UDK 612.76.014.412+618.
8-009.11-031-089-28

CODEN : ZDVEA7

YU ISSN 0350-0063

zdravstveni vestnik

JOURNAL OF SLOVENIAN MEDICAL SOCIETY

5th CONGRESS OF I. S. E. K.
Ljubljana, June 21.—25. 1982

Edited by Franjo Gračanin

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ARTICLES

This yournal is regulary indexed and/or abstracted in: BIOLOGICAL ABSTRACTS, BIOMEDICINA SLOVENICA, CURRENT CONTENTS/CLINICAL PRACTICE, EXCERPTA MEDICA, INFORMATION RETRIEVAL ABSTRACTS JOURNALS

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CODEN : ZDVEA7

YU ISSN 0350-0063

...tveni vestnik

GLASILO SLOVENSKEGA ZDRAVNIŠKEGA DRUŠTVA

5. KONGRES I. S. E. K.
Ljubljana, 21.—25. 6. 1982

Uredil: Franjo Gračanin

VSEBINA
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To revijo redno indeksirajo in/ali abstrahirajo: BIOLOGICAL ABSTRACTS, BIOMEDICINA SLOVENICA, CURRENT CONTENTS/CLINICAL PRACTICE, EXCERPTA MEDICA, INFORMATION RETRIEVAL ABSTRACTS JOURNALS

Zdrav Vestn	Letnik 51	Supl. I	Str. 1—218	Ljubljana	Junij 1982

Tehnični organizator / Technical organizer:
KOMPAS YUGOSLAVIA, Congress agency MAGISTRAT, 61000 Ljubljana, Pražakova 4, tel. (061) 327 761

Tehnični urednik / Technical Editor:
J. Laurenčič

Tisk / Printed by:
Ogrizek-Treve

Naklada / Printed in
500 izvodov / copies

ZV glavni in odgovorni urednik / Editor-in-Chief:
M. Kališnik

Urednik / Editor:
M. Lavrič

Tajnica uredništva / Secretary of the Editorial Office:
K. Jovanovič

UDK 612.76.014.421+618.8-009.11-031-089.28
CODEN : ZDVEA7
YU ISSN 0350-0063

Zdravstveni vestnik LETNIK 50 — ŠTEVILKA 12

ZDRAV VESTN 51 /1982/ - 5th CONGR I.S.E.K. LJUBLJANA 82

BIOMECHANICAL MODELS OF BIPED LOCOMOTION

Miomir Vukobratović, Robotic Department, Mihailo Pupin Institute, Beograd, Yugoslavia

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 12 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 paper 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 efforst 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 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 kinematical system properties and on the cosmesis 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 broadeness, and voluminosity of the simulation results concerning symmetrical and asymmetrical gait, as well all the relevant derived dynamic values, 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 [4], 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 kinematics to the lower extremities (gait type), are practically identical with the measured and recorded values on the human baing. This conclusion concerns first of all the character and values of the dynamical reaction forces 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 dynamics.

It should be pointed out that the zero-moment point is of great practical significance. It is the conditional name for the instantaneous point at 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 τ be the zero-moment point (Fig. 1). According to D´Alambert´s principle, the conditions of dynamic equilibrium, called the dynamic connections, are

$$\begin{aligned} \sum_{i=1}^{n} (\vec{\rho}_{\tau i}\times(\vec{F}_i+\vec{G}_i) + \vec{M}_{Fi})\vec{e}_x &= 0 \\ \sum_{i=1}^{n} (\vec{\rho}_{\tau i}\times(\vec{F}_i+\vec{G}_i) + \vec{M}_{Fi})\vec{e}_y &= 0 \end{aligned} \qquad (1)$$

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where: $\vec{\rho}_{\tau i}$ is the vector from point τ to the center of gravity of the i-th element; $\vec{F}_i$, $\vec{M}_{Fi}$ are the main vector and main moment of the inertial force of the i-th element; $\vec{G}_i$ is the weight of particular elements; and $\vec{e}_x$, $\vec{e}_y$ are unit vectors of orthogonal axes, X and Y though the point τ.

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

$$\sum_{i=1}^{n} (\vec{M}_{Fi}+\vec{\rho}_i\times\vec{F}_i)\vec{e}_\zeta = 0 \qquad (2)$$

where $\vec{\rho}_i$ is the vector from the zero-moment point to the penetration point of axis ζ on the contact surface between the foot and ground, and $\vec{e}_\zeta$ is the unit vector of axis, ζ.

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].

The inertial forces, $\vec{F}$, and moments of inertial forces, $\vec{M}_F$, in equations [1], [2], in the general case, can be written as a linear form of generalized accelerations and a square form of relative angular velocities

$$\vec{F} = \sum_{i=1}^{n} \vec{a}_i\ddot{\phi}_i + \sum_{i=1}^{n}\sum_{j=1}^{n} \vec{b}_{ij}\dot{\phi}_i\dot{\phi}_j$$
$$\vec{M}_F = \sum_{i=1}^{n} \vec{c}_i\ddot{\phi}_i + \sum_{i=1}^{n}\sum_{j=1}^{n} \vec{d}_{ij}\dot{\phi}_i\dot{\phi}_j \qquad (3)$$

where: $\vec{a}_i$, $\vec{b}_{ij}$, $\vec{c}_i$, $\vec{d}_{ij}$ are functions of generalized coordinates. If the prescribed $\tilde{\phi}$ coordinates are separated from the set of coordinates ϕ_i, and the ϕ^* coordinates computed from dynamic connections, the general system [1], [2] can be written in a concise form as:

$$\sum_{i=1}^{n} \vec{c}_i\ddot{\phi}_i^* + \sum_{i=1}^{n}\sum_{j=1}^{n} \vec{d}_{ij}\dot{\phi}_i^*\dot{\phi}_j^* + \vec{g} = 0 \qquad (4)$$

where: $\vec{c}_i$, $\vec{d}_{ij}$ are vector coefficients depending on $\tilde{\phi}$, $\dot{\tilde{\phi}}$, $\ddot{\tilde{\phi}}$, and $\vec{g}$ is a vector coefficient depending on $\tilde{\phi}$, $\tilde{\phi}^*$, $\tilde{\phi}$, ϕ^*. Eq. 4 with the imposed repeatability conditions [1, 2, 3]:

$$\phi_i^*(0) = \phi_i^*(T/2) \qquad (5)$$

where T is the step period, gives the compensating synergy (motion of upper part of the body) at the basis of perscribed synergy of the lower extremities (gait pattern). Since the adopted gait is symmetrical, the repeatability conditions for the coordinates ϕ^* can be written for the half-step.

The next step is to define the driving torques for the purpose of performing the anthropomorphic mechanism. To define these torques, we "break" the chain along the axes of particular joints. Thus for the part of the system not connected with the ground, the total torque of each joint can be computed according to [10]:

$$\vec{M}^k = \sum_{i=1}^{n} \vec{c}_i^{\,k}\ddot{\phi}_i + \sum_{i=1}^{n}\sum_{j=1}^{n} \vec{d}_{ij}^{\,k}\dot{\phi}_i\dot{\phi}_j + \vec{g}^k \qquad (6)$$

where: the indes, k, shows the number of the joint where the kinematic chain of the mechanism has been "broken". Since the coordinates, $\phi(t)$, are known, equation (6) makes it possible to determine all the driving torques M_i, i=1,...,n.

At a basis of the semi-inverse method for the forming of mathematical gait models, sufficiently instructive models were developed based on the adopted mechanical configuration in Figure 2. In the course of development, the algorithm for construction of the mathematical gait model acquired all the properties of a computer method, which resulted in a program package of significant flexibility consisting of the modification of the initial information block about the gait type, the geometry of members of the anthropomorphic mechanism, its masses and tensor of inertia.

SOME SIMULATION RESULTS

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

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

- S - step size (length)
- T - step period (duration), cadence
- displacement of ZMP (zero-moment point)
- overlapping % (double-support phase duration).

Both symmetrical and asymmetrical gait were treated. In that way all basic cases were taken into consideration, which should be significant for real biped system behaviour.

Due to limited space, very condensed examples of the complete simulation results will be given here. Therefore, energy analysis concerning only symmetrical gait types upon level ground will be presented. Also, symmetrical gait cases are given with single-support and double-support phases. All analyses have been performed for prescribed synergy of the lower extremities (desired type of leg motion), illustrated in Figure 3. Also for the single-support type of gait, five cases of ZMP displacement were adopted (Figure 4), while ZMP displacement for the double-support phase will be shown later. In Figures 5 to 7 are presented the corresponding values as functions of gait speed increase via the parameters T and S for various ZMP-laws. Also total mechanical work of all actuators and the maximal values of the vertical component of the reaction force (Figure 6) augment for greater S or smaller T, faster gait in both cases.

In Figures 8 to 10 are presented the corresponding values as functions of gait speed increase via the parameters T and S for various durations of the double-support phases.

In Fig. 11 is presented the compensating actuator as function of the difference of the support phase duration for the left and right leg ΔT in the case of asymmetric gait.

In Figures 12 to 14 are illustrated the corresponding values as functions of gait speed by augmenting the step size of the right leg for S=0,8 and T=1,5 sec with asymmetric gait cases. Mechanical work during full step augments for faster gait with actuators both for compensation in the sagittal and frontal planes (Figures 12).

Total mechanical work of all actuators and the maximal values of the reaction force vertical component augment for faster gait (Figures 13, 14).

CONCLUSION

The paper was intended to prepare the simulation basis for the forming of the criterion for the evaluation of biped gait. Compared with the preceding results at the basis of dynamic models of various complexity and symmetrical gait, in this paper were presented cases of asymmetrical gait, which are of practical importance. expecially in the evaluation of pathological gait. Notably interesting are the new simulation results of gait with different degrees of overlapping, which renders also the possibility of a more realistic gait evaluation. Beside the dynamical reaction forces in the foot - support surface contact, the energetic analysis, acceleration analysis of the compensating part, for a series of characteristic ZMP - patterns, a rather broad basis was created for concluding about the functional connection between the basic dynamical performances of biped gait and the relevant kinematic-dynamical parameters, which yields also the potential possibility for quantification of the mentioned performances in the scope of a criterion for the validation of biped gait.

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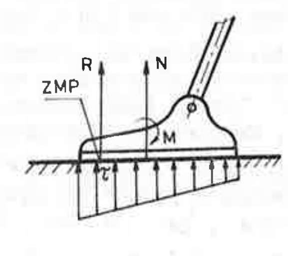


Fig. 1. Zero Moment Point (ZMP)

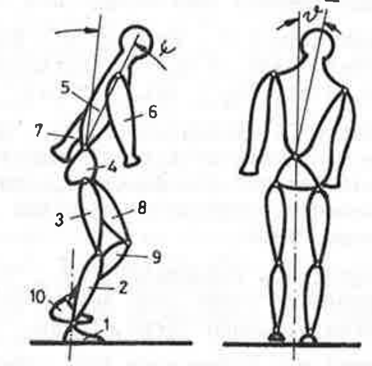


Fig. 2. Adopted Mechanical Configuration

T.1. Mass and inertia DATA

NR. of PART.	MASS $(kg\ sec^2 m^{-1})$	LENGTH (m)	Proper moments inertia $(kg\ sec^2\ m)$ J_x	J_y	J_z
1(10)	0.152	0.121	0.00006	0.00055	0.00046
2(9)	0.612	0.397	0.00815	0.00815	0.00056
3(8)	1.090	0.431	0.01705	0.01570	0.00272
4	1.222	0.182	0.02080	0.00990	0.01960
5	2.140	0.417	0.13400	0.12320	0.01950
6	0.468	0.785	0.00447	0.00447	0.00036

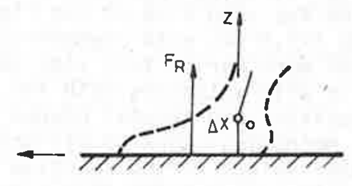
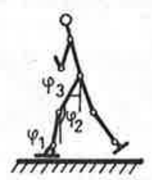


Fig. 4. ZMP displacement



Smooth level walk

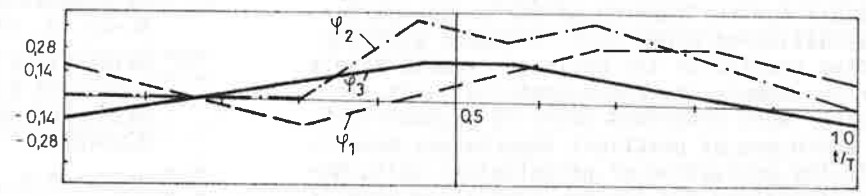


Fig. 3. Adopted Trajectories of Legs

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)

Case	t(sec)	Δx(m)
I	0.÷T/2	0,0
II	0 ÷ 0,3 0,3÷T/2	0,0 0,035
III	0 ÷ 0,5 0,5÷T/2	0,0 0,035
IV	0 ÷ 0,2 0,2÷T/2	-0,02 0,0
V	0 ÷ 0,2 0,2÷0,6 0,6÷T/2	-0,02 0,0 0,035

case I
II
III
IV
V

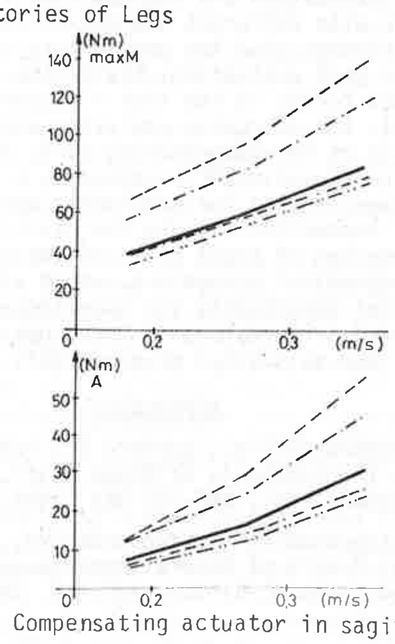


Fig. 5. Compensating actuator in sagittal plane

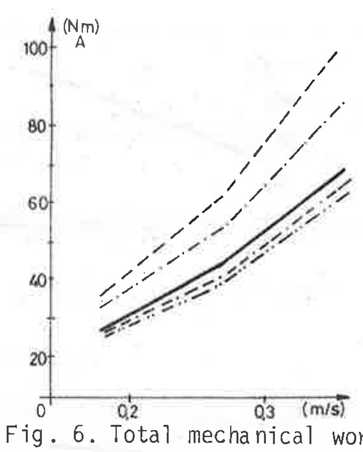


Fig. 6. Total mechanical work

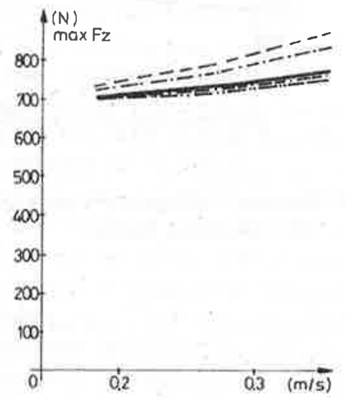


Fig. 7. Maximal values of vertical component of reaction force

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)

ZMP LAW		
Δx [m]	Δy [m]	t [sec]
0	d	0÷T/2 for p=0
0,5S	d	0÷τ(τ=Tp/400)
0	0	τ÷(T/2-τ)
-0,5S	d	(T/2-τ)÷T/2

p=20%
p=30%
p=40%
p = 0%

T: Step period sec; p: % of double support phase duration;

d: Semi-distance of feet in frontal plane

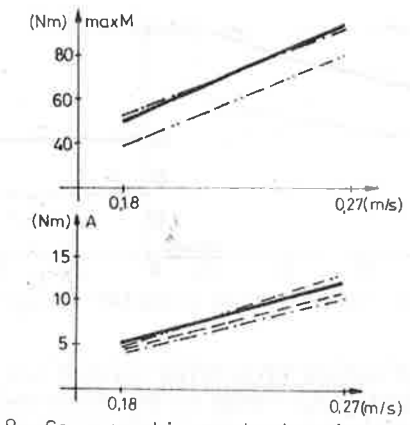


Fig. 8. Compensating actuator in sagittal plane

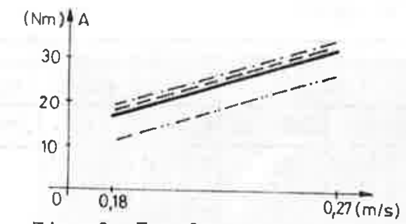


Fig. 9. Total mechanical work

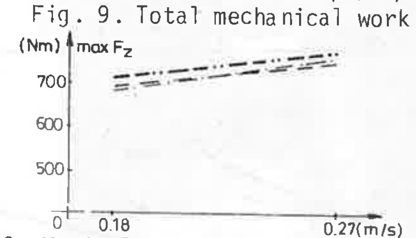


Fig. 10. Maximal values of vertical component of reaction force

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

Support phase of left leg t(sec)	S	ZMP law	Support phase of right leg t(sec)	S	ZMP law	ΔT
0-0,75	0,8	I	0.75-1,5	0,8	I	0,00
0-0,72	0,8	I	0,72-1,5	0,8	I	0,06
0-0,70	0.8	I	0,70-1,5	0,8	I	0,10
0-0,65	0.8	I	0,65-1,5	0,8	I	0,20
0-0,60	0,8	I	0,60-1,5	0,8	I	0,30
0-0,75	0.6	I	0.75-1,5	0,6	I	0,00
0-0,72	0.6	I	0,75-1.5	0,6	I	0,06
0-0,70	0,6	I	0,70-1,5	0,6	I	0,10
0-0.65	0,6	I	0,65-1,5	0,6	I	0,20
0-0.60	0,6	I	0,60-1,5	0,6	I	0,30

case: T = 1,5 S = 0,8

T = 1,5 S = 0,6

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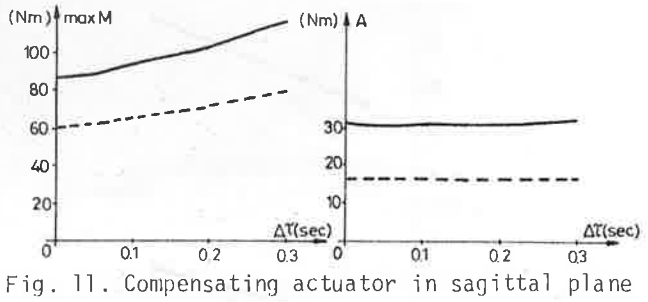

Fig. 11. Compensating actuator in sagittal plane

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)

Support phase of left leg			Support phase of right leg			ΔS
t(sec)	S	ZMP law	t(sec)	S	ZMP law	
0-0,75	0,8	I	0,75-1,5	0,80	I	0.00
0-0,75	0,8	I	0,75-1,5	0,85	I	0,05
0-0,75	0,8	I	0,75-1,5	0,90	I	0,10
0-0,75	0,8	I	0,75-1,5	1,00	I	0,20

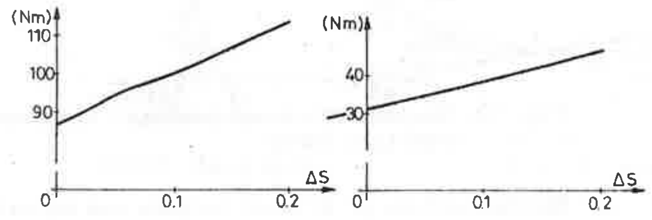

Fig. 12. Compensating actuator in sagittal plane

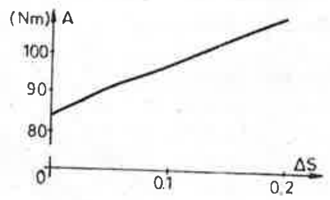

Fig. 13. Total mechanical work

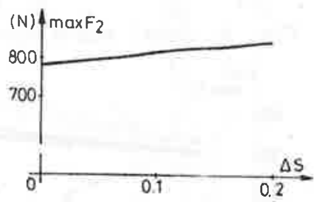

Fig. 14. Maximal values of vertical component of reaction force

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ANALYSIS AND SYNTHESIS OF PARETIC GAIT - MEASUREMENTS AND ELECTRICAL STIMULATION

Miroljub Kljajić, Matija Maležič, Uroš Stanič

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

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 consisting of patient indications and selection, qualitative medical evaluation together with measurements of the biomechanical paramethers, 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 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 out 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).

