integrator can be started by a remote push button. The integration is stopped by pushing the button again at the end of the selected maximum period, depending on which comes first.

Applications.

Generally spoken, the MFA can be applied as a method of investigation in all kinds of neurological and orthopedic disorders, but also in pure kinesiological studies with specially adapted electronics to elucidate disturbances like
MASSETERIC SILENT PERIOD IN PATIENTS SUFFERING FROM TEMPOROMANDIBULAR SYNDROME OR WITHOUT TMJ PAIN-
DISORDER.

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Beginning with Catterall's article in 1976, function of disorders of the masticatory system with associated facial pain has been recognized as an important clinical entity. The temporomandibular joint (TMJ) pain-dysfunction syndrome occurs mainly in young women and consists in a dull, achin pain centered in the ear or preauricular area and radiating variably to the temple, the side of the neck, and the angle of the jaw. The pain is usually continuous, unrelenting, unilateral, often exacerbated by non-dilator movements. Physical examination reveals evidence of joint dysfunction, as limitation of movement, deviation of the jaw on opening, and clicking in the joint. In spite of the fact that TMJ pain-dysfunction syndrome is a common cause of facial pain (Finnic and Tucker, 1978), it is frequently mistaken for other headache.

Various etiologies have been hypothesized (Guralnick et al., 1978), but pain and dysfunction are generally believed to result from spasms of the muscles of mastication (Schwarz, 1984). Tenderness involving masticatory, facial, posterior cervical, temporal and suboccipital muscles can be seen in TMJ pain-dysfunction in association with unilateral or bilateral (mixed) headache (Keck and Keeler, 1981).

On the other hand, pain and symptoms of dysfunction of the masticatory system are relatively common and probably occur also in patients suffering from migraine or other types of headache. Duration of the masseteric silent period during voluntary clench, recorded by electromyographic technique, has been proposed as a useful diagnostic measurement in TMJ dysfunction (Bessette et al., 1971; Bailey et al., 1975); the purpose of the present research is to investigate this electrophysiological parameter in patients suffering from TMJ pain-dysfunction with or without associated tensional or tension-vascular (mixed) headache as well as in patients with tensional or mixed headache who presented masticatory muscle tenderness.

METHODS

The study was performed in 40 patients (26 females, 14 males); they ranged in age from 21 to 70 years (median 41) attending the Headache Center of the Neurological Clinic of the University of Pavia. Only patients who reported also masticatory muscle tenderness were selected for this study. Each patient was interviewed about his symptoms related to headache, family history, as well as details of past medical history. All patients were evaluated neurologically and surgically every 4 weeks in the Clinical Unit of the University of Pavia. Following a second interview about their symptoms, each patient underwent a stomatognathic examination. Radiographs of the TMJ, skull, and other instrumental examinations useful to diagnosis were performed. After completion of this procedure, patients were assigned to diagnostic categories according to criteria established by an Ad Hoc Committee on Classification of Headache (1982). Tensional and mixed headache were considered in the same group. TMJ pain-dysfunction syndrome was diagnosed according to criteria proposed by Reik and Hult (1981). The TMJ pain-dysfunction patients were divided into two groups: with and without associated tensional or mixed headache.

An electromyographic investigation was performed according to Bessette et al. (1971). Each subject, without any therapy for three or more weeks, was positioned in dental chair with his back erect and the plane of occlusion in a horizontal position. To elicit jaw jerk reflex, a saline-irrigated plunger, delivering a tap to the mandibular symphysis and simultaneously triggering the beam of the oscilloscope, was employed. Bipolar silver disc surface electrodes were taped over the right and left masseter muscles and position was standardized. EMG trace was recorded during a maximal clench. Silent period was measured according to Bessette et al. (1971).

Ten normal subjects were also studied. They never had a history of pain in the TMJ area, any overt evidence of a dental malocclusion or any masticatory muscle tenderness.

RESULTS

The results obtained by electromyographic investigation are summarized in Table 1. This table also indicates the criteria for the classification of each group and the number of cases for each single group. The mean age was homogenous between the different groups.

When the means of the duration of the masseteric silent period were compared by the Student t test among different groups, we found a statistically significant difference between group A and C (p < 0.001), A and Normal subjects (p < 0.001), B and C (p < 0.001), B and Normal subjects (p < 0.001).

<table>
<thead>
<tr>
<th>TABLE 1</th>
</tr>
</thead>
<tbody>
<tr>
<td>Duration of masseteric silent period (ms)</td>
</tr>
<tr>
<td>Mean ± SD (10)</td>
</tr>
</tbody>
</table>

**STATISTICAL EVALUATION (Student's t test)**

<table>
<thead>
<tr>
<th>Subjects</th>
<th>A vs C</th>
<th>B vs C</th>
<th>Normal subjects vs C</th>
</tr>
</thead>
<tbody>
<tr>
<td>p (NS)</td>
<td>p &lt; 0.01</td>
<td>p &lt; 0.001</td>
<td>p &lt; 0.001</td>
</tr>
</tbody>
</table>

**CONCLUSIONS**

An abnormal prolongation of silent period produced by the jaw jerk reflex in patients with TMJ pain-dysfunction syndrome was demonstrated in several studies (Bessette et al., 1971; Bailey et al., 1974; Collesano et al., 1979). The mechanism of this alteration is not well known. The masseter spasm of the masticatory muscle has been hypothesized as a source of active inhibition of the masseteric motorneuron pool (Bessette et al., 1971). In tensional and mixed headache patients, this reflex can be seen also in the muscle of mastication without evidence of a dental malocclusion. The masseter spasm of the masticatory muscle has been hypothesized as a source of active inhibition of the masseteric motorneuron pool (Bessette et al., 1971).

REFERENCES


**HEADACHE GROUPS**

A: TMJ pain-dysfunction syndrome B: TMJ pain-dysfunction syndrome with associated tensional or mixed headache C: Tensional or mixed headache with masticatory muscle tenderness.

**SUBJECTS**

A = B NS
B = C p < 0.001
A = Normal subjects p < 0.001
B = Normal subjects p < 0.001
C = Normal subjects NS

**NS** = Not Significant
POSSIBILITY OF DETECTING THE RESTING-INNERVATION BY EMG DURING THE FACILITATION ACC. TO VOJTA (ESP. MMC).

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INTRODUCTION

The congenital myelodysplasia (MMC) is the most frequent damage of the peripheric nervous system in infancy. Until recently no efficient therapeutic possibilities existed. We are of the opinion that Vojta's physiotherapy - (the reflex locomotion) - is the most qualified therapy for these cases of Spina bifida. Early treatment is necessary. As a rule the disconnection between the paretic segments and the supraspinal neuronal levels is not complete. During the facilitation of the complex pattern according to Vojta we have seen clinically the following:

a) Activation of the paretic groups of muscles,
b) Motor development,
c) Prevention of secondary muscular atrophy,
d) Segmental anaesthesia is to be minimized.

In the Kinderzentrum München we have tried to document the neurological modifications since 1976.

METHODS

Systematically we have documented the objective improvements by long-term assessments and by electromyography. We employed clinically the resting motor-innervation by kinesiologic examination and by evaluation of the facilitation-effectivity of Vojta's reflex locomotion. The usual EMG was made by coaxial electrodes of the most paretic segments. First the maximal innervation (often provoked dissociatively) and then the effect of Vojta's reflex-facilitation were recorded without displacement of the electrodes. The EMG was recorded by a Medelec MS6.

RESULTS

In this investigation we have up to now examined 66 children with myelodysplasia. About 70% were of lumbosacral type. Only 45% of them were treated early, that means the beginning of the facilitation took place in the first year of life. We could not always record the EMG at the beginning of the therapy because many children were not treated in our centre. During the provoked maximal activation a typical pattern of isolated motor units was seen. The frequency-discharge of action potentials was stable. During Vojta's reflex-facilitation the following changes often occurred:

1) Increase of the frequency-discharge.
2) Stabilization of the frequency of motor units.
3) Recruitment of motor units.
4) Recruitment and "after-activity" of motor units after the effective facilitation.

In some cases of follow-up-studies (21 children) we have observed an increase of the discharge of motor units after one or more years of therapy not only during facilitation but even in the voluntary maximal innervation.