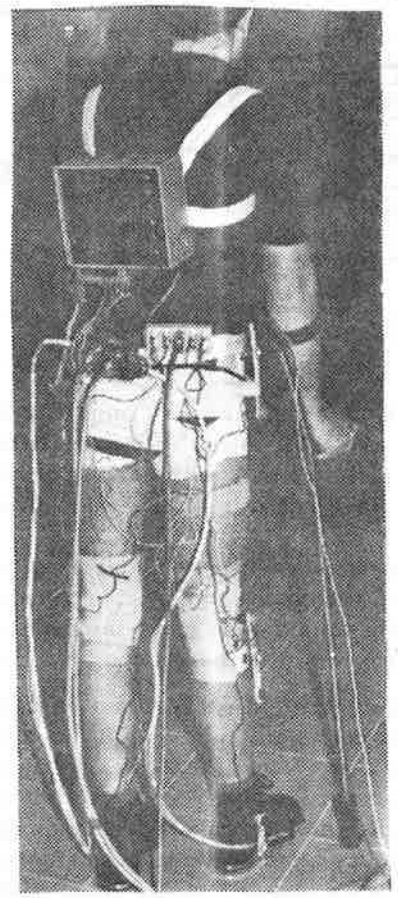

Fig.1: Paraparetic patient with six-channel stimulator, sagittal plane goniometers, force shoes , and force crutch during measurements

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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.

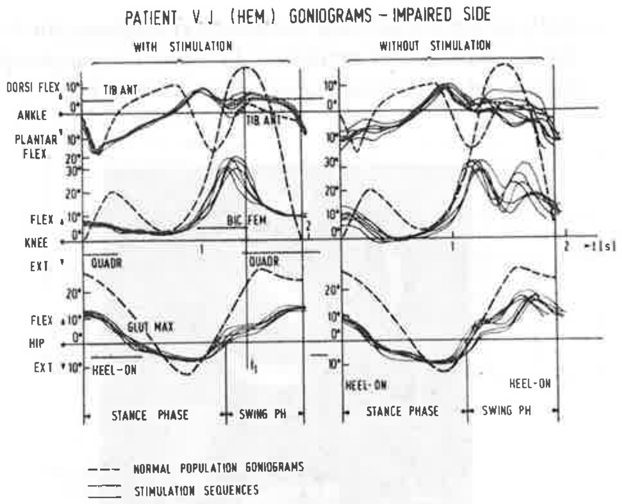

Fig.2: Goniograms of joint angles of hemiparetic gait (full lines) with and without stimulation together with normal ones (dashed line) and stimulation sequences.

With the simultaneous stimulation of m.tibialis anticus, m. biceps femoris, m.quadriceps, and m. gluteus 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 plotts 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).

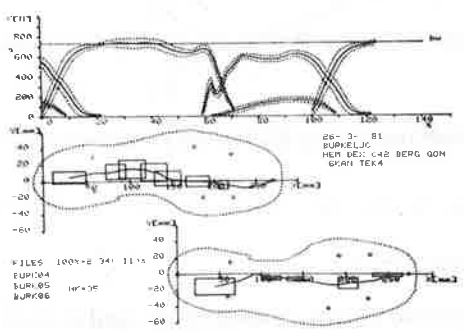

Fig.3: Ground reaction forces and their spatial distribution under the feet and crutch during gait.

Similar print-out have been designed for goniograms 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 sinchronisation (Fig.4. - bottom diagrams) (9).

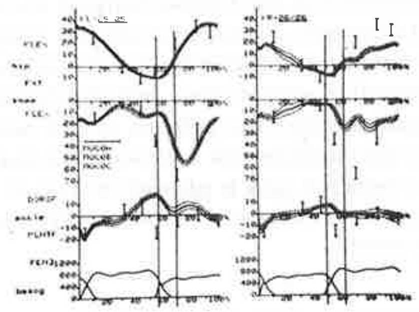

Fig.4: Goniograms of hip, knee, and ankle sinchronised by force- basograms for left and right leg.

CONCLUSION

The approach to the analysis and synthesis of paretic gait has been discussed. The developed measurement instrumentation has been evaluated as usefull 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.

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BIOCYBERNETIC ASPECT OF LOCOMOTION BY EXTERNAL CONTROL

A. Kralj [1] and R. Lehr [2]

[1]Visiting Professor, Department of Electrical Engineering and Pritzker Institute for Medical Engineering, Illinois Institute of Technology, Chicago, Illinois 60616, on leave from Edvard Kardelj University, Ljubljana, Yugoslavia.
[2]Department of Anatomy, School of Medicine, Southern Illinois University, Carbondale, Illinois 62901, USA.

INTRODUCTION

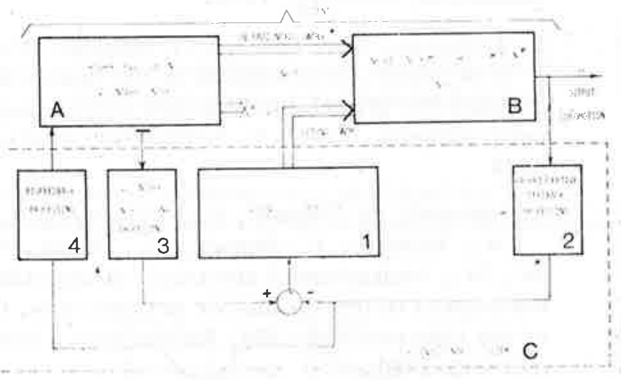
Our aim is to provide a review of the problems in external control for the restoration of locomotion. Biocybernetic aspects are introduced to highlight current achievements in the field and to pinpoint sound directions for further research work. Our emphasis will be on the severly impaired neurological patient and the demands these patients place on the design of external controls.

BIOCYBERNETIC ASPECTS

A biocybernetic approach is useful for analysing pathological neuromuscular mechanisms and defining sound possibilities for the introduction of external controls. In this paper we are going to use a systems theory approach in order to indicate the state of the art in the field of external control. In this simplified and short overview the state of the art will be given in:

- gait analysis and synthesis for
- understanding the neurologic disorders and special problems in each category of patients in regard to the
- advantageous use of the neuromuscular system mechanisms and
- the characteristics of the currently available external mechanical and functional electrical stimulation (FES) control means in gait restoration.

Systems theory demands that the system be understood before a control is applied to it. This is best exemplified for our purposes by reference to the following figure.



In this block diagram the patient is the system, and is represented by blocks A and B. The external control system is represented by block C. Block A, the higher levels of the CNS are excluded from the external control function. Those functional elements which are subject to external control are marked with an asteric (*). Theoretically if the transfer function of B is known then for any input the corresponding output can be determined. Even for a missing natural input (marked X) we can compute the required external control for obtaining a given output.

Philosophically for any system to obtain a given output we can 1) apply the needed inputs or 2) modify the internal components of that system. Our current knowledge of the CNS is such that modification of the internal components is limited at present to destructive invasion. Perhaps in the future we may be able to "rewire" the internal circuits to provide adequate control of locomotion. For the time being application of the needed inputs is more feasible. As a consequence the systems approach requires not only a knowledge of transfer function but an understanding of different patient disorders.

Patient locomotion disorders may be classified in two ways:

A) neurologic patients (hemiplegia, cerebal palsey, multiple sclerosis, spinal cord injury . . .) who have pathologically organized neural control of locomotion but intact natural structures and

B) traumatic patients (amputees, 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 less demanding group for external control. Here the goal is to substitute a biomechanical device for the missing structure and to use the patients 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 biofeedback display unit (4). It is this 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 neurologic patient can be assisted by mechanical bracing, FES, or a combination of both. From a biocybernetical 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 however limited 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

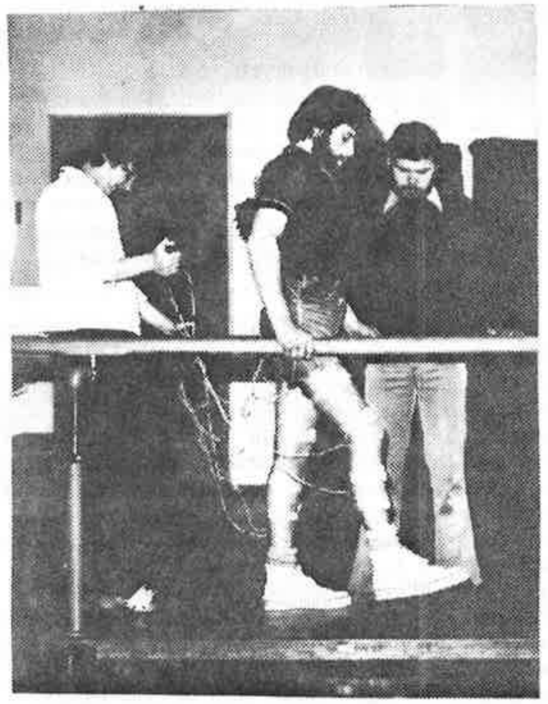
Without an arguement from anyone in the field a rather long list of task or topics to insure progress in the restoration of locomotion via external control could be produced. Given the uniqueness of each patient, the multiple disciplines involved in this complex problem and the rapidly developing technology the attempt to do so seems almost impossible. Yet in spite of this complexity the biocybernetic approach allows a definable goal to be assumed and the basis for its attainment to be persued. Remember progress has been made in spite of the complexity and it is important to site a few critical tasks of common interest to a broad group of researchers. Three come quickly to mind.

1. Understanding the biomechanical events in normal and pathological gait
2. Understanding the neurological 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 mechanicisms. The physical finds him/herself unable to use this information in such a way as to lead to clinical interpretation (9) particulary in answer to the question of why an individual patient performs a given way and the clinical reason for it. Why? Because for a given motion deficiency there may be different medical reasons responsible. For example: inadequate ankle dorsiflexion found in a pathological gait can be the result of weak dorsiflexors or because of spastic plantor flexors. But even when the clinical reason is accurately described, still 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 even less about its disorganization 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 neurologic locomotion disorder is still debatalbe, namely should the criteria be a comparison to the normal or is it 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

Biocybernetic 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.



In Fig. 6 a complete lesion T-10 paraplegic patient exerting biped gait in parallel bars is shown. The gait phases, double stance and single stance are controlled via efferent FES, while the swing phase is performed by FES afferent triggering of the flexion reflex withdrawal movement and hence the dissected spinal cord is used to synchronize and maintain the flexors contracted at the hip, knee and ankle joint. The latter is an example that the higher structures of the spinal cord can be externally controlled in our case even by the pheripheral inflow of information (2, 3). It has been demonstrated that FES is useful for restrengthening of disused muscles (3, 4) and that chronic stimulation converts fast muscle fibers to slow fibers which are less fatigable by altering their metabolism. In addition, it has been demonstrated that prolonged FES can improve and modify voluntary control and the spinal cord pattern of muscle contraction in hemiplegic patients (5). Today it is known that FES can be used for facilitation of various events in regard to neural control. Also FES can be used for spasticity and muscle tone alteration (6, 7, 8) but at present the mechanisms causing these changes are not explained clearly.

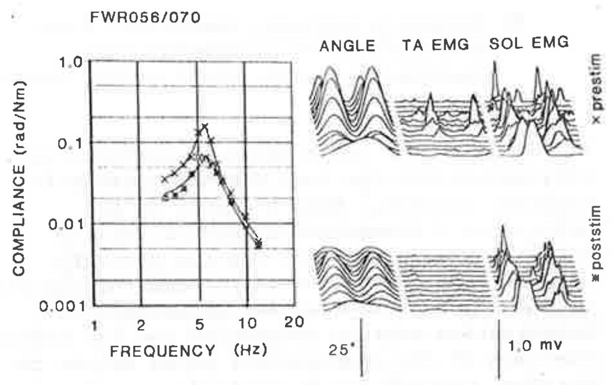


Figure 7, redrawn after (7) clearly illustrates that after only 30 minutes of FES for flexion reflex elicitation at the peroneal nerve resulted in improved joint compliance and diminished EMG in dorsal flexors. FES is an effective means for clonus reduction in patients suffering from cerebrovascular, craniocerebellar 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 excitability observable in clonus inhibition for up to 3 hours (6). The author of this work also demonstrated that FES applied afferent inflow at 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 biocybernetic aspects an interesting result showing that indeed the CNS organization can be altered and hence controlled to a large extent using different input loci. FES can also provoke therapeutic affects (10) and produce reorganization of the neuromuscular control as was shown in hemiplegic patients (5). From a biocybernetic 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 neurologic 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.

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

Kasper L.Boon, Hermie J.Hermens, Gerrit Zilvold
Technical University Twente, Rehabilitation Centre 't Roessingh, Enschede, the Netherlands.

INTRODUCTION

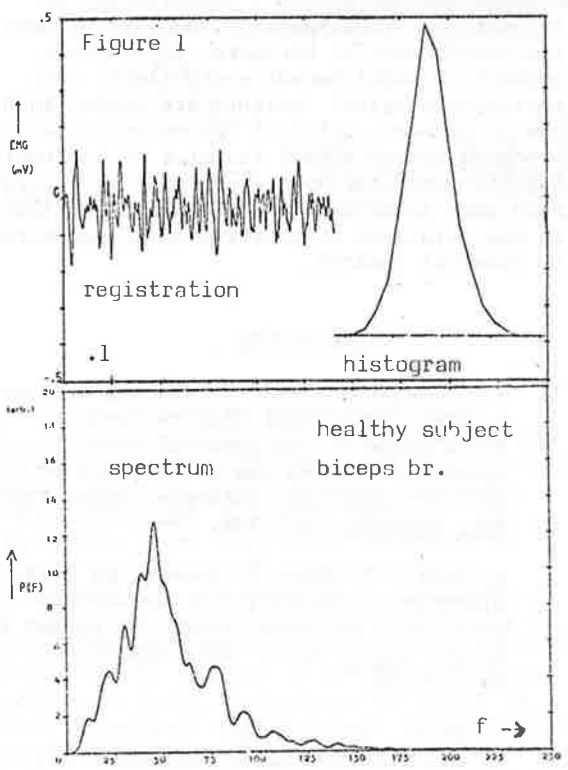
Surface EMG introduces interesting features like:
1. the pick up area of surface electrodes is relatively large so 'grosso mode' an impression of large parts of a muscle will be obtained. With many neuromuscular disorders muscles will be more or less totally involved.
2. with several neuromuscular disorders follow up investigations are necessary. It is convenient to use surface EMG in these cases because it does not introduce any physical or psychological discomfort.

It can be demonstrated that changes on the level of motor units are also detectable in surface EMG (e.g. Lindstrom, 1977).

WHICH PARAMETERS CAN BE USED

The fact that usually a noisy pattern is found is certainly one of the reasons why surface EMG is not very popular in clinical practice. The theory of signal analysis shows how such a noisy signal can be identified in terms of a probability density function (PDF) and a power spectral density function (or shortly 'spectrum'). The PDF describes the behaviour of the signal with respect to its amplitude. An estimation of the PDF is obtained by the amplitude histogram. The spectrum describes the behaviour of the signal with respect to its variability in time. It is important to notice that parameters like the number of peaks, can also be derived from such a spectrum (Papoulis, 1965).

With the amplitude histogram we have to realize that EMG is measured by means of amplifiers that are AC-coupled. This means that the mean amplitude will be 0. It can be shown statistically that surface EMG will usually lead to a Gaussian distribution function. This distribution can solely be described by the mean value and the standard deviation (SD). Since in our case the mean value is 0, we can conclude that the SD is the only parameter of interest. Unfortunately this parameter is not mentioned in the well known ISEK-report: Units, Terms and Standards in the Reporting of EMG Research.



It will be shown in Part II that with certain disorders significant deviations of the Gaussian distribution can be found. These deviations can be quantified by means of the statistical well known parameters skewness and kurtosis.
The spectrum cannot be described by means of a single parameter. The choice of the spectrum parameters is more or less arbitrarely.
We examined:
Fpmax: this parameter is the frequency at which the maximum of the spectrum is found.
First peak or bending point (Ffire): this parameter is related to a mean firing frequency.
Center of median frequency (Fcenter)
Characteristic points: F-6dB and F-10dB: these points indicate the 6dB and 10dB points in the spectrum.
Characteristic power; Prel 100: this parameter indicates the relative power (in %) in the spectrum above 100 Hz.
Furthermore parameters are investigated that are based on the parameters already introduced, viz.
Reciprocal innervation coefficient (RIC): this is the quotient between the SD of antagonist EMG activity and the SD of agonist EMG activity. This parameter is used e.g. by Prevo (1979).
EMG/force gradient (EFG): this parameter indicates the increase of EMG (indicated by the increase of SD) divided by the increase of force (in Newton).
Finally we studied also some parameters that are commonly used, viz.:
Integrated EMG (IEMG): mean absolute EMG (per second).
Mean top-top amplitude per second (Vtt).
Mean number of tops per second (Ntop)

METHODS AND MATERIALS

A subject is seated in a chair. His pronated 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 (MS6) 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 AD-converter). The spectrum is calculated by means of a modified Blackman Tukey algorithm in which a Papoulis window is used (band with 250 points).

RESULTS

During a pilot investigation the EMG activity of biceps and triceps is registrated 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 Ntop 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 Ntop, F-6dB, F-10dB, Fcentr 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. As an example table I shows some results obtained at a force of 24N.

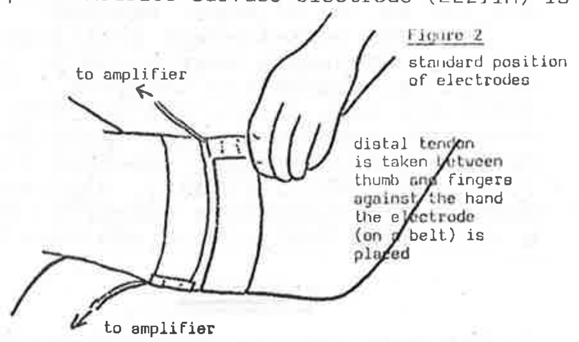


TABLE I (variability between 10 subjects)

PARAMETER	MEAN	S.DEV (in % of mean)
SD (uV)	80	27
Fcenter (Hz)	50	10
F-6dB (Hz)	70	30
F-10dB (Hz)	100	10
Prel 100 (%)	8.3	30
EFG (uV/N)	2.9	40
RIC	.14	24
Ffire (Hz)	15	20

Furthermore during 15 days every day a 'standard measurement' is performed with two healthy subjects. In this way the reproducability can be studied. The next table shows some results at a force of 32N.