CONCLUSIONS

EMG investigations have proved that the complex facilitation pattern acc. to Vojta (reflex-locomotion) is able to activate the motor units. Therefore, in children with MMC - with incomplete spinal lesion - the clinical improvement under Vojta's facilitation, documented by EMG, should be understood as a recruitment of the resting motor neurons as setting up of those for a better isometric function of the recorded paretic muscles.

Therefore, the EMG results are supporting the clinical assessments of motor development and Vojta's neuropathological hypotheses and empiricism. We have seen similar EMG-samples even in other diseases like polio, myelitis transversa and other incomplete spinal lesions. On the other hand the EMG-investigation during the facilitation increases the possibility to quantify the primary damage and to recreate the peripheric neuron.
INTRODUCTION

Primary facilitation in the traumatic paraparesis of plexus brachii during the neonatal phase is absolutely necessary. Vojtá's segmental and complex pattern of facilitation (flexor-locomotion) minimizes the extent of the peripheral paresis or even heals it. This is our favorite clinical empiricism.

Even with very strong lesions of the segment C7 - C8 (type klumpke) we have noted that it is possible to prevent the severe atrophy of the hand if there is no complete neural lesion, we have tried to document the neural activation under the facilitation and the difference to possible voluntary maximal innervation by EMG.

METHODS

see: H. Böker, Possibility of detecting ... (5th Congress of IMK)

RESULTS

Since 1978 we have recorded 125 in 17 children with perinatal traumatic plexus paralysis brachii, mostly in a long-term study (9). Not all of these children were treated since the neonatal phase, but the beginning of the treatment was mostly after six months of life or later.

Clinically we have seen the following (however, not in all children):

1) Prevention of scoliosis-development in the first year of life
2) Normalization of the phase and even of the isometric activity in approximately normal motor patterns
3) Minimization of trophic and growth disturbances.

The EMG is able to document quantitatively the degree of the paresis and the neural recovery. Under Vojtá's segmental and complex pattern of facilitation we have always recorded a good influence of the discharge of frequency of the motor units as well as a recruitment of the effects were even stronger than the effects in the EMG-patterns of FNC.

In some long-term studies we have recorded (after one year of therapy) an increase of the motor-unit's frequency in the voluntary maximal innervation (always provoked). The density of EMG-patterns with voluntary innervation became similar to the facilitation-patterns.

CONCLUSIONS

The EMG investigations have shown that there is an increase of motor-unit activity under Vojtá's facilitation. There is a great correspondence between the long-term results of EMG and clinical repair. If the treatment is very early we can recognize that a recreation of the neural damage is approximatively possible and the innervation defect remains small, even in older defects there exists the possibility of activation of the nerves and of the segmental levels by recruitment of the motor units.

In virtue of these neuropsychological proofs we are of the opinion that the rehabilitation therapy is the therapy of first choice in peripheral neurological damages.

In clinical practice, spasticity denotes a characteristic syndrome of abnormal patterns of muscular activity, leading to hypertonus and deficit of active range of motion. Meckel reported that the response of a spastic muscle to lengthening is not the same during passive movements as during voluntary movements. Studies of passive and voluntary movements in cerebral paretic children are rare.

The aim of our study is to relate the EMG output of passive and voluntary movements with the existence of muscular hypertonus and with the range of motion of voluntary movements.

Up to now 21 cerebral paretic children were investigated in our clinic. 10 girls and 11 boys in the age from 6 to 17 years. All children were premature or suffered amnialysis at birth. 9 of them suffered paraparesis, 10 quadriaparesis and 2 children suffered hemiparesis. 7 of the patients were wheelchair bound; 1 could walk with the aid of crutches; 3 children walked without any assistance.

There was a control-group of 23 normal children in the same age range.

For the recordings the children were坐在 in a specially constructed chair. The seat, back and arm of this chair were adjustable, so that the children could sit in the most comfortable position. The thighs and hips were fixed with straps. The lower leg hung freely. Muscle activity was recorded with surface electrodes. The electrodes were placed in the quadriceps and hamstring muscles. An electrogoniometer was tied to the knee to record the angular displacement. In a gentle way passive extension and flexion movements of the knee were performed by the observer. Due to contractions, full extension was often not reached. After a pause, voluntary extension-flexion movements were recorded. The EMG and the integrated EMG of both muscles were recorded the angular displacement was also recorded.

The peak values of the integrated amplitude during the extension were computed. The mean of these peak values of 15 passive and active movements were used for further calculations. For the angular displacement the mean values of the 15 voluntary movements was also taken for further calculation. The recordings in the patients and normal subjects were performed once. For the determination of the hypertonus we used a four-point scale:

0 = complete lack of resistance
1 = weak resistance
2 = moderately strong resistance
3 = strong resistance

Both quadriceps and hamstrings were tested. The score of hypertonus of both muscle groups were added together. In the recordings of the passive movements of the normal subjects there was no electrical activity in either the quadriceps or hamstrings muscle. In spite of the longening the muscles no response activity occurred.

In the recordings of the patients four different categories were found.

Category I: In 23% there was no response activity seen in both muscle during the passive movement.

Category II: In two patients there was only activity in the quadricepsmuscles during the passive movement. The mean activity was 16.5 μV. (S.D. 0.9 μV)

Category III: In 23% there was only response activity on lengthening in the hamstrings muscle. The mean activity was 41 μV. (S.D. 20 μV)

Category IV: In this category, 43% of the patients had response activity in both the quadriceps and hamstrings muscle on lengthening during passive extension flexion movements. The mean activity in the quadriceps was 31 μV. (S.D. 20 μV) The mean activity in the hamstrings was 65 μV. (S.D. 27 μV)

It is well known that the hamstrings muscle are often more spastic, while the quadriceps muscle often more parietic.

Yet in the 3 patients of category IV there was only response activity during passive
A Positive Neuromuscular Output in Passive and Voluntary Movement to Generalized United Children

The absence of response activity in the hamstring muscles might possibly be attributed to the transcallosal inhibition of these muscles according to the procedure of Eggers.

These four categories were well related to the existence and severity of the muscular hypertonia. The muscular hypertonia was scored with the help of a four-point scale. The scores for the quadriceps and hamstring muscles were added together.

Category I: The patients without any response activity during passive movements showed only a very small increase in the normal tone.

The mean score for the hypertonia was 0.6 (S.D. 0.5).

Category II: The patients with response activity in the quadriceps muscle showed a score for hypertonia of 5.

Category III: In the patients where only response activity in the hamstrings occurred, there was a mean score for the hypertonia of 7.8 (S.D. 1.1).

Category IV: In this category where the response activity was the most prominent, the mean score for hypertonia was also the highest 16.5 (S.D. 2.7).

So, indeed, a positive correlation seems to exist between the passive activity during passive movements and the hypertonia.

With the aid of an electromyometer the active range of motion was recorded. The mean displacement in the normal subjects is 70° (S.D. 8°). The angular displacement in patients expressed as a percentage of the normal value.

Category I: The mean angular displacement in this category is 72° of the normal (S.D. 14°).

Category II: This 2 children show a mean angular displacement of 57° of the normal (S.D. 28°).

Category III: Activity in the hamstrings seems to impair the voluntary movement more than activity in the quadriceps.

The mean angular displacement in category I is 47° of the normal (S.D. 14°).

Category IV: The deficit of the voluntary movement in the biggest children in this category. The mean angular displacement is only 33° (S.D. 26°).

So the active range of motion in the patients is clearly impaired compared with the normal subjects. The positive correlation seems to exist between the response activity during passive movements and the deficit of active range of motion.

The mean activity during voluntary movements in the 23 normal subjects and the 24 patients is very different. For calculation the mean value of the integrated peak amplitude of the 15 movements was used.

In the normal subjects this mean amplitude of the agonistic quadriceps muscle amounts 590 uV (S.D. 22 uV). The mean amplitude of the antagonist activity of the hamstrings muscle is 33 uV (S.D. 15 uV). Patients show low amplitudes of the quadriceps muscle compared with the normal subjects. It is only 65 uV (S.D. 14 uV). The amplitude of the agonistic hamstrings muscle is consistent, to higher 49 uV (S.D. 17 uV).