TABLE II (variability at one subject)

PARAMETER	MEAN	S.DEV(in % of mean)
Ntop	197	9
SD (uV)	95	20
Ffire (Hz)	17	7
Fpmax (Hz)	45	15
Fcenter (Hz)	50	6
F-6dB (Hz)	80	6
Prel 100 (%)	6	30
RIC	.15	20
EFG (uV/N)	2.2	20

A rather linear relationship is found between IEMG, Vtt and SD. Usually SD=K.IEMG with $1<K<1.3$ (force between 8 and 32N) and SD=K.Vtt with $.6<K<.7$. Ffire is a rather independent parameter: it is related to the firing frequencies of motor units. F-6dB is considered as less important because it fluctuates to much between individuals. Ntop and Fcenter are very reproducable. Prel 100 can change significantly as will be shown in part II.

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EFFECTS OF GROUND ELECTRODE POSITION ON ELECTROMYOGRAPHIC POTENTIAL.

Jean P. Boucher and Robert J. James, Dept. of Exercise Science, University of Massachusetts, Amherst, MA, USA.

Myoelectric potentials recorded with bipolar surface electrodes are frequently utilized as a quantitative measurement of muscular activity in kinesiologic studies (Boucher & Lagasse, 1981). This past decade, several developments were achieved in the area of automatic quantification (Boucher & Lagasse, 1979), as well as in the area of electrode placement (Zuniga et al, 1970) in electromyography (EMG). Zuniga et al (1970), for example, reported that the amplitude of myoelectric potentials recorded with surface electrodes is related to the position of the exploratory electrodes over the studied muscle. These authors demonstrated a ligitimate concern regarding the skin preparation and position of the exploratory electrodes, without paying much attention to the position of the reference or ground electrode. Furthermore, the position of the ground electrode is often not reported and/or inconsistant throughout the reviewed literature.

The purpose of this study is to investigate the effects of different ground electrode positions upon the amplitude and frequency of myoelectric potentials recorded with bipolar surface electrodes.

METHODOLOGY

A total of nine college age male students participated as subjects in this study. Beckman bipolar surface electrodes (Ag/Ag-Cl) were placed over the predetermined motor point of the biceps brachii (long head) with a center to center distance of six centimeters. Two ground electrode positions were compaired: over the belly of the muscle (between the two exploratory electrodes), and a bony area (ipsilateral ulnar styloid process) (figure 1). The same EMG signal was fed into two independant amplifiers (Cyborg, J33), with two different ground electrodes (figure 1), and in turn was displayed on a dual trace storage oscilloscope.

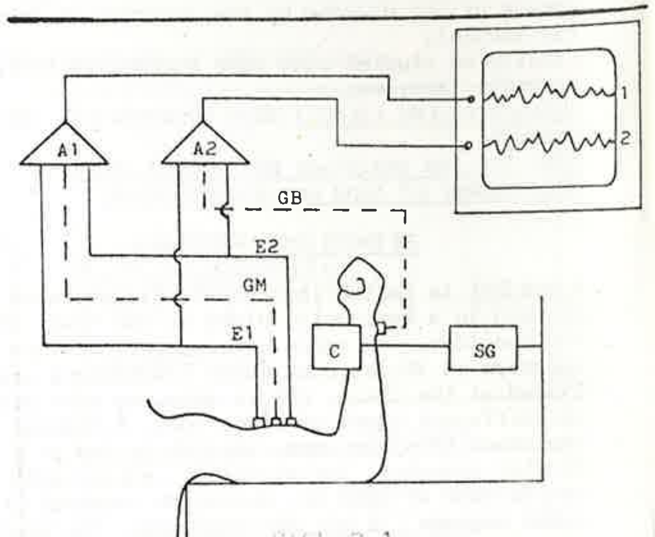

FIGURE 1

Experimental electrodesconfiguration and instruments arrangement. E1 & E2: exploratory electrodes; GM: ground electrode on the belly of the muscle; GB: ground electrode on the ulnar styloid process; A1 & A2: differential amplifiers; C: wrist cuff; SG: strain gauge.

The subjects were asked to perform four isometric contractions against a wrist cuff attached to a strain gauge (figure 1), two maximum voluntary contractions and two at 10% of their maximum.

The mean peak-to-peak amplitude and the mean frequency over a 100 ms time period was quantified (Boucher & Lagassé, 1979) for each trial. The raw EMG signals were also decomposed into their frequency components by power spectrum analysis (SPSS, SPECTRAL). Finally, all experimental conditions were compaired with a split-plot analysis of variance model.

RESULTS

The mean amplitudes and frequencies for all experimental conditions are presented in table 1. The 0.206 mean difference between bone and muscle ground electrodes, and the 0.874 mV difference between levels of muscular contraction, represent a significant increase (p .01) in EMG potential amplitude. In terms of EMG potentials frequency, the 106 Hz increase measured between ground electrode positions is significant (p .01), whereas the 14 Hz mean difference between levels of contraction failed to reach the significant level. However, the interaction between contraction levels and electrode positions represent a significant factor (p .01). Furthermore, figure 2 contrasts the spectral density distributions of the two signals (GB & GM) taken on one MVC trial. The signal with the ground electrode on the bony area (GB) represent a narrow band unimodal signal with a peak frequency of 93 Hz, whereas the second signal (GM) is a very broad band multimodal signal with a peak frequency of 120 Hz and secondary peaks at all 40 Hz harmonics.

TABLE 1

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).

	AMPLITUDE (mV)		FREQUENCY (Hz)	
	MVC	10%	MVC	10%
BONE	0.779	0.105	127	125
MUSCLE	1.184	0.111	164	196

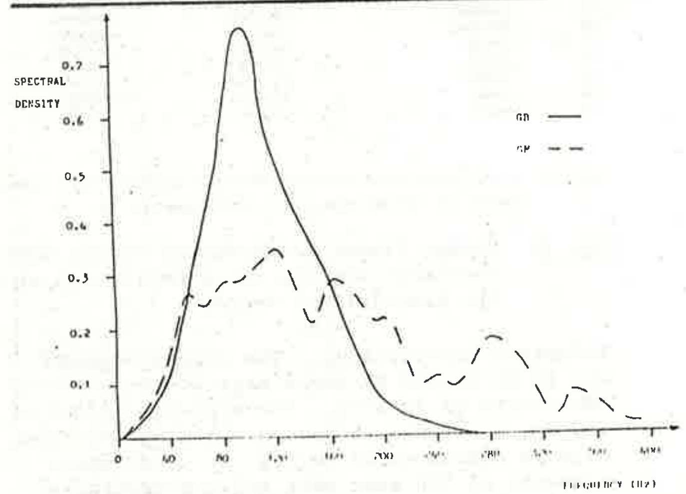

FIGURE 2

Typical spectral density distributions for the EMG signals with the ground electrodes on the bony area (GB) and on the belly of the muscle (GM).

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 Basmajian & Blumenstein's (1981) advices. These authors staded 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 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 artifact probably still exist. However, such drastic differences clearly reveal the need for ground electrode position standardization and also the need for further research on the topic of EMG modalities.

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SURFACE EMG SPECTRAL ANALYSIS AND ITS APPLICATION TO DIAGNOSTIC CLASSIFICATION

G.F. Inbar and A.E. Noujaim, Department of Electrical Engineering and the Julius Silver Institute of

Biomedical Engineering, Technion - Israel Institute of Technology, Haifa, Israel.

ABSTRACT

The present work deals with the spectral characterization of the Surface EMG signal and its application to diagnostic classification. Traditionally such data is from subcutaneous needle electrodes which may have to be inserted several times to obtain the desired signals. This clinically painful procedure was bypassed through the use of appropriate surface electrodes.

Pinelli [1977] showed that the measurement of MUAP (Motor Unit Action Potentials) parameters (shape, rate, and phase) has great clinical interest. Lindstrom [1970, 1974, 1977] and Agarwal & Gottlieb [1975] have shown that the Surface EMG signal consists of two major components. The first component is the firing rates of the MU (Motor Units) that contribute the low frequency components below 40 Hz. The second component is the resulting frequency spectra of the MUAP that contribute the high frequency components above 20 Hz. The frequency band of the spectrum is dependent upon the electrodes' geometry. Therefore it is clear that under appropriate surface measurement, the MUAP parameters are present in the spectrum.

Surface EMG was recorded from the biceps with fixed muscle length at 50% MVC (Maximal

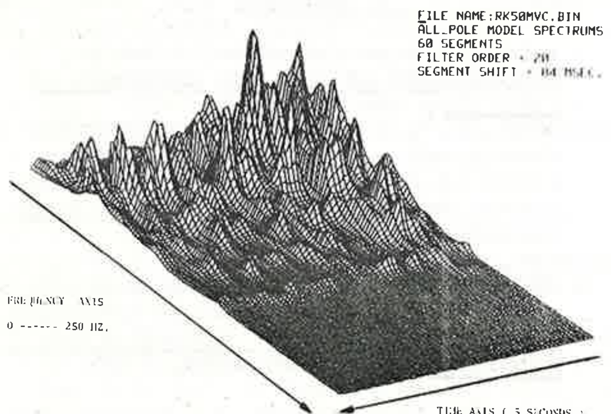

FIG 1: SIXTY ALL-POLE SPECTRUMS (20 COEFFICIENTS) AS A FUNCTION OF TIME

Fig. 1: Sixty all-pole spectrums (20 coefficients) as a function of time.

ABNORMAL GROUP THAT PARTICIPATED IN BUILDING THE DISCRIMINANT VECTOR :

1) KY (R)
2) GM
3) IA (R)
4) IA (L)
5) KSY

NORMAL GROUP THAT PARTICIPATED IN BUILDING THE DISCRIMINANT VECTOR :

1) RK
2) PS
3) JA
4) YK
5) EG

DISCRIMINANT RESULTS FOR THOSE WHO PARTICIPATED IN DISCRIMINANT VECTOR CONSTRUCTION :

ABNORMAL GROUP : 1) KY (R) = - 3054258
2) GM = - 3054244
3) IA(R) = - 3054251
4) IA(L) = - 3054252
5) KSY = - 3054244

NORMAL GROUP : 1) RK = 773553
2) PS = 773556
3) JA = 773559
4) YK = 773557
5) EG = 773549

THRESHOLD PROJECTION OF THE ABOVE TWO GROUPS IS THEREFORE = - 1140352

DISCRIMINANT RESULTS FOR "STRANGERS" :

NAME	NEEDLE DIAGNOSIS	PROJECTION RESULT	DISCRIM. DIAGNOSIS	
JY	ABNORMAL	-13987318	ABNORMAL	
SR	ABNORMAL	-4992499	ABNORMAL	
HP	ABNORMAL	-15685375	ABNORMAL	
KY (L)	NORMAL	9436016	ABNORMAL	X
SI	NORMAL	2339655	NORMAL	
YY	NORMAL	1518450	NORMAL	
MR	NORMAL	6585276	NORMAL	
ZR	NORMAL	-550181	NORMAL	

FIG. 2 : FISHER LINEAR DISCRIMINANT RESULT WITH 5 ABNORMALS AND 5 NORMALS SHARING THE CONSTRUCTION OF THE DISCRIMINANT VECTOR.

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

Voluntary Contraction). The signal bandpass 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.

The "Remnant" signal was used to help find the optimal filter order because the normalized error test $(1 - V_{p+1}/V_p)$ 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 give 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 of 500 msec. After 10 segments it was found that the coefficient average started to converge. 10 segments were used to identify a record.

For classification, the Fisher Linear Discriminant technique was used for the 20×10 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 5 normals and 5 abnormals sharing the construction of the Fisher Linear Discriminant Vector, 87.5% classification was achieved on a sample of 3 abnormals and 5 normals using clinical needle electrode diagnostic classification as a reference. This result is demonstrated in Fig. 2.

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TIME AND FREQUENCY DOMAIN ANALYSIS OF EMG PATTERNS AT INCREASING FORCE LEVELS.

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

* Centro di Teoria dei Sistemi, Dipartimento di Elettronica del Politecnico di Milano
** U.I.L.D.M. Sezione di Milano
*** Clinica Neurologica II dell'Università di Milano, Milano, Italy

INTRODUCTION

On the basis of anatomo-pathological 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 plyphasic potentials of reduced amplitude in early stages of neurogenic atrophies 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 first 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 20-30-40% of the maximum voluntary effort in the case of myopathic and peripheral neuropathic 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 CNEMG findings each muscle has been assigned to one of the following two Groups:
i) myopathic muscles (5)
ii) peripheral neurogenic muscles (9)
The test was carreid 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 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 analogic equalized filter and then sampled at 5000 Hz.
The recorded EMGs have been processed by a PDP 11/34 minicomputer provided with four disc units (two of 10 Mbyte and two of 5 Mbyte), 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 Mbyte discs, each signal occupying 192 blocks of 512 byte each, approximately corresponding to 10s temporal epoch while the programs are resident on the 10 Mbyte-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 800 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 FFT algorithm with decimation in time. A cosine window has been adopted. The mean value and standard deviation have been calculated for the following parameters of the Power Spectrum, over 48 blocks of 1024 points:
. mean frequency
. median frequency
. skewness
. kurtosis
The mean frequency has been computed following the recommendation of ISEK 4th Int. Cong. (Boston, 1979) by means of the formula :

$$f_{mean} = \frac{\int_0^{\infty} f S(f)\, df}{\int_0^{\infty} S(f)\, df}$$

where S(f) is the Power Spectrum. The median frequency is the frequency at which the Spectrum is divided into two regions with equal power: it has been computed according to the formula :

TABLE I: TIME DOMAIN PARAMETERS

SUBJECT	Pathology	ZEROCROSSINGS		POSITIVE MAXIMA /T/		MEAN AMPLITUDE /A/		T/A RATIO	
		Δ_{32}	Δ_{43}	Δ_{32}	Δ_{43}	Δ_{32}	Δ_{43}	Δ_{32}	Δ_{43}
B.C.	M	- 8.30	10.40	- 8.40	6.40	34.40	22.60	- 0.24	0.00
M.S.	M	3.70	- 2.90	2.50	- 6.70	80.30	128.40	- 0.32	- 0.16
M.C.	M								0.02
R.G.	M	72.20	-50.40	43.60	-28.50	145.20	51.10	- 0.06	- 0.16
T.N.	M	29.40	- 2.90	15.90	- 3.50	124.50	70.50	0.00	- 0.02
C.G.	PN	4.60	19.70	9.00	6.90	91.60	122.60	0.00	- 0.14
D.M.	PN	27.80	- 1.20	18.90	- 3.00	-251.90	255.40	0.21	- 0.17
D.E.	PN	5.80	- 4.90	- 4.70	- 4.30	10.30	67.70	- 0.15	- 0.25
L.E.	PN	-11.60	2.10	-10.30	2.00	182.60	59.00	- 0.15	- 0.01
M.A.	PN	19.10	17.00	11.20	6.30	183.00	167.90	- 0.02	- 0.01
P.M.	PN	20.50	17.40	8.70	6.40	98.50	455.50	0.00	- 0.05
N.R.	PN	1.50	5.50	8.50	8.00	20.20	- 0.40	0.01	0.05
S.I.	PN	1.10	22.80	4.10	13.50	60.00	119.50	- 0.10	- 0.06
T.S.	PN	24.70	7.30	27.50	7.00	124.40	24.40	0.01	0.01

$$\int_0^{f_{med}} S(f)df = \int_{f_{med}}^{\infty} S(f)df$$

where S(f) is the Power Spectrum

RESULTS and DISCUSSION

All parameters have been computed for the EMG signals measured at each of the 3 contraction levels: 20%, 30% and 40% of the maximum. Changes of their values from 20% to 30% and from 30% to 40% are given as Δ_{32} and Δ_{43} respectively . Pathology M refers to myopathic subjects where as pathology PN refers to peripheral neuropathic 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 than in neuropathic ones, whereas at a further increase from 30% to 40%,recruitment cannot take place in the myopathic muscle while it occurs in the neuropathic muscles almost to the same extent as before.These phenomena are evident in the behaviour of the zerocrossings and the positive maxima increments Δ_{32} and Δ_{43} .
Less indicative is the behaviour of the mean amplitude increments and the T/A ratio.

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 Δ_{32} and Δ_{43} increments are generally within 20% of the 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 force exerted by the contracting muscle does not influence the shape of the Power Spectrum,provided that the effort is kept between 20% and 40% of the maximum.

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FREQUENCY DOMAIN ANALYSIS OF EMG SIGNALS IN THE FOLLOW-UP OF NEUROGENIC PATIENTS

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

* Centro di Bioingegneria del Politecnico di Milano
** Clinica Neurologica II dell'Università di Milano
*** U.I.L.D.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 therapeutical treatment in the course of time especially when the development of the pathological conditions may be difficult to predict, as in hereditary and paraneoplastic 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 (Maranzana 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 increased number of giant potentials and decreases with increased 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; Maranzana Figini 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 MUAPTs. 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 (Petajan, 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

Mus pool composition	Skewness of the Power Spectrum
20 N , 0 GP , 0 PNP	5.45 ± .291
18 N , 0 GP , 1 PNP	5.26 ± .278
14 N , 0 GP , 3 PNP	4.83 ± .251
8 N , 0 GP , 6 PNP	4.45 ± .234
2 N , 1 GP , 7 PNP	4.93 ± .370
2 N , 3 GP , 5 PNP	6.77 ± .547
0 N , 6 GP , 2 PNP	9.91 ± .701
0 N , 7 GP , 0 PNP	11.67 ± .710

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 MS6. 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 2kHz cut-off frequency to prevent aliasing phenomena. The Fourier Transform has been computed by means of an FFT algorithm with decimation in time. A cosine window has been adopted. The mean value and standard deviation have been calculated over 48 blocks of 1024 points for the skewness of the Power Spectrum.

RESULTS AND DISCUSSION

The results are reported in Table II. For the

Table II: Values of the Skewness for the Guillaim-Barré patients in the follow-up program

Patient	I Examination	II Examination	III Examination
G.A. (age 42)	3.60 + 0.65 (6.10.80)	4.23 + 1.74 (10.21.80)	11.5 + 2.76 (4.30.81)
T.S. (age 17)	4.78 + 1.16 (5.22.81)	6.38 + 2.04 (10.29.81)	
D.M. (age 43)	3.59 + 0.75 (6.23.81)	10.5 + 8.89 (10.29.81)	
P.M. (age 27)	?25.1 + 6.42? (3.5.81)	2.77 + 1.70 (10.29.81)	2.62 + 0.92 (2.25.82)

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 unsteadyness of the Contraction and requires further investigations.
For Patient P.M. the first EMG test was performed some weeks after the onset of the desease. 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.

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DATA COMPRESSION STRATEGIES APPLIED TO EMG PATTERNS.