It is quite clear that the amount of activity in the agonistic quadriceps muscle of the patients is very low. These pathological values are due to the existence of pa- ralysis in the quadriceps muscle.

The high values of the antagonist hamstrings muscle are due to the hypertonia. It is evident that the activity in quadriceps and hamstrings muscles during voluntary movements are pathological.

However, the values of the activity of the quadriceps and hamstrings muscles during voluntary movements do not show a correlation with the four-point scale for hypertonia and the active range of motion do. This is in contrast with the passive movements.

Looking at these results it seems justified to draw the following conclusion.

A positive correlation exists between the response activity on lengthening in passive movements in the quadriceps and hamstrings muscle and the hypertonia of these muscles, measured clinically on a four-point scale.

A positive correlation exists between the response activity on lengthening in passive movements in the quadriceps and hamstrings muscle and the deficit of active range of motion of the knee.

The EMI activity during voluntary movements shows pathological values. There is a decrease of the activity in the agonistic quadriceps muscle and an increase of the antagonist hamstrings muscle.

Up to now these values during voluntary movements do not show a clear correlation with the four categories as the hypertonia and active range of motion do.

More research will be done to confirm these conclusions.

Introduction

The purpose of the project is to provide hand- and quadruplegics with control over elbow and forearm rotation through FES. The control aspect is not to provide a system which will compensate for every external or internal parameter variation, nor to provide unnecessarily fast response times. Rather it is to turn a highly nonlinear plant (Stein, undated), potentially unstable, into a stable, linear, and predictable system. Compensation for changes in loads, for final error, can be carried out by the user, making use of his highly developed learning skills. This implies that the purpose of an FES controller is to give the user the impression that he is operating a linear system. A controller then can be developed whose output is proportional to the command input for a given external load (including the mass of forearm), with a final error proportional to the load. This also provides the user with a measure of load feedback (Nam, 1968).

To achieve the required elbow flexion/extension and wrist pronation/supination, just three muscles are employed: biceps (flexion, supination), triceps (extension), and pronator teres (pronation).

The agonist and antagonist muscles and joint dynamics can be represented by 4th order systems in which the actuator models are highly nonlinear (Inher & Yaffe, 1976). Those of particular concern are the muscle gain (defined as torque output divided by pulse width input, at a given stimulus current) and the damping. The gain depends both on pulse width and joint angle. At least partially negate this dependence, mean gains for the biceps and triceps are measured by the computer, prior to each series of experiments, at three elbow angles. A parabolic fit is made to these data using the computer to compensate for the nonlinearities.

A linear simulation program has been developed. This is used to examine system response and stability and assist in the design of linear controllers.

Description of Equipment

A minicomputer system has been developed based on a 'Digital' Mic-11. This uses a single constant current stimulator, together with a multiplexer and timer operated by the computer, to provide independent control of up to fifteen electrode pairs. Generally a 20 Hz stimulus rate is used, with the torque being controlled by the pulse width (50 microseconds maximum) with a typical amplitude of 100 mA. In experiments discussed in this report, the subject's arm was constrained, by means of a purpose built, curvilinear, to a horizontal plane, with the elbow free to rotate about a vertical axis. The curvilinear incorporates a stick mounted on a freely rotating platter which can, however, be fixed at a desired angle for isometric measurements. The stick carries two strain gauge bridges, to measure elbow flexion/extension torque and wrist pronation/supination torque. The wrist is located on the stick by a goniometer measuring forearm rotation, which is locked when torque measurements are made. A second potentiometer, mounted under the platter axis, measures elbow rotation.

Description and Comparison of Controllers

The essential aim of the controllers discussed below is to increase system stability, allowing an increase in loop gain and hence in linearity. Four strategies were studied:

A. Open loop, no controller.
B. Closed loop, no controller.
C. Closed loop with a linear controller. Different linear filters, developed with the aid of the simulation program, were implemented. The most promising used a pair of complex zeros to cancel the dominant poles of the plant, together with a second order Butterworth filter to stabilize the controller.
D. Closed loop with variable coactivation. Stains and others have modelled changes in muscle activity (as well as stiffness) with increasing active state. Various workers (Poli et al, 1979) have suggested that the
natural strategy employed by the human operator uses agonist/antagonist coactivation (which increases damping) to arrest fast movements. A controller was thus developed having a coactivation term which is large for a small output error and high angular velocity, but low or zero otherwise. There are numerous methods for comparing control system performance for this report a standard input was used (shown as a broken line in the figures), and the outputs (the elbow angle) compared. Parameters for each controller were fixed to achieve the best response. In Fig. 1, the open loop response, illustrating the effectiveness of measuring biceps and triceps gains separately. It also shows that reasonable results can be achieved with open loop, if the plant parameters are known. The three closed loop controllers are compared in Figs. 2 to 4. The closed loop response without a controller becomes unstable for a very low loop gain (about 1.0, when the model predicted instability at 1.5). The response is inferior to the open loop case, as is to be expected for such gains. The necessity of a controller can be seen, though at this stage it is not appropriate to say which of the two examined is superior.

Fig. 1 - Open loop response. (a) No gain compensation (note weak triceps response compared to biceps, and biceps 'dead space'). (b) Triceps and biceps gains calculated separately, and compensation made for their variation with angle.

Fig. 2 - Closed loop response with a proportional controller.

Fig. 3 - Control by cancellation of the plant's dominant complex poles.

Fig. 4 - Control by variable damping, using restitution.

In addition to work detailed above, control of wrist rotation is being examined, together with techniques of parameter identification.

References


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INTRODUCTION

The use of functional electrical stimulation (FES) to restore motor patterns in patients suffering brain damage has received a great deal of attention over the past few years and several studies recently reviewed by Vodovnik (4) have described apparent rote training of reorganization of the central nervous system due to repetitive FES. FES utilization to facilitate the acquisition process of novel tasks in asymptomatic subjects, however, still remains scarce (1,2). Moreover, its effects on the activation patterns of antagonist muscles which are known to be modified during the learning process (3) are relatively unknown. This study was designed to investigate the effects of a FES training program on the activation patterns of antagonist muscles.

METHODS

The subjects for this study were 20 males, ranging in age from 19 to 35 years. They were tested while executing a maximal speed horizontal arm adduction followed by a forearm flexion in the same plane of motion. Normand et al. (3) described the actual procedures followed during execution of the movement and the testing apparatus utilized. Each subject participated in three testing sessions during which he repeated the maximal speed arm adduction-forearm flexion movement five times each with the right and left upper segments. On each repetition, surface electrodes placed over the motor points of the pectoralis major, posterior deltoid, biceps brachi and triceps brachi were utilized to monitor the electromyographical activity of these muscles. Potentials were placed in line with the shoulder and elbow joints were used to record arm and forearm displacement. Electromyographical as well as displacement signals were amplified (Electronics Pacified, 8810/1) and recorded on photographic paper by a multichannel (fiber optic cathode ray tube recorder (Glowrecorder, Xicorder 1858), running at a speed of 200 mm/s.

Between testing sessions one and two, each subject was involved in eight physical practice training sessions; during each physical practice training session, he executed 100 practice trials of the right upper segment maximal speed movement. Testing sessions one and three were separated by eight FES training sessions during which each subject received a total of 800 stimulations applied to the pectoralis major, posterior deltoid, biceps brachi and triceps brachi of his left upper segment. A major feature of this study was that the electrical stimulation of the muscles of the left upper segment was based on the temporal sequence secured from each subject's record of his contralateral right upper limb while he executed the maximal speed arm adduction-forearm flexion movement with his right upper segment.

Four Stoelting (Model no. 5800/39) square wave stimulators delivering pulses of 1 ms duration, 45 volts above rhomboid and at a frequency of 100 Hz and a Stoelting (model no. 58000) four channel universal timing module were used to stimulate the muscles.

DATA ANALYSIS

For each trial, several variables were derived from the records. Total movement time (TMT) was the time interval between the onset of movement and 30° of forearm flexion, pronator movement time (PMT) was the time elapsed between movement onset and 45° of shoulder adduction and distal movement time (DMT) was the interval between the initiation of forearm flexion and 75° of forearm flexion. The duration of antagonist muscle activity was quantified as the time from the onset of pectoralis major (PM) and triceps brachi (TB) activity as during the execution of the movement. The time elapsed between the onset of antagonist muscle activity and the instant when at least 15° of adduction or 30° of flexion were the posterior deltoid-target latency (PMT) and the triceps brachi-target latency (TBL).

RESULTS

The results pertaining to the effects of 800 practice trials of the movement on the
Influence of Functional Electrical Stimulation on Premotor and Motor Reaction Time,


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Introduction: Physiatrists managing hemiplegic patients have often to face the problem of the drop, stiff and painful shoulder. Literature on this topic is not rich. In our statistics, taken on elderly hemiplegic patients, the drop and painful shoulder, that soon becomes also stiff, occurs with a percentage of about 25% of all the cases observed. Till now, according to the literature, the only effective therapy is prevention. In our department from some years we used electrostimulation of the deltoid muscle with the aim of reducing hypertonus and spasticity: lately we are using this therapy also in quite hypotonic hemiplegic shoulder, extending the stimulation to supra- and infraspinatus muscles. In order to investigate on the real value of this therapy, a comparative experiment has been made recently in our department.

Material and Methods: Twelve hemiplegic patients with drop painful shoulder (6 males and 4 females; age: 48 to 75; time from stroke, 1 to 6 months) have been randomized in two groups of 6; the 1st group received electrostimulation of deltoid, supra- and infraspinatus muscles for passive movements; the 2nd group received short waves, ultrasound, metabolic treatment; Parameters: Electrostimulation were for deltoid muscle: single square impulses, 10 msec width, 0.5 Hz; for supra- and infraspinatus muscles, that are more difficult to be specifically excited because less easily isolated, exponential impulses, 200 to 300 msec width and 0.5 Hz. EMC of middle deltoid and pectoralis major and sometimes of supra- and infraspinatus, X-rays and photographs of the shoulder and kinesthetics assessment have been made before and after treatment. We made use of two EMG and ENG of deltoid muscle; moreover we detected the raw EMC of pectoralis major while manually stretched through an abduction movement for 30° to 60° and external rotation for about 20° of the arm (depending on articular state of liberty): at a speed of about 40°/sec, and held in position for new seconds; sometimes we detected ECG from both deltoids for a comparison between affected and sound sides. The number of treatment sessions has been less than 15 in both groups; in particular, the 1st group received from 15 to 50 electrostimulations and kinesthetics (average, 25.5); the 2nd group had from 10 to 15 sessions of ultrasound, short waves and massage and from 20 to 45 sessions of kinesthetics (average, 29.6).

Results and Comments: Diagnostic remarks. Seven of the twelve patients (about 60%) had more or less remarkable hyperreflexia also of supra- and infraspinatus muscles besides the deltoid; 9 patients (about 75%) showed spasticity of pectoralis major.

We registered also 3 cases with electromyographic features of peripheral denervation of deltoid and infraspinatus muscles, as well as hypotonia in an early stage and oscillations of high amplitude (about 300V) in the stage of reinsertion; those cases of peripheral denervation were confirmed by EMG/nerve conduction curves, demonstrating mainly an increase of the base and a strong reduction of accommodation ratio. This could be correlated with a relapse and an incomplete recovery of the nerves, due to their chronic traction. We think that such an important possible component of the syndrome must be stressed. Therapeutical remarks. As for the deltoid atrophy, it was reduced in the 6 patients of the 1st group and in 3 of the 2nd, but increasing in the 3rd group. Voluntary activation of deltoid muscle was at least doubled up in every patient of the 1st group, while in the 2nd group the increasing was more moderate and 1 patient had no improvement at all after 25 sessions. Pectoralis major spasticity strongly reduced or abolished in the 1st group, not much or not reduced in the 2nd one.
Multichannel functional electrical stimulation of skeletal muscles in the process of walking in an effective method of rehabilitation for patients with multiple injuries of supportive-locomotor system. It allows to simultaneously train a large number of muscles under conditions similar to those of normal muscle functioning during the locomotor act. It also helps to correct many movements in order to normalize the biomechanical structure of gait and to form a more correct gait.

In the CHIP an 8-channel system has been designed, which contains an electrical stimulator, flexible multilayer electrodes, synchronisation sensors of the angles at the leg joints, pressure sensors in the sole of the footwear, as well as systems of fixation and connection. A table electrical stimulator provides series of voltage pulses for 6 pairs of electrodes with time division of channel impulses and switch division of electrodes. Synchronisation can be performed from one as well as from two sensors of analogous or discrete type along any front and at any level. The stimulation program consisting of 2 x 10 cycles is set by means of 160 switches. The duration of the program may be fixed or it may be adapted to the previous step. Sound, visual or digital indication is stimulated, as well as outputs for recorders. For convenience of use we have provided the generations of inner rhythm of the stimulation program, a large range of smoothly regulated parameters of stimulating signals.

System tests on patients have shown its high effectiveness, resistance to disturbances, convenience for use with corrections aims in patients of various etiology.
MEASUREMENTS OF GAIT CORRECTION WITH HEMIPLEGIC PATIENTS BY MEANS OF ELECTRICAL STIMULATORS

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INTRODUCTION

Rehabilitation programs for patients whose motor activity has been affected by diseases of the central nervous system include also reeducation of gait. For this specific purpose there are used various one- or six-channel electrical stimulators applied during therapy or as orthoses, stimulating one or more muscle groups. Beside clinical closely observed results of gait correction the measuring shoes (Klijajč 1979, Krajnik 1981) and electromyonometers (Bajić 1976) for an objective evaluation were included. The objective of our research was to determine the correction of the heel-on, and examine the changes accompanying the transfer of weight onto the affected side, the value of ground reaction forces on the affected extremity and the crutch, the changes in the symmetry and the velocity of gait, and the correspondence of motion in joints.

METHODOLOGY AND EXPERIMENTS

The experiments covered three groups of hemiplegics after stroke:
- 10 hemiplegics for a one-time stimulation of 6.2 muscle groups and 1 muscle group;
- 10 hemiplegics for electrical stimulation (ES) of 4 to 6 muscle groups up to 10 weeks, and
- 10 hemiplegics for one-time stimulation by means of an implantable system via the peroneal nerve.

In the first case measurements were performed successively to each instance of stimulation.
In the second case measurements were carried out before the application of ES, after the programs had been concluded, and after six months.
In the third case measurements were performed before and after the implantation, and six months later.

DESCRIPTION OF THE EXPERIMENT

The patient M.J., 45 years old, admitted to the institute eight months after stroke was selected for the programs of one-time ES on the basis of a clinical analysis of gait (Ačimović 1975) yielding 37 negative points, which represents the sum of anomalies to be observed during gait without stimulation. They were evident primarily from a bad postural control during the swing phase, a deficient control of the knee during the swing and the stance phase, characterized by a strong extension thrust, a strongly pronounced equinovarus in both phases of gait, as well as a week push-off.

ES was applied in the first phase on m. glutaeus maximus, m. quadriceps, knee flexors, m. quadriceps surae, m. peroneus and m. triceps brachii.

In the second phase we stimulated m. quadriceps and m. peroneus. In the third phase stimulation was performed via m. peroneus. In case of six stimulated muscle groups the number of negative points, namely 37, fell to 18 only. A partial correction (about 50%) of observed anomalies during the swing and stance phase, was achieved. The velocity of gait increased, the stride time shortened from 2.97 to 2.23 s. The transfer of weight onto the impaired extremity occurs immediately upon the heel-on and displays an even distribution during the stance phase (Fig. 1).

The trajectory of ground reaction forces in the direction of gait has a more central course and is no longer dislocated. There can be observed an increase in the vertical component of ground reaction forces acting upon the affected lower extremity, its distribution becomes more uniform, the crutch reaction force is weaker (Fig. 3).

The increases in the vertical component of ground reaction forces acting upon the affected lower extremity, its distribution becomes more uniform, the crutch reaction force is weaker (Fig. 3). There exists a more adequate symmetry with respect to the healthy extremity (Fig. 1). Electromyonometers show a correction of the extension thrust of the knee during the stance phase as well as a correction of the equinovarus in both phases in accordance with the clinical picture (Fig. 2).