M. Maranzana Figini, ISEK member, R. Molinari and G. Sommariva, Students,

Centro di Teoria dei Sistemi, Dipartimento di Elettronica del Politecnico di Milano, Milan, Italy

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:

i) allow allocation of the muscle examined to a pathological class,
ii) preserve all information necessary to obtain from the reconstructed EMG signal the Time and Frequency Domain parameters which are at present of 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 digitation is regarded as the output of a linear filter whose input is the white noise (Fig. 1).
The problem is then a classical identification problem: a model must be determined which is able to describe the behaviour of the system (in this case the neuromuscular 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:

i) to select a class of models
ii) 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 AutoRegressive (AR) and the AutoRegressive-Moving-Average (ARMA).
The AR(n) model describes the process in the following way:

$$y(t) = a_1 y(t-1)+a_2 y(t-2)+...+a_n y(t-n)+e(t)$$

with:

$a_1, a_2, ..., a_n$	AR parameters
n	model order
$e(t)$	white noise

The ARMA (p,q) model describes the process in the following way :

$$y(t) = a_1 y(t-1)+... a_p y(t-p)+e(t)+ c_1 e(t-1)+... c_q e(t-q)$$

with:

$a_1, ... a_p$; $c_1, ... c_q$	AR and MA parameters
p ; q	model order
$e(t)$	white noise

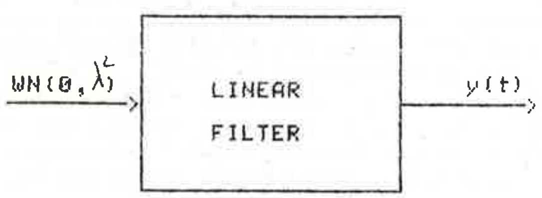

Fig.1

RESULTS and DISCUSSION

i) 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 Fabbro, 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.

ii) Reconstruction of the EMG signals
Diagnostic 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(Akaike, 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 (Amstrom, 1979). After diagnostic checking of the prediction error (Anderson and Portmanteau 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.

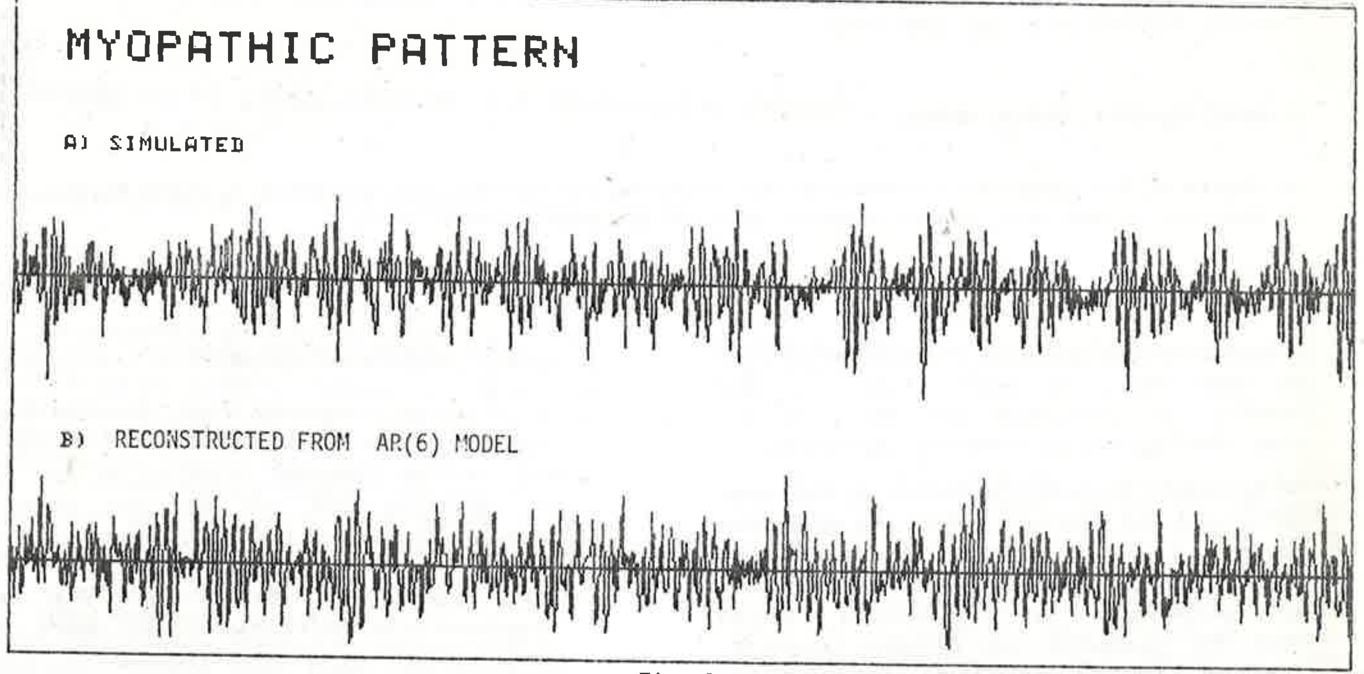

Fig. 2

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ACKNOWLEDGEMENT

The authors are deeply indebted to Sergio Bittanti, Centro Teoria Sistemi CNR, Milano, for helpful discussion and challanging suggestions.

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COMPUTER SIMULATION OF SKI JUMP (ABST)

NUNNARI GIUSEPPE, RODANO RENATO

1. Centro di Bioingegneria - Fondazione Pro Juventute Don Gnocchi, Via Gozzadini 7, 20148 Milano
2. Politecnico diMilano, Piazza Leonardo da Vinci 32, 20133 Milano, Italy

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 Bioingegneria 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 aquipment adopted and the take-off events.

An useful mathematical model, implemented on a PDP 34/II computer, describing the ski jumper has been developed.

The model consists of two main parts:
- the first analyzes the inrunning phase;
- the second analyzes the airborne and landing phases.

The first part allows to compute the velocity of the athlete during the inrunning 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 initial velocity, the dynamical parameter of the athlete and the geometrical parameters of the outline landing are assigned.

The initial velocity of the airborne phase has been obtained with the vectorial sum of the computed velocity at the end of inrunning and the measured velocity produced during the take-off.

The velocity of the take-off has been measured with a force plate where the athletes performed many take-off utilizing an useful equipment, able to provide sperimental condition similar to the real one.

These results have been compared to those obtained by other authors through cinematographic measurements. The validity of the model is confirmed by the agreement between the experimental and computed results. Therefore the model seems to be useful to evaluate the influence of different parameters on the final performance.

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ANTICIPATORY EMG ACTIVITIES RELATED TO A VOLUNTARY MOVEMENT

S. BOUISSET and M. ZATTARA, Laboratoire de Physiologie du Mouvement,

Université de Paris-Sud, F-91405 ORSAY - Cedex, France

The existence of postural activity prior to and during a movement was established in Man by BELENKII 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 (ZATTARA et BOUISSET,1980; BOUISSET et ZATTARA, 1981).

Movements of antepulsion-flexion of the upper limbs have been investigated according to three conditions: unilateral flexions with no additional inertia (OUF), with an additional inertia (IUF) and bilateral flexions(BF). The dynamic asymmetry of the movements increased from BF to OUF and to IUF.

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 (BRENIERE et al., 1981). Accelerometers fixed on splints were bound to the wrist of the moving upper limb and to various body segments. This made it possible to measure the tangential acceleration of the arm (AW_i) and to determine the onset and the sign of the anticipatory antero-posterior local accelerations. The activity of the Anterior Deltoïdeus (AD) and of the main muscles of the lower limbs, pelvis, trunk and scapular girdle, on the ipsi-(i) and contra-(c) lateral sides, were recorded by surface electromyography (EMG). The instruction 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 AD_i and/or AW_i were recorded. The other muscles and accelerations were recorded by rotation in order to make all the relevant comparisons possible. The onset of the activity of AD_i and the onset of AW_i were used respectively as time origins for the EMG and biomechanical activities.

RESULTS

Before the activation of the AD_i, a sequence of inhibitions (-) and excitations (+) concerned muscles of lower limbs, pelvis and trunk. These anticipatory EMG activities were organized according to a pattern reproductible 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 $Soleus_i$ (-), Tensor Facia $Latae_c$(+)/ Rectus $Femoris_c$(+), $Semitendinosus_i$(+)/Gluteus $Maximus_i$(+), Erectores $Spinae_c$ (+); 2) for the BF, anticipatory EMG only concerned the pairs of Soleus, Semitendinosus, Gluteus Maximus and Erectores Spinae; 3) for both types of movement the anteposition of the EMG activities of the lower limbs and pelvis increased with the dynamic asymmetry of the forthcoming movement.

The onset of AW_i was also preceded by anticipatory acceleration at the general level as well as at the local one. The shape of these anticipatory accelerations was reproductible for one subject and consistent from one subject to another. The body's center of gravity was always subjected to a forward 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 ones; 2) for BF,identity was the rule; 3) the anteposition 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 volun-

S Buisset AnticipatoryEMG Activities Related to Voluntary Movement

tary 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 UF, 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 in the rest 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 reproductibility 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 preprogrammed, a result which is in accordance with BELENKII et al. (1967) and with WEISS and HAYES (1979 and personal communication). also.

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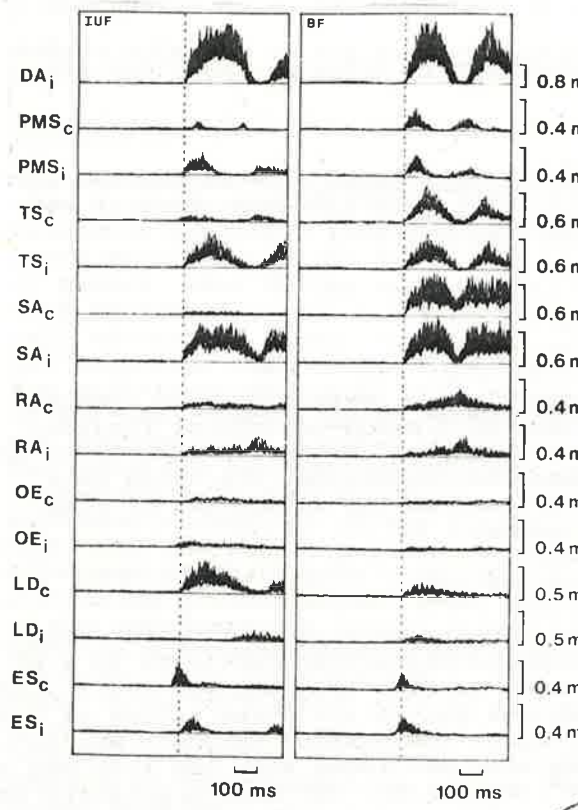

Fig. 1 EMG activities of the main muscles of the trunk and the scapular girdle for unilateral flexions with additional inertia (IUF) and for bilateral flexions (BF).

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

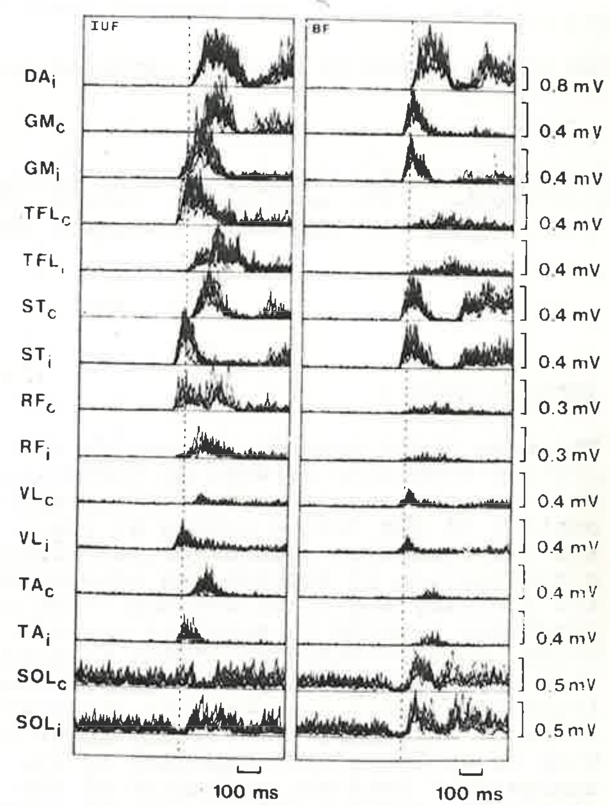

Fig. 2 EMG activities of the main muscles of the lower limbs and pelvis for unilateral flexions with additional inertia (IUF) and bilateral flexions (BF).

As for figure 1, the figure represents, for one subject, the rectified and smoothed EMG activities of five trials, superposed by synchronising the records on the onset of the Anterior Deltoïdeus (DA_i). From bottom to top: Soleus (SOL), Tibialis Anterior (TA), Vastus Lateralis (VL), Rectus Femoris (RF), Semitendinosus (ST), Tensor Facia Latae (TFL), Gluteus Maximus (GM).

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THE RELATIONSHIP OF APPLIED EXTERNAL MOMENTS AND MYOELECTRIC ACTIVITY ABOUT THE KNEE.

T.P. Andriacchi[1], G.B.J. Andersson[2], R. Örtengren[3], and R. Mikosz[1]. [1]Dept. Orthop. Surg., Rush-Presb. - St. Luke's Med. Center, Chicago, Ill., USA; [2]Dept. Orthop. Surg. I and [3]Dept. Clin. Neurophysiol., Sahlgren Hospital, Göteborg, Sweden.

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 antagonistic 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 antagonistic 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, .05 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 kneegoniometer 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, TL, 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 VL) 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 considered to be extensors of the knee when 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.

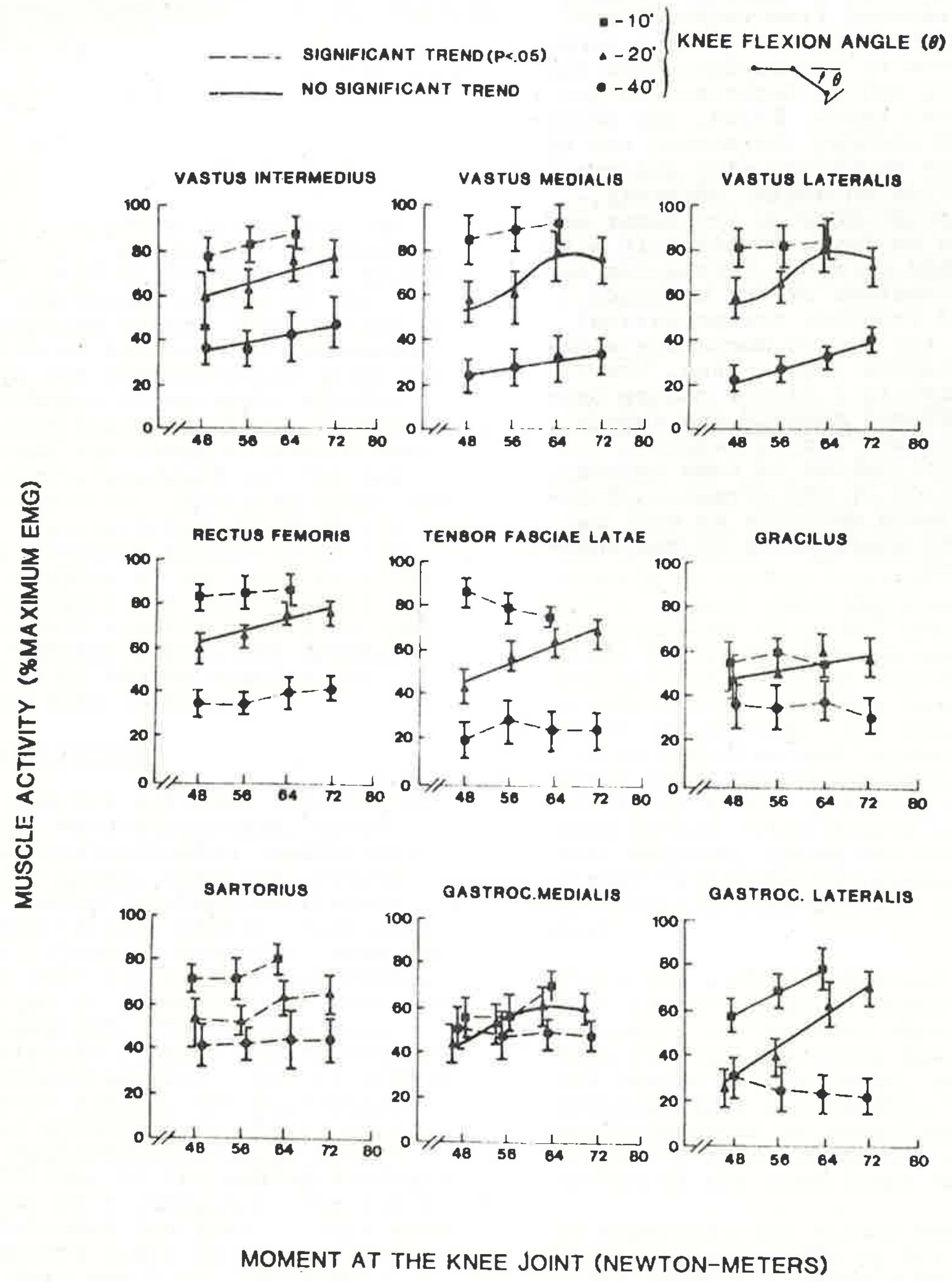
As expected, the flexor muscle groups primarily responded to the external loads tending to extend the joint. In general, the knee flexors appear more efficient at 10° of flexion than at higher flexion angles. The hamstring muscles showed a significant increase in

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 phenomena 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 antagonistic muscle activity as well as changes in the moment arms of the individual muscles.

MUSCLE ACTIVITY VS. EXTERNAL FLEXION MOMENT AT KNEE



SKELETAL MUSCLE FORCE, PRESSURE AND MYOELECTRIC SIGNAL

P. Parker, L. Körner, C. Almström, G. Andersson, U. Johnsson, P. Herberts, R. Kadefors
Chalmers University of Technology, Dept. of Applied Electronics; Project Lindholmen AB; East Hospital, Dept. of Orthopaedic Surgery; Göteborg, Sweden.

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 (Kadefors, 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 fourty-five years who volunteered for the investigation. A pressure catheter was introduced into the biceps brachii muscle under local anesthesia. In seven subjects wick-catheters 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 spcial 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 CRO. 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-one 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 70 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 0.962 (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 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, the average correlation coefficient during five minutes of loading amounting to for load/pressure 0.971 (SD = 0.02) and 0.916 (SD = 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 limits 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.