Two channel stimulation shows a slight increase in amplitudes, namely, from 18 to 21.5. The stride time decreases from 2.23 to 1.79 s. A certain interference is to be observed in the trajectory direction displaying a dislocation in the last third of the course. No essential change is evident with respect to the weight transfer and the value of ground reaction forces. Evidence from

the goniograms is a change in the knee extension which is more jerky than in case of six-channel stimulation (Fig. 2).

One-channel ES via m. peroneus, too, discloses an increase of anomalies to 26 during gait. The velocity of gait increases - the stride time is 1.74 s. Goniograms again show a more jerky extension during the stance phase, whereas other parameters are not subject to changes.

When ES is interrupted, the anomalies recur to the same extent as before the experiment, only the velocity of gait remains a little higher - the stride time amounts to 2.03 s (Fig. 3).

Control measurements performed six months later yielded similar results even though the patient continuously applied the one-channel surface ES (Fig. 4). Therefore, we decided to apply an implantable system making possible ES via m. peroneus. The measurements after the implantation showed an increased velocity of gait with decreased stride time from 2.22 to 1.90 s.

All the other biomechanical data approached or the normal or the parameters of the unaffected side (Fig. 5).

In the second group of hemiplegics (ES 10 weeks), more pronounced effects on the correction of gait could be observed during the stimulation period than at the control six months later, except increased velocity of gait.

CONCLUSIONS

Results similar to those just described were obtained also in other groups of patients. They mostly corroborate the clinical observations and complete them from the quantitative

In a motion of an incision ...
point of view. By this method of measuring we successfully evidenced quantitative biomechanical data, particularly with respect to the changes in the transfer of weight, loading of the affected lower extremity, and changes in the velocity and the symmetry of gait. The computer out-prints made possible a not-time-consuming and up-to-date analysis of the results and the differences with regard to the application of various modes of correction when stimulating one or more muscle groups.

The results as obtained do not provide a complex explanation since an evaluation of gait cannot be given for all planes; nevertheless, they serve as sufficiently objective data, which should be regarded as in favour of a clinical application of the measuring system described.

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Fig. 1. The difference between No ES (———) and six-channel ES (———) (M. F. – H. D.)

Fig. 2. Goniograms with two channel (———) and 6-channel ES (———) (M. F. – H. D.)

Fig. 3. Goniograms with No ES (———) and one-channel ES (———) (M. F. – H. D.)

Fig. 4. Goniograms with No ES (———) and standard deviation (———) (M. F. – H. D.)

Fig. 5. The difference between No ES (———) and impl. one-channel system (———) (M. F. – H. D.)
According to a belief as prevailing in the world, the use of functional electrical stimulation (FES) is indicated primarily in cases of upper motor neuron lesions (Vodovnik et al. 1978). For this reason, some European and U.S. centers nowadays extensively apply FES on patients with spastic hemiparesis and cerebral palsy children (Gračanin 1978). With these groups of patients FES of the peripheral muscle is mostly used to prevent spastic equinus or gait correction. For the same purpose the application of FES has recently become very popular in cases of spastic paraparesis (Turk et al. 1979). FES has proven successful also with individuals suffering from complete paraplegia; its objective has been to help these patients to maintain a standing position by the stimulation of m. quadriceps, whereas by provoking flexion reflex responses of the lower extremities some patients even managed to walk while making use of a suitable support (Kralj, Bajd and Turk 1980). In all of these cases there were involved patients with upper motor neuron lesions.

Contrary to this we wished to apply FES also on certain patients with lower motor neuron lesions. In the course of gait, patients with lesions of motor roots, caused by diseases of the lumbar-sacral spine, above all heredine of intervertebral discs, manifest foot-drop (similarly as hemiparetic patients). We chose 5 such patients with one-side foot-drop, namely, 3 women and 2 men, aged between 42 and 60. These were cases with a partial lesion of motor roots after the impairment had lasted for several years and the motor deficiency no longer spontaneously improved. With all of them excitability of motor fibers of the peripheral nerve had been preserved as much as to enable us to apply such parameters of electrical stimulation as normally used with patients with upper motor neuron lesions, namely: pulse duration 0,6 ms, frequency 40 Hz, stimulation intensity up to 50 V. With each of the five patients the Ljubljana functional electrical peroneal brace (type PEMA-10) was successfully applied as an orthotic aid for the correction of foot-drop during gait. So far, however, there has not been assumed a uniform attitude towards the application of electrical stimulation with patients suffering from complete lesion of peripheral nerves. Certain institutions extensively resort to the stimulation of denervated muscles, whereas others are much more restrained or even warn against it. It is clear, of course, that by electrical stimulation lasting a few minutes or some hours a day the atrophy of denervated muscles cannot be prevented, though it might be slowed down to a certain extent.

Patients with a denervated muscle belonging to the peripheral nerve manifest a well pronounced foot-drop during gait if not corrected by an appropriate brace. With three patients suffering from this type of lesions electrical stimulation was applied to correct foot-drop. All three were male patients able to walk, one of them (aged 30 years) having a flaccid paralysis due to an impaired lower portion of the spinal cord as a result of L1 fracture, whereas the remaining two (aged 33 and 44 years) had a one-side lesion of the ischiatic nerve on the hip level due to a trauma. With all of three of them the muscularity of the peripheral nerve was completely denervated. Some months had elapsed since the occurrence of injury and with none of them more than a year. There were chosen the following stimulation parameters: pulse duration 38 ms, frequency 25 Hz, large disk surface electrodes were positioned above the muscular fascia on the anterolateral side of the shank. By an appropriately selected duration of a pulse train and suitable triggering we obtained a satisfactory dorsal flexion of the foot during the swing phase of the patients gait. Instead of a patient receiving stimulation while sitting and remaining quite passive all the time, functional stimulation forces the patient to be active, which exerts a much more favorable influence upon the lower extremities due to a better blood circulation, as well as upon a better fitness of the organism as a whole.

While not doubting the effectiveness of FES with selected patients with absolute partial lesions of the nerves of peripheral type, there remains an unsolved problem of the therapeutic value of denervated muscles. Should such stimulation be already applied on the lower extremities, we would consider it more profitable for a patient to walk instead of resting and remaining passive during the application period.

REFERENČNI:


HYPOTHESIS

Injured lower extremities' rehabilitation can be accomplished in the process of running by means of purposeful condition modification of athletes' and environment interaction and muscle activity control electric stimulation.

METHODS

12 middle and long distance runners underwent a 12-week treadmill running program (at a speed of 3-6 m/sec) during rehabilitation period after functional traumas of knee and ankle joints. In order to decrease shock loads on lower extremities we used upright elastic pulls fastened to an athlete's waist, decreasing his body weight by 12-20%.

Muscle activity control was performed by analogy computer on the basis of current movement phase information.

Biomechanical running parameters were registered in the process of training. Support and flight time was determined by special optic electronic device (error ±20 sec). Shock acceleration on athlete's waist was measured by "Bruel & Kjaer" accelerometer. Cinematic characteristics were registered by "Actionmeter-500" camera. Film analysis was performed by "Ergonomic analysis system". Before and after the experiment strength capacities of flexors and extensors of feet, ankles and things were determined.

RESULTS

Rehabilitation period of the subjects in comparison with that of the athletes who did not use the method of movement control decreased by 4 weeks. Differences of support and flight phases between sound and injured extremities decreased from 6 to 2%.

Values of shock acceleration of sound and injured extremities became insignifi-
DISCUSSION

The results confirm previous findings that the activity detected in the intercostal muscles was arising within these muscles since it was detected with wire, and wire only detected activity 1-2 mm from the tip. The activity was different in timing from that of the diaphragm, being present when activity in the diaphragm was absent. It was found well

from the diaphragm in the upper spaces of the chest; it was absent during spinal shock. It was present in the intercostal muscles when the diaphragm was silent during generalised spasm. The nature of the activity showing after-spike discharges was different from normal interference patterns. This activity thus arises locally in paraspinal muscles, and we believe it to be reflex, the afferent link being in the intercostal nerve, the stimulus being deformation of the inferior thoracic wall causing stretching of the muscle spindles of the intercostal muscles resulting in reflex contraction of these muscles.


Figure 1b. Channel 1 - S, spirometer trace; Channel 2 - B/C-5, eighth intercostal space surface electrode; Channel 3 - B/C-6, eighth intercostal space, wire electrode; Channel 4 - D, diaphragm.

Figure 3b. Channel 1 - S, spirometer trace; Channel 2 - B/C-5, third intercostal space, wire electrode; Channel 3 - D, diaphragm, intra-oesophageal electrode; Channel 4 - A/C-5, fourth intercostal space, wire electrode. Activity detected with wire in the third and fourth spaces is well removed from the diaphragm.

INTRODUCTION

Our purpose is to focus attention on the occurrence of a previously unreported phenomenon, dyspnoea during generalised spasms in tetraplegic patients. Thirteen patients with complete or incomplete spinal injuries ranging from C5 to T6 as the last intact segmental level had an intra-oesophageal electromyographic electrode passed as part of an investigation into their ventilatory mechanics. During these ongoing investigations, inquiry was made about breathing during generalised spasm. Three patients reported dyspnoea during a generalised spasm. This phenomenon was not present immediately after injury when the patient was in spinal shock. It was found during a generalised muscular spasm; the earliest time it was elicited was one to six weeks after injury and was accompanied by a sensation of breathlessness. We suggest that this phenomenon is a result of a spasm of the diaphragm.

METHOD OF INVESTIGATION

Three patients with complete lesions of the cervical cord were studied. The patients were investigated electromyographically using bipolar fine wire electrode for the intercostal muscles. An intra-oesophageal electromyographic electrode was used to record the diaphragmatic (ME). The combination of intra-oesophageal and bipolar fine wire electrodes allow the differentiation of the diaphragmatic and intercostal (ME). Simultaneous spirometry was carried out, Silver and Leib (previous presentation).

OBSERVATIONS

Three patients described the dyspnoea in their chest as uncomfortable, but not painful, being a tightness below the ribs, as though someone was sitting on the chest, or as an explosion of air within the chest. The dyspnoea was a continuation of generalised body spasms which were initiated by sudden movements such as having their position changed, when they recorded passive movements to the area of the cords or as part of another reflex, for example palmar stimulation. The common pattern of general spasm of the whole body was clenching of the fingers, flexion of the elbows, adduction of the shoulder or extension of the whole arm with extension of the fingers; whether the arms were flexed or extended the lower limbs were involved, also either flexing or extending. The most common method of producing a generalised spasm was by taking a deep breath. Voluntary breathing was not possible during the most component of the generalised spasm nor could the dyspnoea be stopped voluntarily. Onset of the dyspnoea always followed the initiation of a general spasm and was subject to the same conditions which influence the onset of the general spasm. These are not present during spinal shock but appear at about six weeks when spontaneously developing in the voluntary muscles. A generalised spasm is more likely to occur when the patient has been maintained in the same position for a prolonged period, in a colder environment or if anxious about a situation. Spasms are reflex in nature and admit thus once a spasm has occurred subsequent spasms are more difficult to eliclt and of decreased intensity. The duration of the dyspnoea following a generalised spasm was variable both between spasms in an individual patient and between patients.

We concluded that the dyspnoea was a spasm of the diaphragm based on EMG activity, spirometric activity and clinical observations. The beginning of the record shows normal ventilation in the patient evidenced by the spirometry trace. (A) A deep breath with an increasing contraction of the diaphragm is noted, which is interrupted by a brief pause during expiration. Diaphragmatic EMG activity indicates two inspirations as preparation for coughs following which the patient was requested to strain by trying to expire against a closed glottis. It should be noted in the diaphragmatic lead that the depressive after-spike discharges are abnormal. (B) Upon inspiration to a deep breath the beginning of a generalised spasm is initiated. The spirometric trace shows that the breath is held in inspiration, expiration is not begun until
SUMMARY

We have not seen this form of dysphonia except in patients with C.5 lesions. The present spirometric, electromyographic and clinical observations suggest that the generalised spasm involves the diaphragm. All three patients had reduced vital capacities. The only patients that showed the full evolution of the spasm to involve the shoulders and diaphragm were those in whom the diaphragm was partially paralysed.


METHOD

Neuromuscular synapse testing was performed according to technical modifications of the method described by Jusid et al. (1977). This development permitted high constancy of recorded muscle signals inspite of voluntary contraction movements.

RESULTS

No significant difference between exposed groups and the control group was found.

SUBJECTS

Two groups (I and II) of healthy agricultural workers and one group of healthy spraymen (III) were exposed to organophosphorous pesticides of different intensities. An additional group of non-exposed agricultural workers (IV) was also analysed for comparative purposes. Before and during exposure, cholinesterase activity was determined, occupational, medical, and neurological case histories were taken, and during the same session electromyographic synapse testing was performed, Jusid et al. (1980).

Electromyographic neuromuscular synapse testing was performed also in two patients with a very low blood cholinesterase activity due to suicidal poisoning with organophosphorous pesticides, Jusid et al. (1977).
signs of electromyographic neuromuscular synapse insufficiency. The maximal afferent and efferent conduction velocities were within the normal range. A decrease of potential amplitude was only seen during the 50 Hz stimulation. The 3 Hz stimulation for 30 seconds before the 50 Hz stimulation provoked potentials of entirely stable amplitude. With an additional 3 Hz stimulation the previous (not bigger) amplitude was obtained already with the second stimulus (i.e., no post-tetanic potentiation was observed).

**DISCUSSION**

In this study electromyographic results are interpreted as not being indicative of neuromuscular synapse insufficiency. The results of subjects who experienced different grades of exposure to organophosphorus pesticides (Groups I, II, and III) do not differ significantly from those obtained in healthy adult controls (Group IV). In healthy individuals, amplitude changes can be found which are probably due to changes in the configuration of the underlying surface of the recording electrode. In a large group of healthy adults, almost the same percentage distribution of different amplitude changes was found as was found for both exposed and nonexposed groups in the present study. The percentage of constant findings decreased from 75% to 44%, with a prolongation of isometric contraction from 10 sec to 60 sec. During re-testing more attention was directed toward making the contraction as isometric as possible; the percentage of constant findings increased again to 72% in spite of the same duration of activation. Accordingly, it should be concluded that every change in evoked muscle potential amplitude should not represent neuromuscular insufficiency.

Stålberg et al. 1978, found also normal "jitter" of single fiber electromyography in a group of organophosphorous insecticides exposed workers with lowered cholinesterase activity.

Acetylcholinesterase (AChE) activity was low throughout the testing analysis of suicide cases. It is well known that in myasthenic patients neuromuscular synapse testing results improve after application of cholinesterase inhibitors, and blood cholinesterase values decrease. Our patients did not show electromyographic signs of neuromuscular synapse insufficiency, although they had very low blood cholinesterase activity values. In view of the above, the following questions should be considered: a) Is blood AChE sufficiently representative of neuromuscular synapse AChE? b) Is AChE as important in neuromuscular synapse function as is generally thought. c) As no neuromuscular synapse changes were found in severe cases of poisoning should the weakness be attributed to spinal cord involvement? Fasciculations were outstanding features of acute poisoning with depressed spinal reflexes, or even enhanced later. AChE is accumulated in the gray substance of the central nervous system, and is present in all anterior horn cells, the cell body having the highest AChE activity.

**LITERATURE**


INTRODUCTION

We chose the electromyographic and the teratographic methods for early diagnosis of vibration syndrome. Both methods permit the strict orientation in functional state of each finger separately in both hands.

MATERIAL AND METHODS

We examined 15 workers exposed to local vibration (I group) and 7 workers working in the same factory, not being exposed to vibration (II group). The exposure to vibration was sometimes difficult to determine, as some of workers are using vibrating tools only temporarily.

We have performed following examinations: 1/ sensory nerve conduction test of n. radialis, n. medianus and n. ulnaris in the region of finger-wrist, by using Elisa electromyograph; 2/ teratographic investigation of dorsalis and palmaris surface of both hands before and after cooling test. The following parameters were analyzed: I/ the subjective and the objective response threshold in mV, 2/ the maximal amplitude of potential perceived from sensory nerve in mV, 3/ the conduction velocity in m/sec; above mentioned parameters were measured for each finger separately. In such a way we have obtained 42 results for each person.