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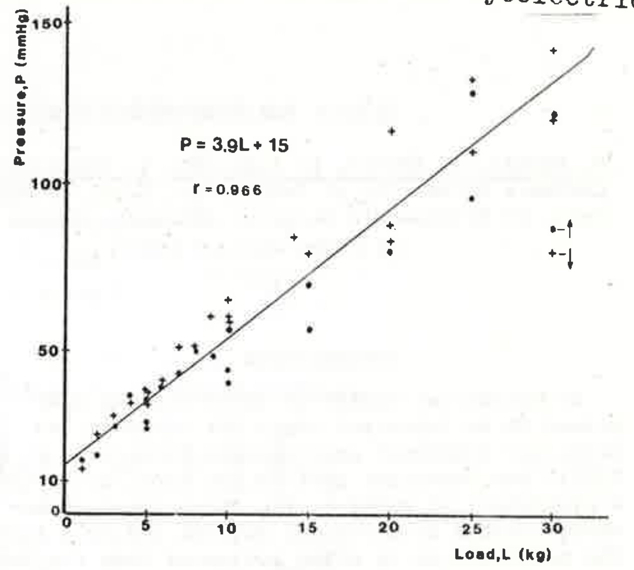

Figure 1 Pressure v.s. Load

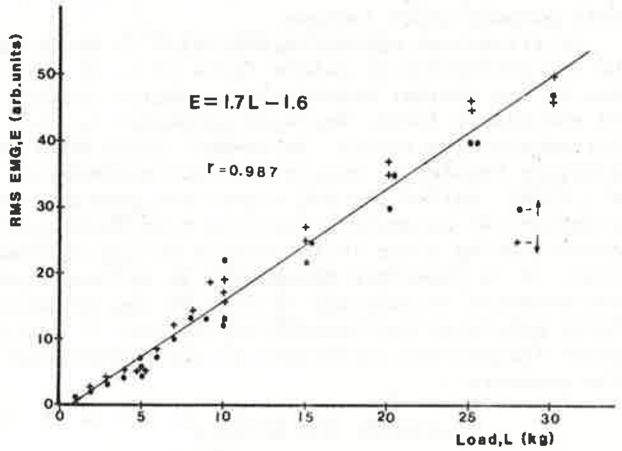

Figure 2 EMG v.s. Load

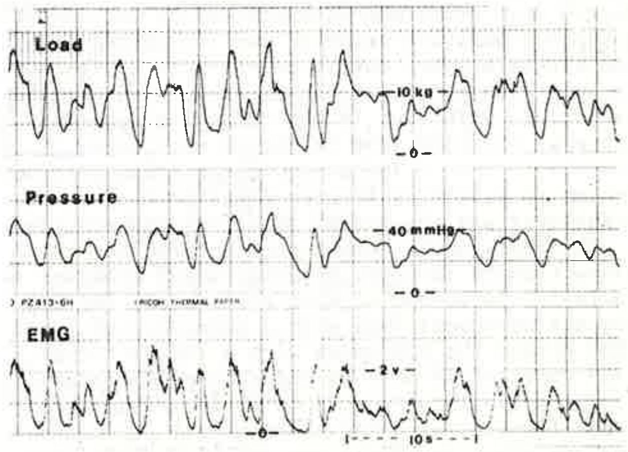

Figure 3 Dynamic Data

TRANQUILIZER EFFECT OF BICYCLING AND RUNNING (ABST)

Marcel Nadeau, Jean-Yves Roy, Laboratoire de motricite humaine, Departement de kinanthropologie, FEPS,

Universite de Sherbrooke, Sherbrooke, Quebec, Canada

The H/M ratio is an objective measure of motoneuronal excitability and de Vries et al (1981) showed that the maximal H/M ratio of the calf muscles decreased by a moderate type of exercise on a bicycle ergometer. Since the calf muscles are much more stressed while running them while bicycling, the purpose of this study was to evaluate the changes in the maximal H/M ratio of the soleus and gastrocnemius muscles following an exercise of the same intensity and duration on a bicycle ergometer and on a treadmill. Ten male subjects, from 19 to 23 years of age, participated in the study. Exercise intensity was adjusted in order to require a heart rate of 120 beats per minute on both ergometers and was maintained for a period of 20 minutes. Oxygen consumption was also measured during the 10th and the 20th minute. The maximal soleus and gastronemius H/M ratio was evaluated before and 10 minutes after the exercise bouts. A three-way analysis of variance with repeated measures on two factors (ergometer, time) was performed in order to test the statistical differences of the distributions.

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RELATIONSHIP BETWEEN KINEMATIC MUSCLE FORCES AND EMG (ABST)

U.P. Wyss, V.A. Pollak, Division of Biomedical Engineering, University of Saskatchewan, Saskatoon

Saskatchewan, S7N OWO, CANADA

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 poisition was chosen for this study as only a few muscles are 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.

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

J.A.RINGELBERG, 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? (Pancherz and Winnberg, 1981)
2. In what plane should the abduction be performed? Inman et al. (1944) did their investigations on abduction in the frontal plane. Johnston (1937) found the scapular plane to be the more physiological one. Freedman and Munro (1966) investigated abduction in the scapular plane in 61 persons. They stated that the scapula, humerus and attachments of humeroscapular muscles all lie 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.

I. MATERIALS AND METHODS

Surface EMG recordings were obtained from four healthy men who were subjected to the protocol on three different days (see TABLE 1-3 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 taperecorder. Later on raw EMG signals were analysed 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.

Muscles	Electrode placement
Deltoid anterior part	at 90°abduction: along line a.c.junction-medial epicondyle
middle part	0°:along line mid-acromion-lateral epicondyle
posterior part	at 90°abduction: along line posterior part acromion-lateral epicondyle
Infraspinatus	below a.c.joint and medial from axilla
Pectoralis major clavicular part	below midclavicular

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

Abduction Angle	30°	45°	60°	75°	90°
Frontal plane	xx	xx	xx	xx	xx
Scapular plane	xx	xx	xx	xx	xx
Frontal plane + 1 kg	xx	xx	xx	xx	xx
Scapular plane + 1 kg	xx	xx	xx	xx	xx
3x MVC frontal plane	-	-	-	-	xxx
3x MVC scapular plane	-	-	-	-	xxx

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 5 kHz-90 dB
- -taperecorder- 5 kHz - 15 inch/sec
- -sampling rate A/D conversion 5 kHz

II. RESULTS

In figure 1-4 the mean EMG values (n=4) with S.D. are presented as a percentage of the EMG 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 x. The signal/noise ratio was insufficient to quantify this EMG. The

MVC was 9.99 kg (mean) $\pm$ 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 anatomical position, it will have a more comfortable situation in scapular plane abduction because of its origin in the supraspinatus fossa.

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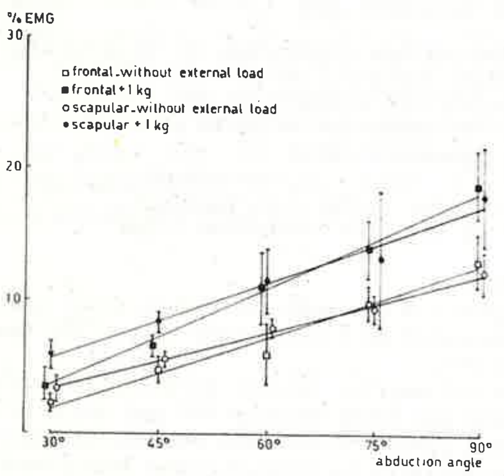

Fig. 1. EMG of the deltoid muscle - anterior part - in different isometric positions of abduction.

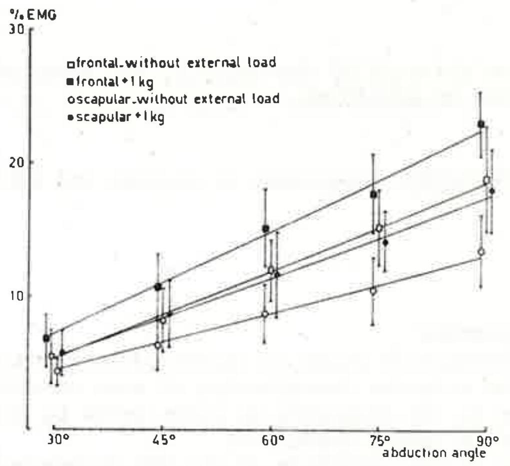

Fig. 2. EMG of the deltoid muscle - middle part.

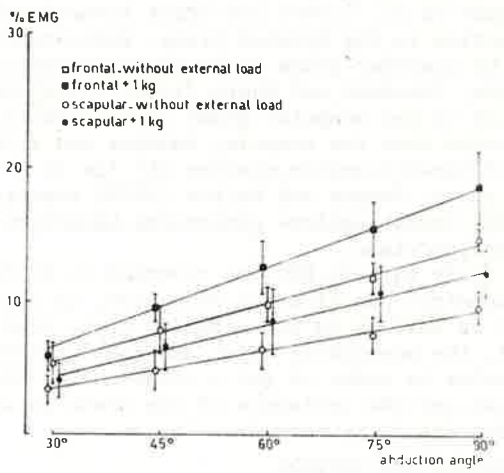

Fig. 3. EMG of the deltoid muscle - posterior part

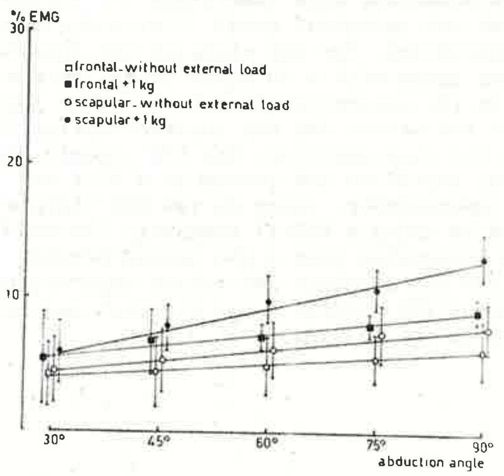

Fig. 4. EMG of the infraspinatus muscle.

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ANALYSIS OF FORCE TRACKING AND EMG INDICATING SENSORIMOTOR PERFORMANCE

Günter Rau, Harald Reucher, Helmholtz-Institute for Biomedical Engineering,

Aachen, West Germany

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, kinesthesis 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 wrist which was exerted in a set-up indicated by Fig. 1. By a force 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.

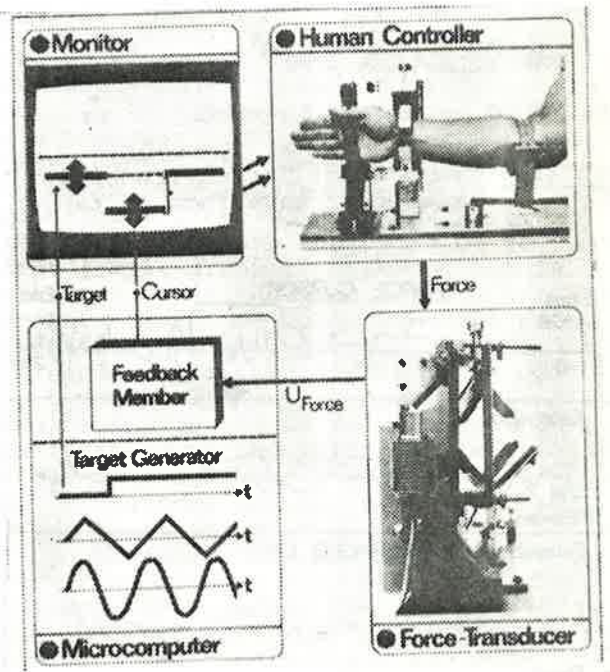

Fig. 1: Target Tracking Task: Experimental Set-Up

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 target was the zero torque level. The advantage of the critical tracking task is to detect the individual performance limit of any subject by increasing the difficulty of the task.

A review of tracking tasks and a comparable set-up was described by Stein and Pioch /2/ for an isometric control element. The position of the target and the cursor as well as the force have been recorded. In addition, the EMG activities are picked up and processed (rectified and averaged). Typical records of target tracking and critical tracking experiments are shown in Fig. 2 a, b.

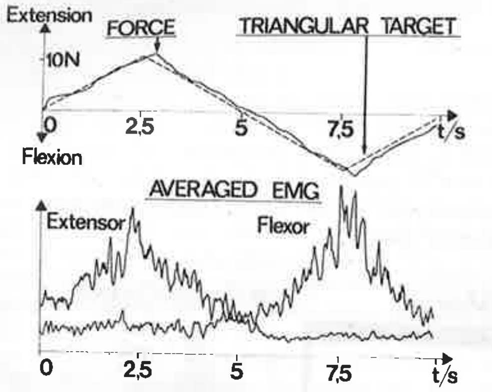

Fig. 2a: Example for Target Tracking: triangular wave form of target position

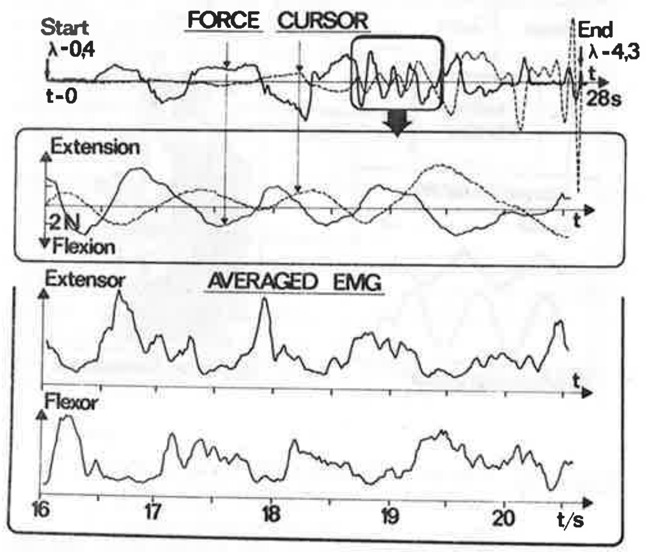

Fig. 2b: Critical Tracking: Force and Averaged EMG

DATA ANALYSIS

a) Target tracking task: In the example of Fig. 2 a, a triangular function of low frequency (0.1 Hz) has been followed. Since the averaged EMG is in good approximation proportional to force one 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 excertion when amplified selectively; this higher frequency components increased at higher levels of force. The basic activity indicates a level of cocontraction in both muscle groups. At these low frequencies a prediction by the human operator seems to play an inferior role.

b) Critical tracking task: A typical example for a critical tracking task experiment is shown in Fig. 2 b. In the upper trace the deviation of the cursor is continously increasing in time for a given value of force, i.e. very small changes in force caused a much larger deviation of the cursor towards the end of the experiment compared to the initial section. In the course of time the human operator seemed to change the strategy (indicated by frame); this section is displayed in detail and demonstrates the coordination in time of the two muscle groups. Both possibilities of force gradation have been used involuntarily in combination: decrease of agonist and increase of antagonist activities which can be seen in details of the EMG's. This type of coordination could not be observed in the initial phase.

RESULTS

Tracking tasks and especially the critical tracking task have been designed in order to study neuromuscular performance. The value λ at the end of the critical tracking task was reproducible for each subject which can be compared, e.g. to results of /2/. The fine structure of the averaged EMG gives insight into the coordination of agonist and antagonist which cannot be obtained by analysis of the mechanical output. For clincial tests of neuromuscular disturbances appropriate criteria will have to be developed to detect and quantify the degree of the disorder.

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CONCENTRIC AND ECCENTRIC CONTRACTIONS OF HIP AND TRUNK MUSCLES WHILE SEATING ONESELF AND RISING.

Ane Friis Christiansen, Ulla Kardorf and Finn Bojsen-Møller

Anatomy Department C, University of Copenhagen, Denmark.

Electromyograms show periods of activity. Combined with a photographic registration of movements of body segments and joints the 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.

While 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 work was to locate the periods of concentric and eccentric contractions in four muscles selected for their importance for the described movements.

Methods

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

The movements, seating oneself and rising from a chair, were registered photographically with strobelight and lighttracks. The strobelight was flashed with 5 Hz and showed thus the position of the body at each 200 msecs. The lighttracks were formed by diodes placed on the shoulder, on the anterior and posterior superior iliac spines, on the trochanter major and at the knee. The diodes were fed by a 50 Hz signal from which, however, one impulse was omitted each second (fig. 1a). The 50 Hz signal was registered on the oscillograph together with the EMG signals (fig. 1b). The missing impulse was used for two purposes: 1) to number correctly (in msecs) the individual light-dark periods of the four lighttracks thus establishing a chronographic correspondence between the tracks, and 2) to link these with the electromyograms.

Results

The photographic registration of the movement: seating oneself, is shown in fig. 1a with the corresponding EMG (1b) and analysis (1c) added. In the photograph each light-dark period lasts 20 msecs. The missing signal is 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 arrows).

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 (1c). The two diodes on the os ilium 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^{o} flattening of the lordosis is kept as part of the sitting position (1c).

As 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 touch-down where the muscle with a concentric contraction erects the trunk.

The rectus abdominis (r.a.) initiates the forward inclination of the trunk, supports the flattening of the lordosis and the abdominal wall and ends with an eccentric contration.

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.

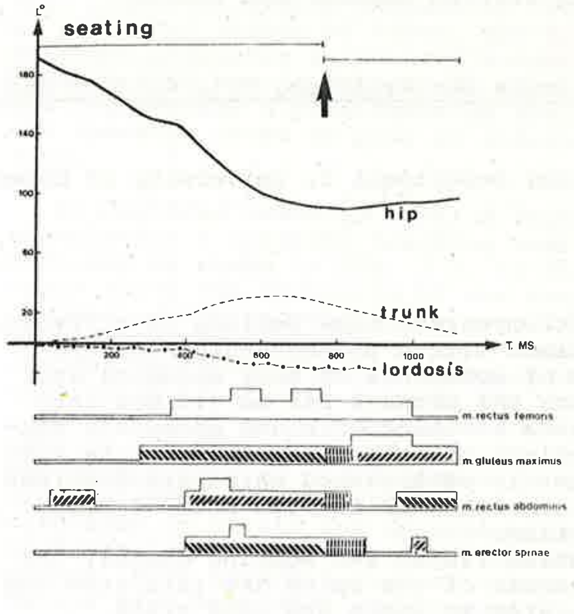

Fig. 1c

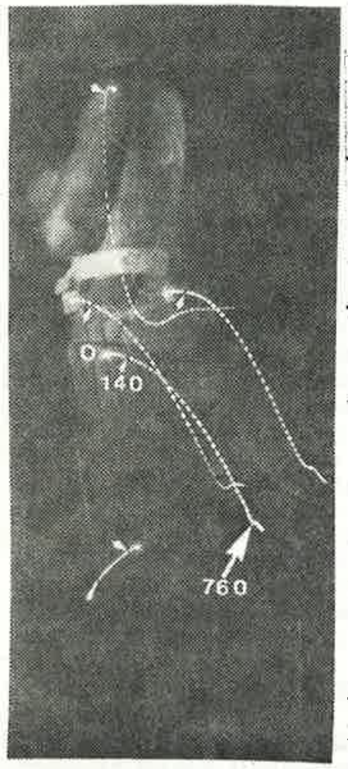

Fig. 1a

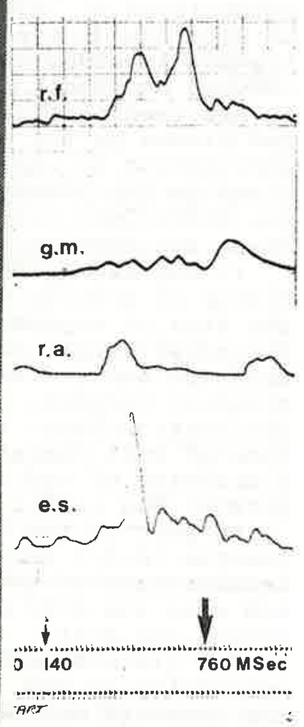

Fig. 1b

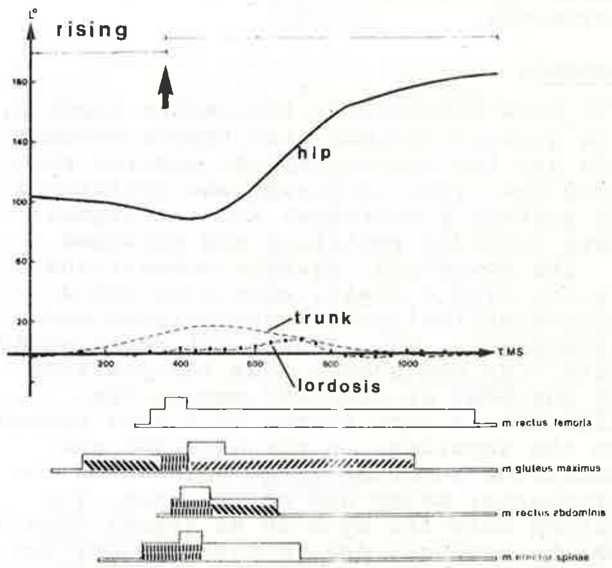

Fig. 2

eccentric contraction

isometric contraction

concentric contraction

BEHAVIOUR OF POSTURE MACHANISMS AT LOADING THE SPINE OF SCHOLL-CHILDREN

Franjo Gračanin, Robert Blumauer, Vesna Zeljić and Gabrijela Vrabič

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 perniciousness 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 analyse 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 mechanisms 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 childrens' 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 disk electrodes in the paravertebral muscles of the thoracic and lumbar region, oblique abdominal muscler, flexors and extensors of the knee. Recording by a Van Gogh polygraph (amplification 175 uV/cm; filter 35, time constant 0,01). The shift of the center of gravity was analysed by a statokinesiometric method (Gračanin F. 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 axis, 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 1 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 mm. paravertebrales reg. thoracalis 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 dex., which display reciprocity. The distribution and intensity of the EMG activity according to Figure i show that:

- in general the EMG activity is greater on the left than on the right;
- it is relatively maximal at the beginning of the experiment;
- while holding the satchel in the right hand there occurs, beside a more strongly expressed EMG activity of the left side muscles, an increased activity in mm. paravertebrales reg. thoracalis dex.. 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 statokinesiometry 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 lateral shifts. With the others we noticed differences in the medium radius of the movement under various test conditions, 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 experiments the data obtained can only be treated as preliminary. However, we can state that the distribution patterns of the EMG activity are constant; the difference is merely in the degree of activity with each child. On the basis of these analyses we can draw a conclusior 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 uncertainty as noticed at the beginning of the experiment.