Temperature measurements we took twice before and after cooling test. For the statistical analysis the results were evaluated in points number.

RESULTS

The dispersion of obtained parameters characterizing the sensory nerve conduction was rather big. The potentials amplitude varied from zero to 20 mV and the conduction velocity from 15 m/sec to 59 m/sec.

We present the typical examples of electromyograms. The first /Fig. 1/ belong to worker by whom all conductibility parameters were within normal limits.

The Fig. 2 presents electromyograms of worker, by whom 26 of total 40 parameters surpassed normal limits ± S.D. By this worker we did not state any evoked potential when stimulating IV and V fingers, in spite of the use a maximal impulse.

On the Fig. 3 and 4 we present the diagrams of termograms of the same workers. Relating to the first the termogram shows normal temperature of finger after cooling test.

The electromyogram with low amplitude find their analogue in the termogram with characteristic lowering of temperature after cooling test.

Concentrated results are examples of obvious correlation of sensory nerve conduction and microcirculation however not in every one examined case the correlation was so obvious.

CONCLUSIONS

It was proved in both groups the statistically significant straight correlation between the accumulated parameters each electromyogram and termograms evaluated in points number.

The pathomorphological changes in sensory nerve conduction are more marked as these in termograms. The diagnostic use of electromyographic and teratography helps in early detection of vibration syndrome.
The present project is undertaken to investigate the acquisition of skill in high skill manual work. The aim of the project is to arrive at an improved training technique for welders. Initially working technique is studied in standardised welding tasks by a group of experienced welders and in a group of persons who are unexperienced in welding.

Parameters of interest in this study include: muscle involvement as measured by EMG; extent of localized muscle fatigue; skillfulness in the pursuit of the task as measured in terms of endpoint stability related to the welding joint.

**METHODS**

In order to be able to follow the acquisition of welding skills, we have used a simulator developed at the MEC Environmental Physiology Unit, London School of Hygiene and Tropical Medicine (1). Figure 1 shows the simulator in an overhead task, and a real welding situation for reference.

The simulator consists of an artificial welding joint with a light-sensitive device travelling underneath the joint. At the tip of a simulated welding rod there is a light source affixed. When the electrode tip is properly positioned at the joint, the light-sensitive device travels on at a realistic welding speed, if not, it stops and gives a click indicating an error. The endpoint stability can be measured using the photosensitive device output. During the task, which requires some 100 s, the welding rod withdraws at a realistic speed. Two welding situations are being subject to study: one overhead and one downhand.

Hypoelectric signals are collected in four muscles during the simulated task. Initially, the upper trapezius and the medial deltoid muscle, and forearm flexor and extensor are monitored. The signals are directly fed into a PDP 11/44 computer and analyzed subsequently. Of primary interest in the EMG part of the signal analysis is to have appropriate means for quality control of the signals. In the project, quality control is being undertaken according to methods developed by Arvidsson (2).

Results of the investigation are presented in terms of EM solaeity level in microvolts p.m., fatigue measure in relative mean frequency change (R), and endpoint stability in mean and standard deviation of simulator sensor output. Graphical output on digital plotter gives information on the time course of the signals.

**RESULTS AND DISCUSSION**

Of major interest in the initial phase of the project has been to investigate the reliability and efficiency of the data acquisition system, and to study the properties of the simulator in the tasks chosen. Figure 2 shows an example of a recording from an experienced welder in an overhead welding task. From the top, the activity levels of the trapezius, the deltoid, the extensor and the flexor are given (A1), as well as the corresponding fatigue plots (right). Outlined data points indicate data not accepted. It can be seen in this example that channel 4 has almost no accepted data points, which indicates very low activity level (muscle not engaged). In channel 2, the fatigue plot shows the characteristic decaying effect attributable to localized muscle fatigue. The analysis gives a fatigue probability of 1 in this channel, whereas in the other channels there is no significant fatigue. At the bottom, the mean and standard deviation of the position signal is given.

In order to validate the method from the point of view of muscle engagement, a study was undertaken where the same person (five experienced welders) worked in the simulator and in real in the two standard positions chosen. The level of muscle engagement was compared. Figure 3 shows the result in terms of a correlation diagram. It is seen that there are some deviations from perfect correlation (r=0.90), but that there is a reasonable good correspondence, taking into account that we have observed some training effects on the simulator. This aspect is being studied presently.

Some preliminary observations in the early phases of the project are the following:

- Muscle engagement is higher in all muscles in overhead welding than in downhand welding;
- Endpoint instability increases predictably with fatigue;
- Endpoint instability, particularly the rapid component, is higher in overhead welding than in downhand welding.

**ACKNOWLEDGMENT**

This investigation is supported by the Swedish Work Environment Fund.

**REFERENCES**

INTRODUCTION

Fatigue of human skeletal muscle is classically studied using two kinds of techniques for characterizing the electromyogram (EMG) recorded from the muscle surface: a temporal analysis (e.g., integrated EMG computation) or a frequential analysis (e.g., mean power frequency - MFP - determination). For isometric submaximal contractions, an increase in the amplitude of the integrated EMG as a result of fatigue is described by several authors (Litwold et al., 1960; Lind and Pettersson, 1979) and associated with failure in excitation-contraction coupling. Likewise, in the course of fatigue, several experiments have shown a decreasing MFP value (Lindstrom et al., 1979; Komit and Tesch, 1979) described as reduction in muscle action potential velocity. In the present investigation, an attempt was made to evaluate fatigueability at the level of an heterogeneous muscular group, the triceps surae.

METHOD

Six healthy male subjects participated in the study. They were seated with the right foot on the plate of an ankle ergometer, the knee being flexed at 150° and the ankle slightly dorsiflexed. The horizontal axis through the medial malleolus coincided approximately with the rotation axis of the mechanical system. In order to perform isometric torques of plantarflexion, the ergometer was linked to a plethysmographic dynamometer. The EMG of each muscle of the group (soleus, gastrocnemius medialis, gastrocnemius lateralis) was detected by means of bipolar surface electrodes. After amplification, these EMGs and the output signal of the dynamometer were recorded on magnetic tape.

Each subject performed three experiments. In the first, after determination of the maximal voluntary contraction (MVC), a series of brief (5s) isometric plantarflexions at eight contraction levels (from 10 to 80% of MVC) were performed by the subject with at least one minute rest between each contraction. In the second experiment, the subject was asked to maintain a torque of 80% of MVC. During both experiments the signal from the dynamometer was used to provide a visual reference to help the subject maintain the predetermined torque (isometric isometric contractions). In the third experiment, the subject had to maintain a given value of EMG ("isometric" contraction) at the level of one muscle chosen as a reference in order to show the differences in the behavior of each muscle inside the group. For this purpose, the EMG of the "reference" muscle was sent to a microcomputer based system (BRUCHE and COULZIN, 1983) in order to calculate in real time the electrical energy of the signal over a 1 second period using a shifting average method. The result was then sent to an external amplifier of an oscilloscope after digital analog conversion. Feedback control was made by the subject keeping the oscilloscope trace at a predetermined level (electrical energy developed by the muscle at the beginning of an effort of 80% MVC).

DATA ANALYSIS

The EMG signals were sampled into a Hewlett-Packard spectrum analyser which computed a mean spectrum by calculating the rms mean value of the 32 spectra obtained from consecutive time windows of 0.5 s duration. The resulting spectrum was defined by 106 points in module and phase in the 4 Hz-156 Hz bandwidth. The module values were transmitted to a microcomputer by means of a GPIB interface system; then, the spectrum was divided into equal bands. The computer calculated the total energy and the MFP of the mean spectrum and the relative energy of each bandwidth. The results were visualized in a graphical or numerical form on a video display unit and printed on an alphanumeric/graphic printer.

RESULTS

During brief isometric contractions the MFP tended to decrease in relation to the total electrical energy for the lowest contraction levels. At the same time, a decreasing in the relative energy of the first bandwidth (4 - 20 Hz) was found (see fig. 1).

For isometric isometric contractions, after some rearrangements in the EMG at the beginning of the test, two kinds of results were found: a continuous increase in energy occurred at the level of each muscle or "alternations" (decrease in EMG of a muscle associated with increase at the level of an antagonist). In the first case, an increase in the relative energy of the first band was always found while in some tests, the classical decrease in MFP was not obvious. In the second case, alternations induced a large variability in the spectral parameters.