An interesting correlation is to be noted between the distribution, the degree of the EMG activity, the medium radius of the movement and the direction 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 without 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, whereat there are no substantial lateral or backward shifts; and finally, a group with which there can be observed, beside a proper EMG activity, also a contralateral shift of the center of gravity while holding the satchel in the left or the right 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 them is undoubtably the group showing no need for a greater activation of muscles and maintaining the center of gravity in its initial position. With the children with whom the EMG activity is more stressed and the center of gravity is subject to compensatory shifting, there can be foreseen a more rapid occurrence of fatigue and weakening, which due to repeated loading can résult in bad posture and subsequent deformations. However, the shift of the center of gravity without the occurrence of the EMG activity - which was not detected in our group of children - represents a major threat to the population of children since with them the mechanisms responsible for maintaining an upward posture behave contrary to one's expectations. The reasons thereto are different, to be sure, be it that the question is of an insufficient control of postural mechanisms, insufficient functions of motor units irrespective of the etiology (humoral, neural etc.). Anyway, it would be advantageous if the children suffering from asthenic constitution, humoral and metabolic disturbances, defects of the locomotor apparatus, the children who, for notobvious reason, tire quickly, as

well as those with neuromuscular diseases were subjected to a polyelectromyographic analysis under loading to reveal the possibility and degree of compensation. In this way, a negative influence of loading upon posture and development of deformations could be avoided with these children.

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AUTHOR'S APPOINTMENTS

Prof. Franjo Gračanin, M. D., D. Sc.
Klinika za fizikalnu medicinu i rehabilitaciju Kliničke bolnice "Dr Mladen Stojanović", Zagreb

Robert Blumauer, M. D.
Šolska poliklinika, Ljubljana.

Vesna Zeljić, M. D.
Gabrijela Vrabič, PT
Zavod za rehabilitacijo invalidov
Ljubljana

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LUMPED - PARAMETER MODEL OF THE HUMAN THERMOREGULATION SYSTEM

Halil BENGİ, Ph.D. and A.Rifat ERGÜL, M.Sc.
Department of Electrical Engineering, Middle East Technical University, Ankara, TURKEY.

ABSTRACT

Body temperature in man is kept in its normal range as a result of the combined responses of various physiological systems involved in thermo-regulation. 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 partitiened into four simple geometrical segments as follows: Head as a sphere; trunk, upper extremities, and lower extremities each as a cylinder (Fig. 1). Furthermore, each body segment has been assumed as consisting of three concentric (coaxial) 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 in each 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_t) and heat conducting (G_t) 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 distributed 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 0° Centigrade

In the model: C's represent thermal capacitances; G's thermal conductances ; M's metabolic heat generation ; W's muscular heat generation; E's evaporative heat loss by sweating.

RESULTS

The computer program has been organised 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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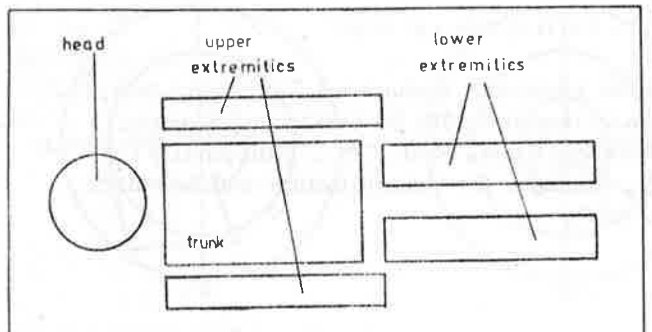

Fig .1. Geometrical shapes used for modeling body segments.

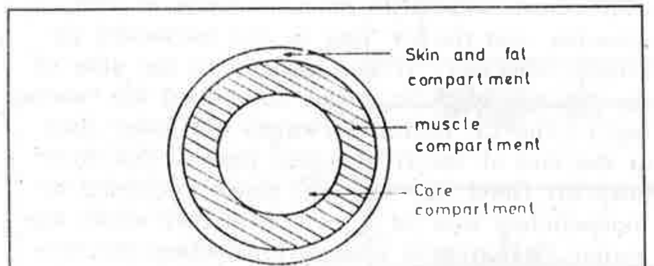

Fig.2. Cross-section of each body segment used in the model.

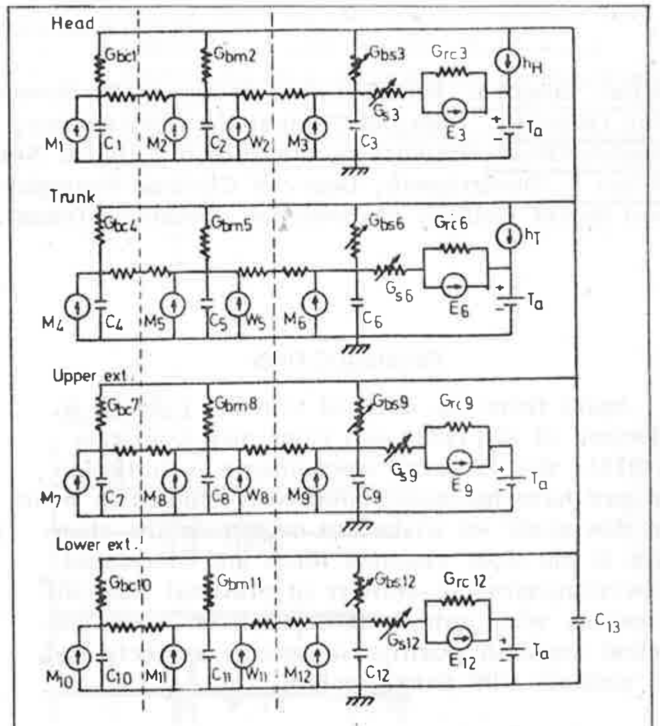

Figure 3. Electrical analog circuit for temperature regulation system in human body

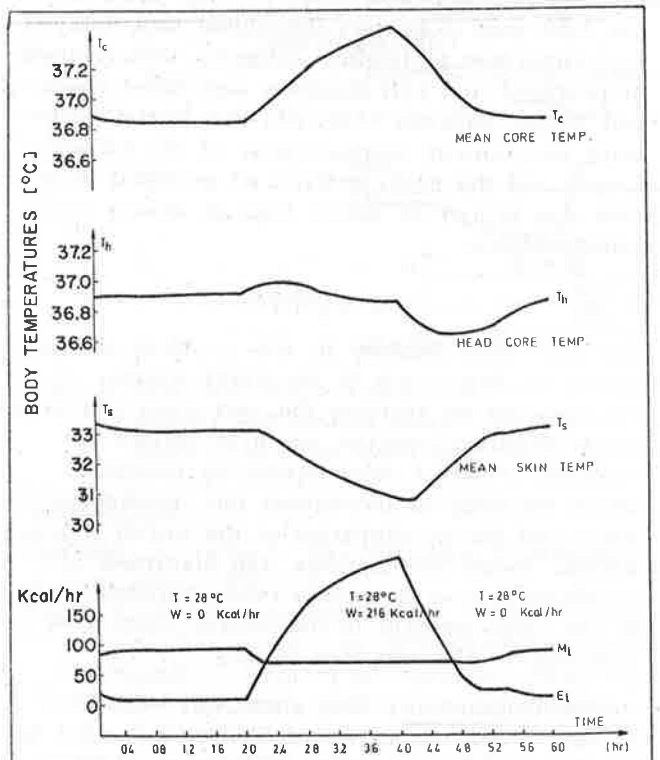

Figure 4. Computed results for two hours moderate exercise (muscular heat generation rate=216 Kcal/hr, three times of the BMR, T_a=28°C, RH=0.6)

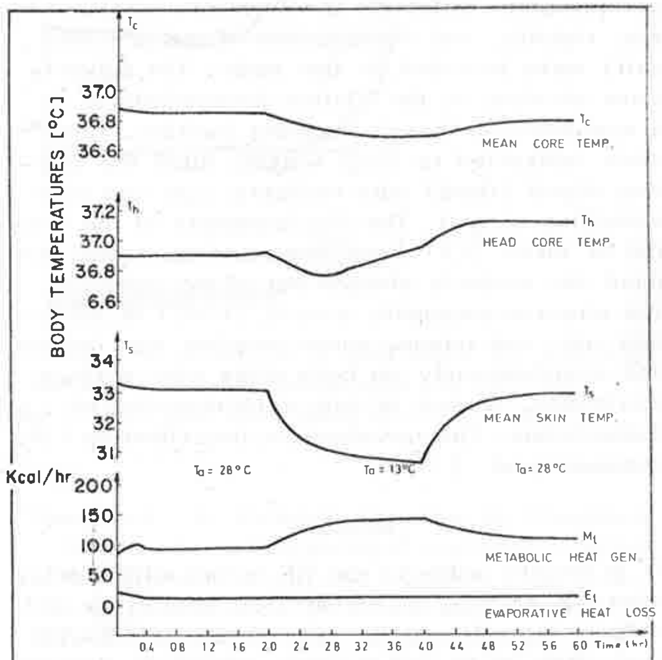

Figure 5. Computed results for two hours exposures to ambient temperature 28°C, 13°C, 28°C (W=0, v=0.1 m/sec, RH=0.6)

POSTURAL CONTROL IN INITIATION OF GAIT

Milan Gregorič, Rehabilitation Centre of SR Slovenia, 61000 Ljubljana, Linhartova 51, Yugoslavia;
Per Odenrick, Dept. of Clinical Neurophysiology, Academic Hospital, 581 85 Linköping, Sweden;
Bogdan Oblak, University Institute of Clinical Neurophysiology, Univ. Med. Ctr., Ljubljana, YU;
Milan R. Dimitrijević, Dept. of Clinical Neurophysiology, Institute for Rehabilitation and Research, and Baylor College of Medicine, Houston, Texas 77030, USA.

INTRODUCTION

Apart from the work of Carlsöö (1966), R. Herman et al (1973) and Cook and Cozzenns (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 L.E. 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 "swing 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 "swing 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.

Acknowledgements: This study was supported by Research Community of SR Slovenia, and by the Bob and Vivian Smith Foundation, Houston, Texas (USA).

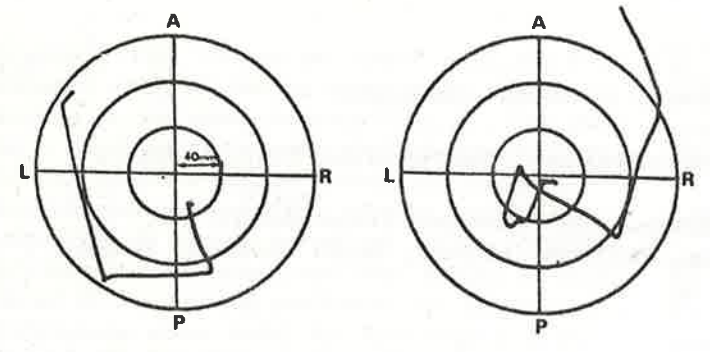

Fig. 1 Trajectory of the displacement of the centre of force on the standing surface on initiation of gait in a healthy subject (left) and in a patient with parkinsonism (right).
A=anterior, P=posterior, L=left, R=right

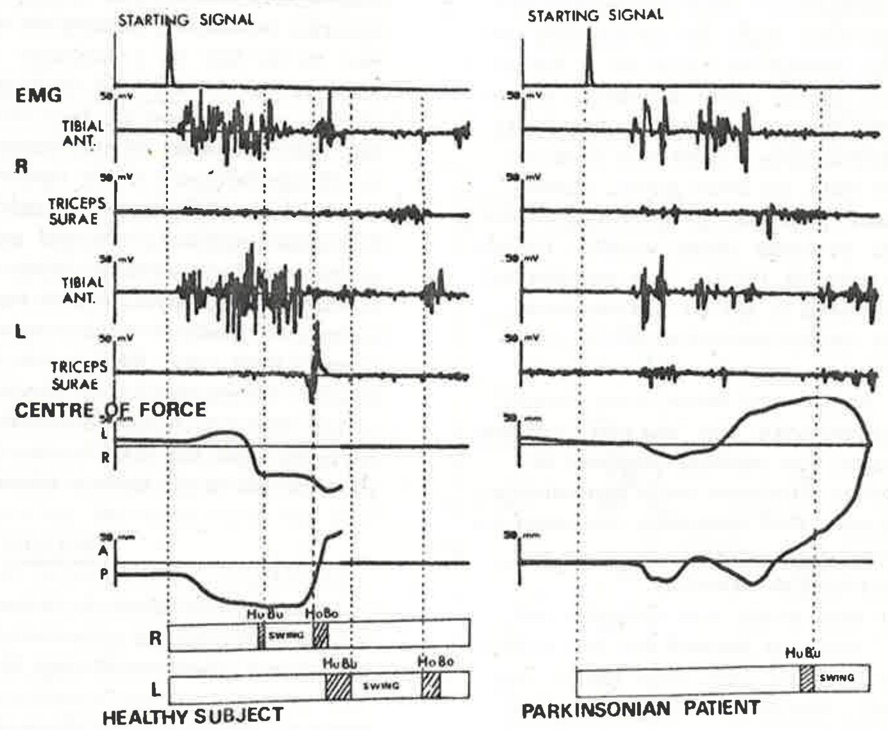

Fig. 2 EMG activity and displacements of the centre of force on initiation of gait after the starting signal. The healthy subject started the gait with the right, and the patient with the left leg. HoBu = the time from Heel up to Ball up ; HoBo = the time from Heel on to Ball on.

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BODY SWAY IN DIFFERENT STANDING POSTURES

Bogdan Oblak, University Institute of Clinical Neurophysiology, The University Medical Centre, 61105 Ljubljana, Yugoslavia;
Milan Gregorič, Rehabilitation Centre of SR Slovenia, 61000 Ljubljana, Linhartova 51, YU;
Per Odenrick, Dept. of Clinical Neurophysiology, Academic Hospital, 581 85 Linköping, Sweden.

INTRODUCTION

The platform stabilometry (statokinesimetry) 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 (statokinesigram and stabilograms) does not give us enough quantitative data on body sway, numerous computer-based processing techniques have been introduced to process these signals (Njiokiktjien 1973, Hufschmidt 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 somehow compared with the 'normal' values. As no 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, Geursen et al. 1975).

METHOD

The subjects (6 male and 1 female, aged 20 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 51.2 s trial two photos were taken by two cameras placed on the left side and at the back of the subject. Electromyograms were recorded simultaneously on six muscle groups of the leg and the back (tibialis anterior, triceps surae and erectus spinae on both sides) using standard bipolar surface electrodes. From the forces the body sway parameters were computed. The mean speed (MS), the RMS value (in sagittal and lateral directions) with regard to the mean position (MP) 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 I).

Table I. Mean values (and standard deviations) of mean speed of centre of force in mm/s for different body positions.

	LEFT LEANING	SYMMETRICAL	RIGHT LEANING
MAXIMAL FORWARD		27.4(±7.2)	
FORWARD	13.9(±6.1)	13.2(±4.7)	14.3(±5.1)
NEUTRAL	13.8(±4.2)	9.5(±3.1)	13.5(±5.2)
BACKWARD	21.5(±5.1)	20.1(±7.2)	22.3(±6.3)

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.

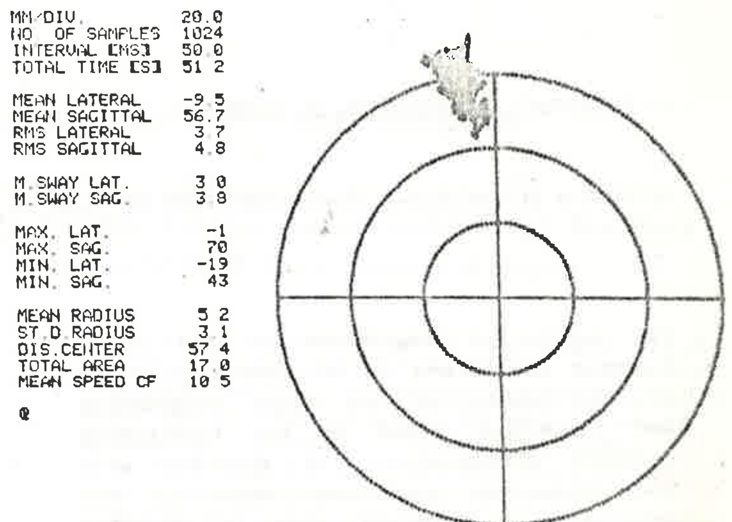

Fig. Typical SKG for forward leaning position.

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 Z forces are proportional to the displacement from the vertical (if the amplitude of oscillations and displacement 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).

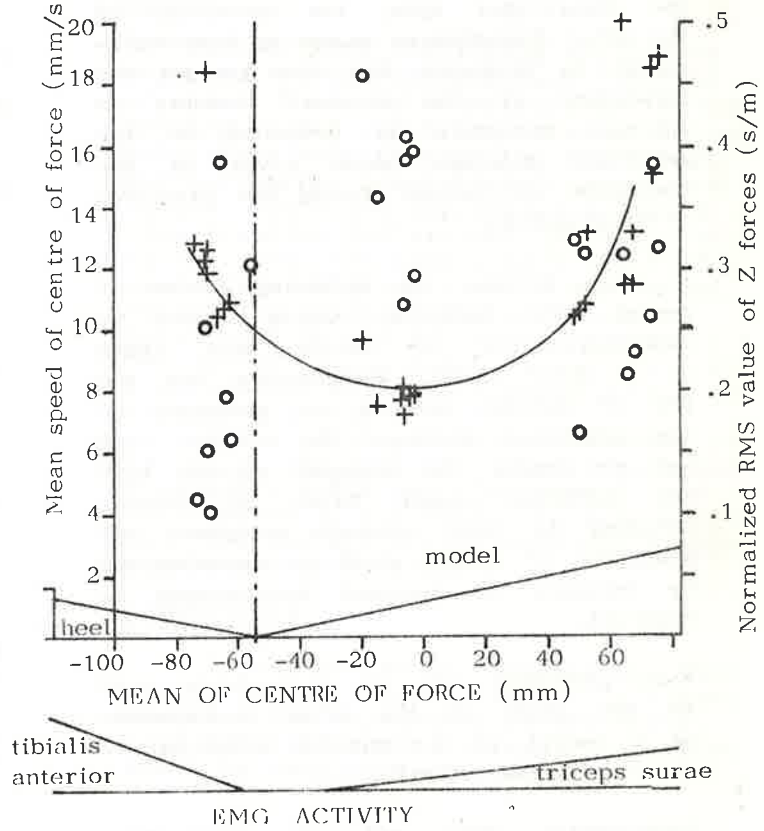

Fig. 2 Typical position-body sway diagram. The mean speed of centre of force (crosses) is more pronounced when the body is not in neutral position. No correlation can be found for RMS values of vertical forces normalized to the body weight and mean speed (circles). For inverse pendulum model this relation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

THE FEEDBACK IN THE SYSTEM OF CORRECTIONS MEANS OF THE SPORTSMEN MOVEMENT DISTURBANSES

A. POCROVSKI, A. RODIONOV, S. SLOBUNOV

ALL-UNION SCIENTIFIC-RESEARCH INSTITUTE OF PHISICAL CULTURE

Two levels of regulation of the motor function are known: to be open-loop, based on the theory of the motor programming and close-loop based on the functioning feedback afferentation. In accodance 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 disturbanses 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 fencer with technical errors caused by deautomatization of skills were taken as a motor model. Mechanograms and the EMG of definit muscles are presented on the graphical display. The subject task was to render the movement pattern with the different speed. After the traning accoding to some concrete programme the stabiliti of skills which is characterized by recovery of movement disturbanses is observed.

That procedure may be succesfully used in the cases of the motor disturbanses as a result of a stressful situation of the competitive actiwiti.

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

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

Motoaki Ishii, Shigemasa Fujii, Hidemaro Yokota, Hiroaki Iwasaki, and Kenji Masuhara
Dept. of Orthopedic Surg. Nara Medical University Kashihara, Nara 634 Japan

INTRODUCTION

As the surgical approaches for the spinal thoracic tumor, anterior decompression or lateral rhachotomy or costotransversectomy 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

Normal adult mongrel dogs were utilized for the study. Dogs were anesthetized with thiamylal sodium and myorelaxant intravenously. Endotracheal intubation was performed to respire artificially with room air. Laminectomies were performed from L1 to L4, and a stimulating electrode was inserted cranially and a recording electrode caudally into the epidural space. SEP monitoring was conducted in the anesthetised dogs as follows: a spinal cord stimulus was applied by means of epidural bipolar catheter electrode. The stimulus current was controlled just over the level to elicit maximal response (duration 0.1ms, 20pps). A bipolar catheter electrode was also used as a recording electrode. A refference electrode (ground) was inserted in the skin.

The average response to 32 stimuli was displayed on an oscilloscope screen which data were recorded for the 10ms period following each stimulus. Control of the stimulus and recording and averaging of the responses were performed by Medelec MS-6. Dogs were divided into 2 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 10 minutes.

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 tensive 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.

M Ishii Sep Changes by Retraction of Thoracic Cord with or without Rhizotomy

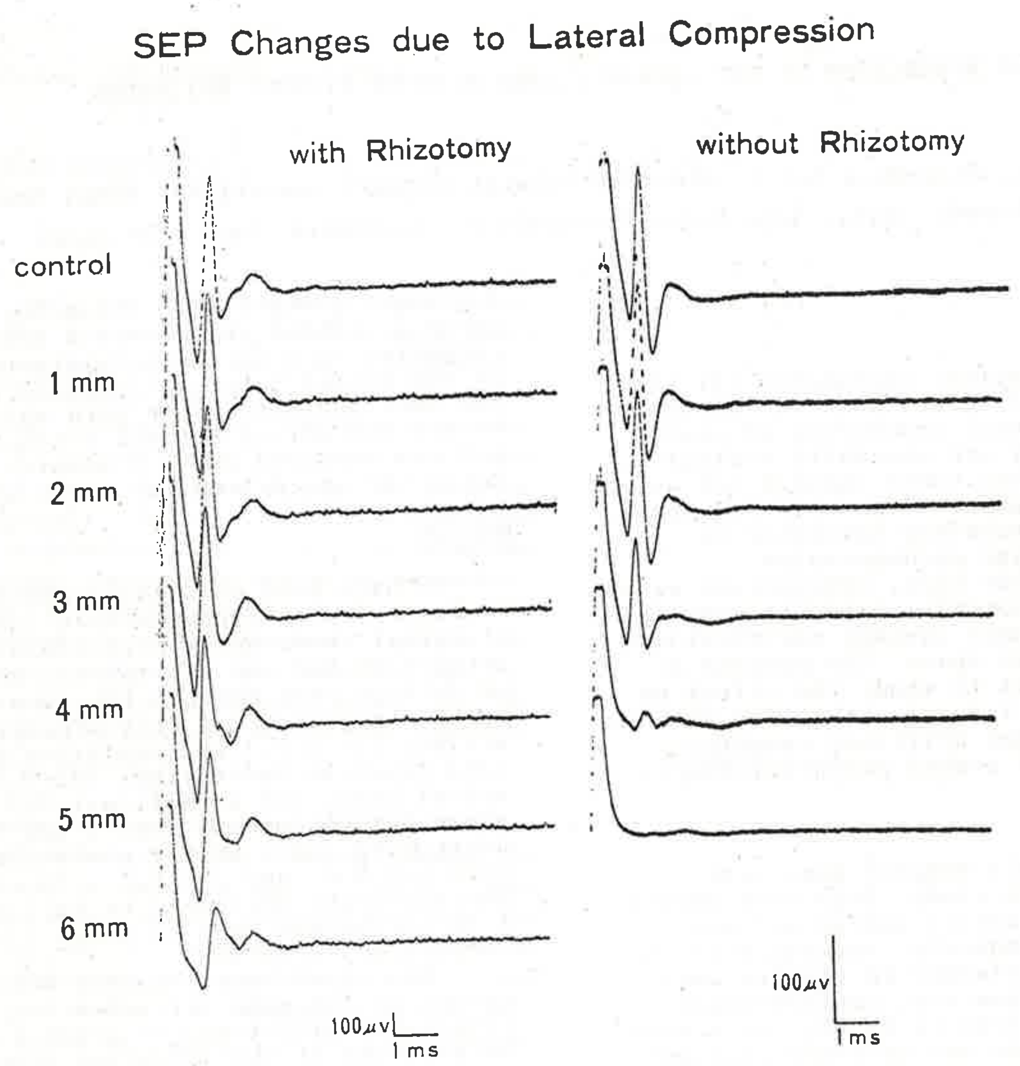

Fig 1

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ELECTROMYOGRAPHIC SIGNAL ANALYSIS FOR EVALUATION OF THE SCOLIOTIC PATIENTS (ABST)

M. BRACALE, M. CESARELLI, C. RAGOSTA, A. MIGNOGNA, C. DE ROBERTO

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

In principle the integrity of the muscolar system of the scoliotic column may be an usefull 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 infact, 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 indeces, 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 indeces record.

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

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QUANTITATIVE BACK MUSCLE ELECTROMYOGRAPHY IN IDIOPATHIC SCOLIOSIS.

Carl Zetterberg[1], Roland Björk[2], Roland Örtengren[2], and Gunnar B.J. Andersson[1]. [1]Department of Orthopaedic Surgery and [2]Clinical Neurophysiology, Sahlgren Hospital, S-413 45 Göteborg, Sweden.

INTRODUCTION

Electromyographic studies of the paravertebral muscles in scoliosis have most often been made to detect neurogenic or myogenic lesions. A comparatively higher myoelectric activity has been found on the convex side in most cases. The differences between the concave and convex side have not been quantified, however, nor have they been related to the degree of the curve. The aims of the present investigation were to study the amplitude and spectral content of the myoelectric signal in relation to scoliosis and load.

MATERIALS AND METHODS

The scoliosis group consisted of girls with adolescent idiopathic scoliosis of uniform morphology. Their mean age was 14.0 years. All curves were primary (major) thoracic, convex to the right and with the vertex at T7 - T9 levels. The thoracic curves had a mean angle of 20.6 degrees (range 5-70 degrees), and the secondary lumbar curves a mean angle of 19.3 degrees (range 0-47 degrees).
The control group consisted of 19 healthy girls with a mean age of 12.6 years. They were physically examined and also investigated by means of Moiré topography.
To load the muscles of the back, the subjects were asked to lie prone and arch their backs for 2 minutes (figure 1). The myoelectric signals were picked up by means of 4 bipolar surface electrodes, placed symmetrically on the back at about 2.5 cm from the midline on both sides at T8 and L3 levels. The signals were amplified and fed to the analog-to-digital converter of a computer for storage and subsequent analysis. The sampling rate was set to 2048 Hz per channel.

The evaluation procedure consisted of signal amplitude estimation and power spectrum analysis. For estimation of the signal amplitude the r.m.s. values were determined in µV. Based on the r.m.s. values an amplitude quotient, Q, was calculated for each spinal level and each subject:

$$Q = \frac{2\ (Vr - Vl)}{Vr + Vl} \qquad (1)$$

where V denotes the r.m.s. values in µV and r and l denotes right and left side respectively. This quotient which minimizes the interindividual variation, facilitates comparison of the side differences between the individuals.

The myoelectric signals were analyzed for frequency changes during the 2 minutes recording, by repeated calculations of power spectra based on consecutive half second signal segments. Shifts of the myoelectric power spectrum along the frequency axis were characterized by means of the center frequency (CF). It has been found that the center frequency decreases exponentially with time in isometric fatiguing contractions. Fatigue curves were also plotted with the frequency value scaled logaritmically. The average slope of the curve in this case was estimated by means of linear regression analysis. This slope, with the opposite sign has been referred to as the fatigue index (Lindström et al 1977).

RESULTS

The scoliosis group showed a higher myoelectric activity on the convex side compared to the concave for both the primary thoracic ($p<0.001$) and the secondary lumbar ($p<0.01$) curves. The myoelectric amplitude dominance on the convex side was found to increase with increasing scoliosis angle in both the primary thoracic ($p<0.05$) and the secondary lumbar ($p<0.01$) curves, (figure 2 and 3). In the control group, there were no significant side difference of the

amplitude values.

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 frequencies 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, (Brussatis 1962, Zuk 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 were also 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 loads on the paraspinal muscles on the convex and concave sides were in proportion to their capacity. An adaptation of the convex sided muscles appears to have occurred to the higher load demand.

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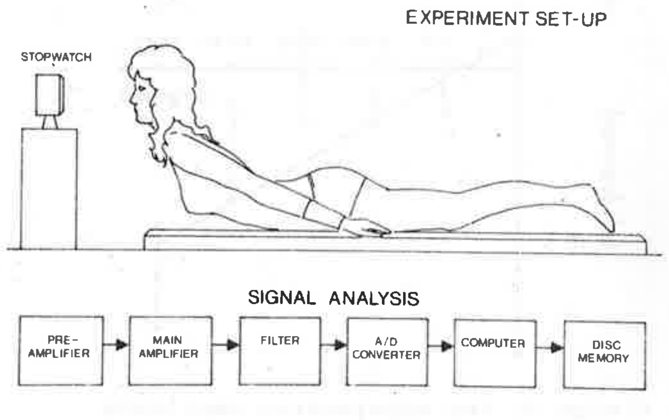

Figure 1. The loading position and the experimental set-up.

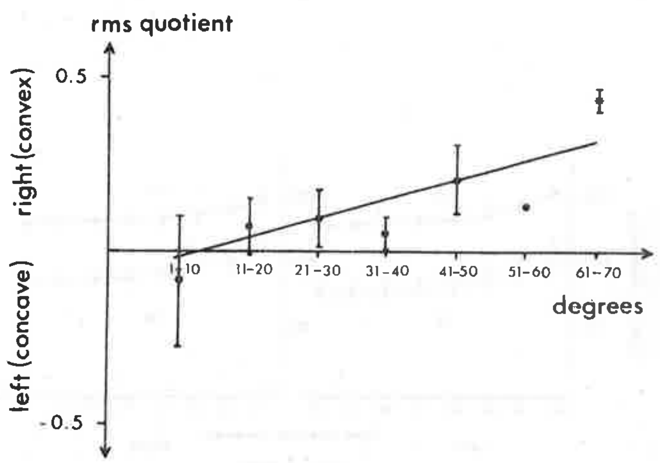

Figure 2. The myoelectric amplitude quotient in thoracic scoliosis in relation to curve degree. Mean values ± SEM. N = 41. All curves were convex to the right.

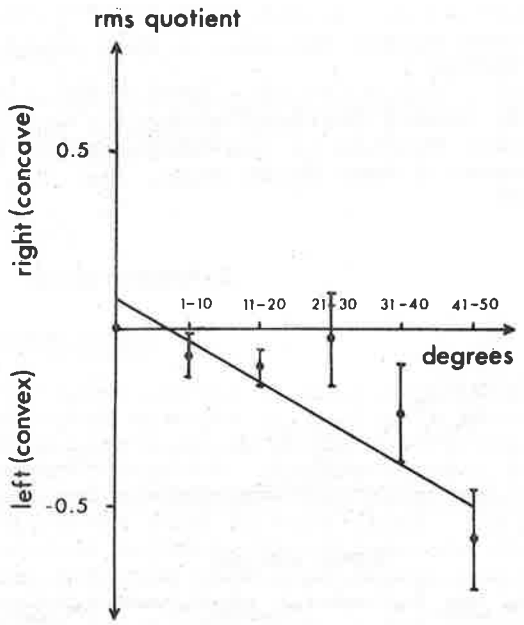

Figure 3. The myoelectric amplitude quotient in lumbar scoliosis in relation to curve degree. Mean values ± SEM. N = 41. All curves were convex to the left.

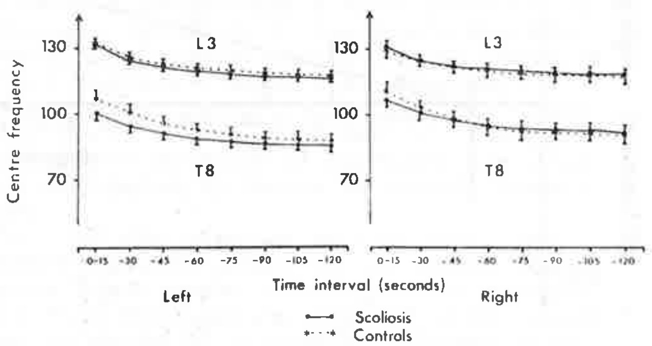

Figure 4. The center frequency curves in Hz during the 2 minute study period. Mean values (SEM). 40 scoliotic subjects and 19 controls.

EQUILIBRIUM REACTIONS IN IDIOPATHIC SCOLIOSIS (ABST)

NEIL ADLER, M.A., R.P.T., ASSOCIATE DIRECTOR, MOTION AMALYSIS LABORATORY,

CHILDREN`S HOSPITAL AT STANFORD

One hundred and twenty cases of idiopathic scoliosis and fifty age-matched controls were tested by a computerized ataxiometer 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 rhytmic 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.

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EMG ANALYSIS BY MEANS OF ASYMMETRY INDEX OF SCOLIOTIC PATIENTS

Dacquino, G.F.(o), Divieti L.(o), Frontino G.(+), Mezzina T.(+), Pissarello M.(o), Rosi M.(+), Villa L.(+)

(o) Dipartimento di Elettronica, Politecnico di Milano.

(+) Centro Don Gnocchi - Parma.

Introduction

Every stimulation procedure adopted so far involves stimulating the convex side of the scoliotic curve (1) . By our procedure(2,3,8)(Scoliotic Patient Electrical Stimulation, SPES), points to be stimulated were selected on the basis of each patient's electromiographic 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 (4). Stimulation was applied by us to the side showing low-level EMG activity.

Asimmetry was evaluated in scoliotic patients' EMG activity by a specially designed Distributed Microprocessor System (see Fig. 1) that receives

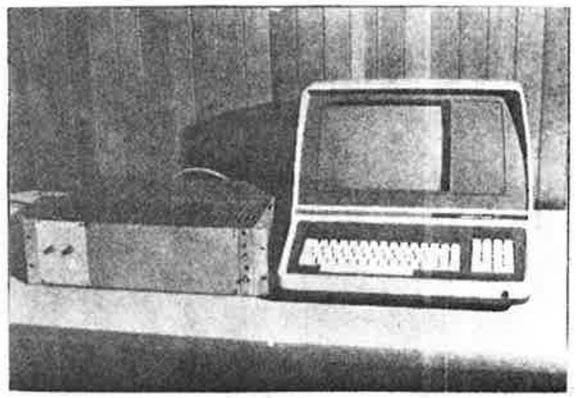

Fig. 1 - The distributed Microprocessor System.

signals form surface electrodes (via a commercially available electromyograph) and processes digitally-converted signals to obtain specific EMG evaluation in real time.

An asymmetry index (AI) taking into account EMG activity in each side of the spine at the apex of every scoliotic curve is discussed in this paper.

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 form the patient's spine are fed into an electromyograph in order to be monitorized. The outputs from the electromyograph are sent to an analog tape recorded and the Distributed Microprocessor System (see Fig. 2).

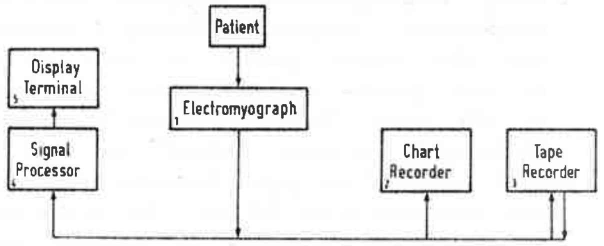

Fig. 2 - Block-diagram of the equipments used.

At power-up, the distributed microprocessor system software performs several test on all critical devices and the communication protocol between the microprocessors to secure fail-safe operation.

The user can select sampling frequency (f_o), acquisition time(T_a) and start command.

The distributed microprocessor system computes partial sums while sampling incoming signals by a double-buffering technique. At the end of the

acquisition time, the distributed microprocessor system computes the AI by the following formula: (5,6,7)

$$R = \int_0^{T_a} |r(t)|\,dt \qquad L = \int_0^{T_c} |l(t)|\,dt$$

$$AI = \frac{R-L}{\max\{R,L\}} \cdot 100$$

(where r(t) is the EMG activity of the right side
l(t) is the EMG activity of the left side
and displays the results on its video terminal.

Since the electromuograph filters signals above 1000 Hz we selected a sampling frequency of 3500 Hz.

The device must operate in a linear range between its saturation and noise levels.

When EMG signals are very low, integrated activity is badly affected by spurius effects, such as superimposed EMG.

Positive AI values resulting from eq.1 corresponds to greater activity on the right side,and negative AI values correspond to greater activity on the left side.

Results

Measurements of AI have been made on 39 curves from 30 scoliotic patients and have been repeated consecutively 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:

sub	time (min)	n	AI			Δ	σ
			mean	min	max		
A	6	14	44	42	49	7	2,42
B	12	24	27	20	36	16	3,43
C	18	29	38	26	55	29	6,43

n = number of measurements taken.

The highest range of variability was observed in patient C who showed a difference of 29 between AI max and AI min. Note that this experiment lasted 18 minutes. Generally it takes less than 2 minutes to repeat the test 4-5 times and the difference Δ is near 10.

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

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

Fig. 3 shows the mean AI of each scoliotic curve as a function of its Cobb degree value. As expected there was no relation between AI and the seriousness of scoliosis.

By measuring AI for every vertebral level of a scoliotic curve it can be found that its maximum value corresponds to the apex of the curvature.