For "isometric" contractions, it was essentially found that the global efficiency of the group was decreasing with time as shown by the decrease of the external torque developed for comparable electrical activities of the muscles in the group. In some tests, small variations and even "silence" in the EMGs of the controlled muscles were found (see fig. 2). No significant variation in the spectral parameters preceded these variations.

CONCLUSION

From these results it appears that, at the level of an heterogeneous muscular group, modifications in EMG spectral patterns can be induced by other causes than fatigue. For example, when "alternations" are present, a decrease in EMG activity can lead to a decrease in MFP and the interpretation of this spectral variation in terms of fatigue could be erroneous (see also BrucHe and couLzin, 1983). In some experimental conditions, the use of MFP as a fatigue indicator seems to be less relevant than the increase in energy of the lowest bandwidth. This result could be due to the fact that the next part of energy is found around the MFP: thus, a slight variation in the median bandwidths can lead to notable fluctuations in MFP. Further work concerning the relative behavior of each muscle inside the group and its interpretation in terms of resistance to fatigue is actually in progress.

REFERENCES


INTRODUCTION

Objective signs of localized fatigue in skeletal muscles can be obtained by amplitude and frequency analysis of the EMG-signal originating from corresponding motor units.

To detect and follow the wellknown changes in the EMG signal spectrum due to fatigue, autocorrelation and FFT-principles (ordinarily off-line) are used in many laboratories. This is an elaborate, but also a rather expensive and time-consuming task. To be efficient a rather large and fast computer system with A/D-converter, a good FFT-algorithm, a substantial primary as well as secondary memory etc., must be used.

A much faster and cheaper approach is to use the zero-cross method. This can be performed on-line by a small micro-computer system without A/D-converter - in fact, at least a tenfold decrease in "computer power" as well as in cost.

MATERIALS AND METHODS

The basic aim of this work was to compare the results from ordinary FFT-analysis with those obtained with the much simpler zero-cross measurement technique. This was done by analysing the same series of fatigue experiments, parts by the FFT-method on a medium-sized laboratory computer and main parts by zero-cross technique on a small table microcomputer (ABC-80). The analysed EMG signals were sampled from different isotonic and dynamic fatigue experiments on the elbow flexors.

The main difference between the methods is the treatment of the signal. In Fourier analysis the signal is decomposed to a set of sinusoidal components. This was done by sampling the entire signal and applying a time window (Ranning), then a FFT-algorithm calculated a set of frequency component parameters from which the signal spectrum was obtained.

Each achieved spectrum can be successfully characterized by a single parameter, the Mean Spectral Frequency (MSF). The MSF is the quotient between the first and the zeroth spectral moment i.e. the center of gravity of the spectrum which makes it quite sensitive to changes in the spectral composition of the signal. This estimator has among others been used by Kwatny 1970, Lindström et al., 1977, Petrofsky 1979, Hagberg et al., 1982.

By definition the zero-cross frequency is the reciprocal of the time between three successive zero crossings of a signal. For a pure sinusoidal signal the normal definition of frequency coincides with the zero-cross frequency. For any other shape of the signal the zero-cross frequency can be interpreted as the mean value of the instantaneous frequency. Due to the randomness of the gross EMG signal then for a larger number of measured points the joined results will have a tendency towards an average of the different frequency components of the signal.

The EMG signal was fed to one of the microcomputers input ports via a base line comparator. The achieved "sign signal" was then scanned by an assembler routine with an Idle loop to await a positive flank and then measured the time until next positive flank. If the measured time interval was accepted it was picked up by the main program which handled the plotting of regression analysis.

The MSF usually shows an exponential decay with time and developing fatigue. This seems also to be true for the zero-cross frequency. To facilitate the comparison between the two methods regression analysis (least square) was performed on both MSF and zero-cross data versus time to the function:

\[ f = A \exp(-Bt) \]

where A and B are regression coefficients.

RESULTS AND CONCLUSIONS

Even if the MSF and the zero-cross frequency are measured in different ways they both represent averages over frequency of the analyzed EMG signal.

Inevitably there is a much larger spreading in each measured zero-cross frequency point than in a MSF value. However already an averaging over 10 measured points renders the variance almost down to the same level as MSF data. This is seen in the histogram in fig 1 where each horizontal bar represents the average of the last 10 points. Major advantage with the zero-cross method are that the results can be obtained during the course of the experiment and with a rather simple and inexpensive equipment.

Both methods suffers from the limitation that they may not give unambiguous results at very low contraction levels. However at higher levels of contraction the findings are that both in sustained and dynamic load situations the results are in practice quite comparable.

REFERENCES


MYOELECTRIC BACK MUSCLE ACTIVITY DURING BUILDING WORK. A COMPARISON OF BRICKLAYERS AND PAINTERS.


INTRODUCTION

The load on the spine has been estimated through measurements of the myoelectric back muscle activity, during bricklaying and painting. The study is part of an epidemiologic survey of low back pain in different types of building work, and is an attempt to quantify exposure of physical load on the spine.

MATERIALS AND METHODS

The study was performed on 12 bricklayers and 21 painters while performing normal work activities. They were all healthy, and physical examination revealed no pathological findings of significance. The myoelectric signals were picked up by means of bipolar surface electrodes which were glued to the skin on both sides of the spine 3 cm lateral to the midline at the levels of L3 and T6. The signals were amplified in body carried amplifiers and fed via a multiwire cable to a signal acquisition system for storage on magnetic tape and subsequent analysis. During the work the different work tasks were coded and the code recorded on the tape.

The analysis procedures were entirely computerized. The purpose of the signal processing was to estimate the amplitude of the myoelectric signals (m.u.u.) and also change in the power spectrum of the myoelectric signals. Mean values and standard deviations of the signal amplitude were calculated for the different classes of work tasks. Also, the variations of the muscle signals were summarized in amplitude histograms.

The purpose of the power spectrum analysis was to study effects of fatigue on the myoelectric signals. It was performed as described by Lindström, Kadesfor, Petersén (1977). The power spectrum analysis was done at the same time as the amplitude analysis in the computer. This enabled us to at the same time perform signal qualification and removal of signal segments disturbed by artifacts (Arvidsson, A., Lindström, L. & Ortengren, R. 1981).

Shifting in the power spectrum towards lower frequencies indicate an influence of fatigue on the muscles. Center frequency variations were summarized in the same way as the signal amplitude variations.

RESULTS

The mean myoelectric amplitude in the lumbar region was significantly higher in the bricklayers than in the painters. Significant changes in the power spectrum were not found for either bricklayers or painters.

CONCLUSIONS

The study shows that the posterior muscles of the lumbar region of the back are more active during bricklaying than during painting, indicating a higher work load. Both work activities permit periods of rest which are sufficient to restore the muscle metabolism. Therefore, there were no indications of local muscle fatigue.

REFERENCE


of frontal fasciculations comprises two controlled lockers, reacting to bending with respect to the horizontal. The sensor is mounted on the back of a patient. At the bending exceeding the established level the sensor generates triggering of an appropriate channel of the electromotor and excitation of muscles.

There were 10 boys and 12 girls at the age from 10 to 16 years affected by infantile cerebral palsy in the form of spastic diplegia with nonsignificant symptoms of a mental retardation. These patients walking independently were given a course of treatment with the help of the orthosis. At their entrance to the clinic the patients were walking on the bent, adducted legs, rotated insido, making jerky movements in the frontal plane. In all the patients a positive Trendelenberg symptom was marked. The course of treatment consisted of 30 sessions each with duration up to one and a half hour of walking on the level ground, with 5-min intervals every 15 minutes for a rest and control of the orthosis adjustment.

The immediate effect of switching on the orthosis is expressed in the distinct visual decrease of the truck flexions in the frontal plane during walking. Trendelenberg symptom disappeared and gait pattern improvement. A decrease of the body flexions on an average to 50 per cent has been registered, with normalization in 9 cases. Increase of stability, improvement of biomechanical parameters of walking relief of walking and awareness of a normal comfort by the patient have been retained after the course of treatment gradually decreasing during a half of a year.

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