Conclusion

The use of AI as a means of determining the

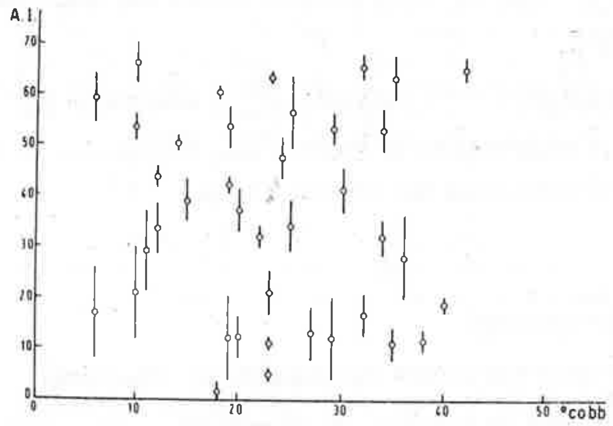

Fig. 3 - A.I. as a function of Cobb degrees

sites where electrodes should be applied to effect electric stimulation of scoliotic patients (SPES) has given good results at the Centro Don Gnocchi (Parma) with 80% of success on 97 curves from 64 scoliotic patients,at the Istituto "G.Pini"(Milano) and Centre des Massues(Lyon)the results of which can be found in (ref. 2 and 3).

By the Distributed Microprocessor System adopted by us it will be 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 for the prognosis of the disease.

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A DISTRIBUTED MULTIPROCESSOR SYSTEM FOR REAL TIME MYOELECTRIC SIGNAL ANALYSIS

Dacquino G.F.(+), Divieti L.(+), Pissarello M.(+), Rosi M. (o), Sardini E.(+), Vanelli P.F.(+)

(+) Dipartimento di Elettronica, Politecnico di Milano

(o) 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.(1-3). 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 cost;
d. computation potential similar to or even higher than that general purpose computers, and
e. reduced space requirements and weight which spells the opportunity to make investigation 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 of 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 pipelined mode.

EMG signal coming from surface electrodes require a sampling rate up to 5 kHz. 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 needles are used instead of surface electrodes to obtain EMG signals.

Difficulties arise with algorithm implementation because the processing unit might possibly be unable to perform the entire operation before next sample is received. To overcome these 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-ported 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-buffering technique. The number of two-ported 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 channels, 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 5K of RAM memory, a serial I/O interface, an arithmetic processing unit (Am9511A), 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 of each of the subunits. Tasks to be

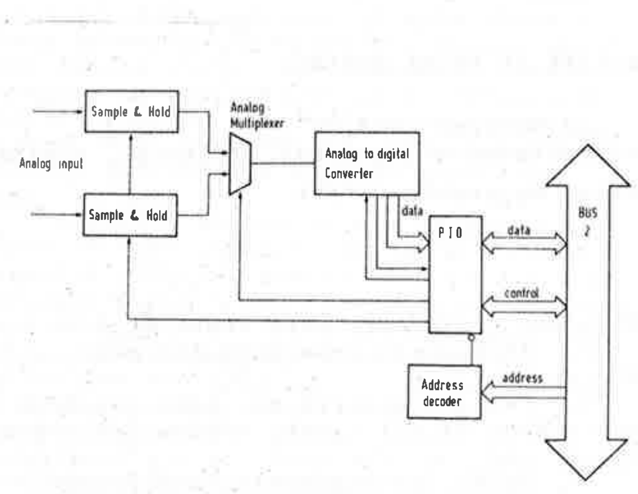

Fig.1 - The acquisition subunit block diagram

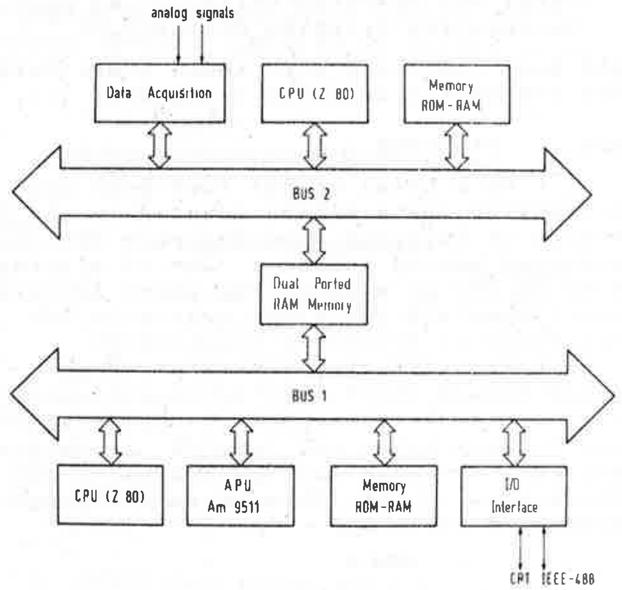

Fig. 2 - The processing subunit block diagram.

performed on the data acquisition subunit are: testing every subunit device, channel selection, channel gain control, A/D control, and communication with the processing subunit via the dual-ported RAM memory.

The entire system is managed by the processing subunit which also interact with the user, who can select one of the specific menu driven programs for system operation, testing, running index EMG signal, EMG sample, acquisition and storage, and EMG sample transfer to a general-purpose computer from one or several channels.

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.

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LOCOMOTOR DISABILITY IN SPINA BIFIDA

Katherine Laughton-Ladd[1] and Herbert Ladd,[1,2]
1. Department of Medical Research, Rehabilitation Institute of Montreal, Montreal, CANADA
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 (myelomeningocoele) 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 inquiry 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 1955 to 1975 was taken from hospital records.* Hospital records also provided ongoing medical histories for these patients for the 20 year period and where possible until the end of the study.
2. Data Classification Analysis: The level at which the myelomeningocoele occurred was recorded, usually in descriptive terms. These have been classified as follows;
   1. Cervical, cervical-dorsal/thoracic,
   2. Thoracic, thoraco-lumbar,
   3. Lumbar, and
   4. 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 (1955-75) or subsequently at the end of the study. These categories are described as follows:

1. Not Able to Walk, either 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.

* These were Institut de Réadaptation de Montréal, l'Hopital Ste-Justine pour enfants, Montreal Children's Hospital, Montreal Neurological Institute, l'Hopital Marie Enfant, Shriners Hospital For Crippled Children and Quebec Society for Crippled Children.

All data collected were coded numerically for computer processing using SPSS (5).

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 there remained 196 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 I.

TABLE I

COMPARISON OF LOCOMOTOR ABILITY-PERCENTAGE DISTRIBUTION

RATED LOCOMOTOR ABILITY	MALES-FEMALES	WITH HYDROCEPHALUS YES	NO
1	6.25%- 8.6%	11.0%	4.0%
2	25.0%- 29.3%	28.0%	27.0%
3	51.25%-38.8%	41.0%	46.0%
4	17.50%-23.3%	20.0%	23.0%
TOTAL	100.0%-100.0%	100.0%	100.0%
NO.CASES	80 116	106	90

Other factors that could be considered to effect ability to walk which were investigated were; the repair of the meningocoele, and secondary complications or deformities of an orthopaedic nature such as paralysis of the lower limbs, foot deformities, subluxated hips, spinal deformities such as scoliosis etc.

Surgical repair of the lesion was done in approximately 80% of all cases, usually within 3 months of birth. Seventy-four percent had orthopaedic problems, and over half of these required multiple surgical interventions to correct or alleviate the condition. These averaged 4.5 per patient mostly before the age of 7. There was little change in this treatment picture over the period studied; in addition, further subdivision of these groups resulted in such small numbers that detailed comparisons could not be carried out when controlling for these factors.

Locomotor ability has appeared to be related to level-of-lesion. That is locomotor ability improved as the level-of-lesion moved distally. (Table 2) Similar to that reported by Hoffer et.al. (3).

TABLE 2

LOCOMOTOR ABILITY AND LEVEL OF LESION

RATED LOCOMOTOR ABILITY	LEVEL OF MYELOMENINGOCOELE 1	2	3	4	TOTAL CASES
1	3(19%)*	8(24%)	0(0%)	2(3%)	13 (7%)
2	4(25)	10(30)	20(32)	17(22)	51 (27)
3	6(37)	11(33)	27(43)	40(51)	84 (44)
4	3(19)	4(12)	15(24)	19(24)	41 (22)
TOTAL	16(100)	33(100)	62(100)	78(100)	189 (100)

Chi-square with 9 d.f.= 34.40; P <.001

* NUMBERS IN ( ) are percentages.

In the above data there were 13 cases rated as not being able to walk; however, 11 of these were considered too young to be able to do so. If these latter cases were to be excluded from the comparison the differences in locomotor ability related to the level of the lesion are no longer significant.

The age of the patient appears to be one factor that indicated the extent of locomotor disability. These data are presented in Table 3.

TABLE 3

LOCOMOTOR ABILITY AND AGE

BIRTH COHORT YEAR

RATED LOCOMOTOR ABILITY	1960 and Before	1961 to 1970	1971 to 1975
1	1(15%)*	3 (3.8%)	11 (22.4%)
2	34(50.0)	13 (16.4)	7 (14.3)
3	25(36.8)	39 (49.4)	22 (44.9)
4	8(11.7)	24 (30.4)	9 (18.4)
TOTAL	68(100.0)	79 (100.0)	49 (100.0)

* NUMBERS IN ( ) are percentages.

It appears in Table 3 that the older the patient the more likely he/she will be confined to a wheel-chair in order to be mobile. Of those born before 1960 (18 years or older) 50% were wheel-chair dependent. This proportion declined to 14% for the youngest group (born after 1971). The optimum in locomotor ability occurred for those patients between the ages of 7 and 17 where almost 30% were rated as being able to walk independently.

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 wheel-chair 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 outcome.

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This research was funded by Health and Welfare, CANADA. Project #6605-1394-53.

PREVENTION OF DOWNWARD DISLOCATION OF THE SHOULDER JOINT

Masafumi Kohno and Atsushi Taguchi, Division of Orthopaedic Surgery, The Japan Red Cross Society, Nagasaki Atomic Bomb Hospital, and Ryohei Suzuki, Nobuyuki 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 loose 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 posibility 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 rhomboideus. 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 rhomboideus 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 according to the added load as compared with 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 rhomboideus, 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 arms 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.

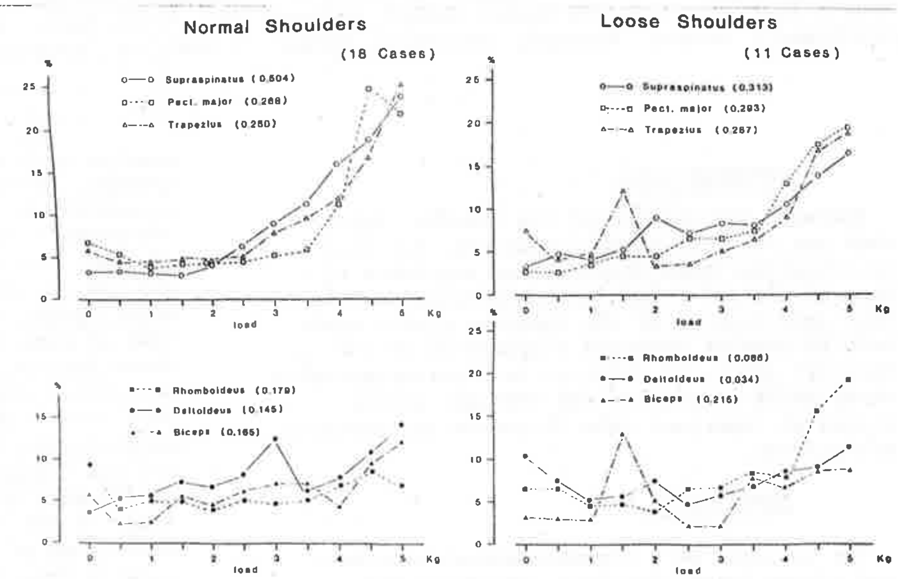

Fig. 2 Relations between load and muscle activities
( ) - correlation coefficient

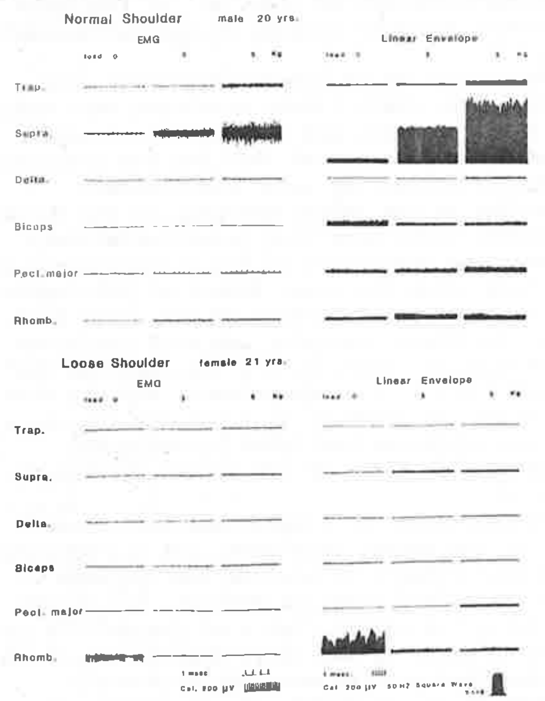

Fig. 1 EMG and linear envelope

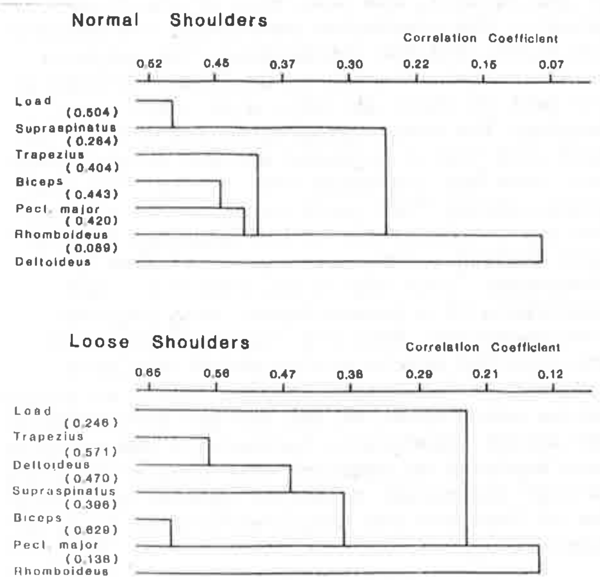

Fig. 3 Dendrogram of cluster analysis

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EMG DISCHARGE PATTERNS IN THE MUSCLE OF LOWER EXTREMITY

Kimihito Ishizaka, Tadaatsu Ito, Yasumasa Shirai, Sumio Kota and Junichi Yamaguchi

Department of Orthopaedics, Nippon Medical School, Tokyo, JAPAN

Introduction

Recently many papers have been published which deal with computer analysis of the motor unit action potentials. However, there are few literatures among them which throw some informations on the amplitude of action potentials dealing with the stabilizing mechanisms of the muscles of the lower extremity in standing position of human being. There is a possibility that the stabilizing mechamism of human being in standing position are clarified by means of quantitative analysis of the amplitude of the motor unit action potentials. We investigated a reciprocal relationship between any one amplitude of action potential and its following 100 consecutive amplitudes of the discharges, and some results were obtained from these analysis to suggest the stabilizing mechanism of the standing position.

Materials and Methods

The subjects ovserved consisted of 86 normal persons and 124 patients with low back pain, that is 56 patients with intervertebral disc herniations of L4/5 level, 40 patients with lumbar spinal canal stenosis and 28 patients with spondylolisthesis. There were 210 cases in total (table 1).

Examined muscles :
Gluteus maximus
Tibialis anterior
Gastrocnemius

Examined cases :	
Normal persons	86
Patients with lumbar intervertebral disc herniations	56
Patients with lumbar spinal canal stenosis	40
Patients with spondylolisthesis	28
Total	210

table 1

Electromyograms were taken with concentric needle electrodes from the gluteus maximus, the tibialis 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 units consisted of 100 units were calculated one by one manually from the oscilloscope screen regarding to the amplitudes and durations, as shown on fig. 1.

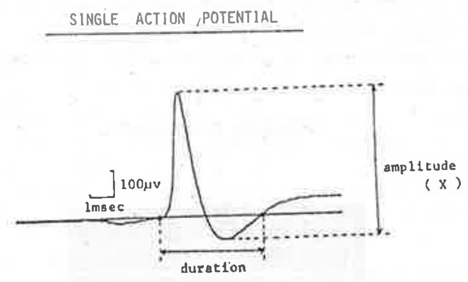

fig. 1

And then fed to the computer these results. The motor unit discharge was finally displayed as functions 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 diagrams.

We had the variable peak to peak amplitudes, such as X1, X2, X3,... so forth to Xn in the consecutive discharges (fig. 2).

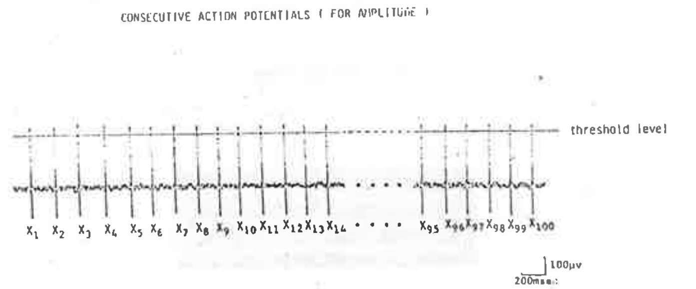

fig. 2

Firstly, we computed the correlation coefficient (r) between X1 and X2, X2 and X3,... so forth to Xn-1 and Xn. Secondly, we computed

K Ishizaka EMG Discharge Patterns in Muscle of Lower Extremity

the correlation coefficient of X2 and X4, X3 and X5,... so forth to Xn-2 and Xn. We did so forth to gain the correlation coefficient (fig. 3).

Equation of the Correlogram

$$\gamma_k = \frac{\sum_{i=1}^{n-k} (X_i-\bar{X})\ (Y_i-\bar{Y})}{(n-k-1) S_x \cdot S_y}$$

$$S_x = \sqrt{\frac{1}{n-k-1} \sum_{i=1}^{n-k} (X_i-\bar{X})^2}$$

$$S_y = \sqrt{\frac{1}{n-k-1} \sum_{i=1}^{n-k} (Y_i-\bar{Y})^2}$$

$$\bar{X} = \frac{1}{n-k} \sum_{i=1}^{n-k} X_i$$

$$\bar{Y} = \frac{1}{n-k} \sum_{i=1}^{n-k} Y_i$$

$$Y_i = X_{i+k}$$

$(n=100)$

fig. 3

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

CORRELOGRAM OF AMPLITUDE

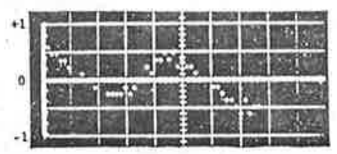

fig. 4

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

AMPLITUDE VARIATION

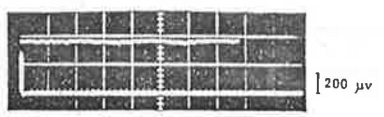

fig. 5

INTERVAL VARIATION

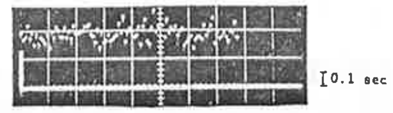

fig. 6

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

Normal correlograms of amplitude (Gastrocnemius)

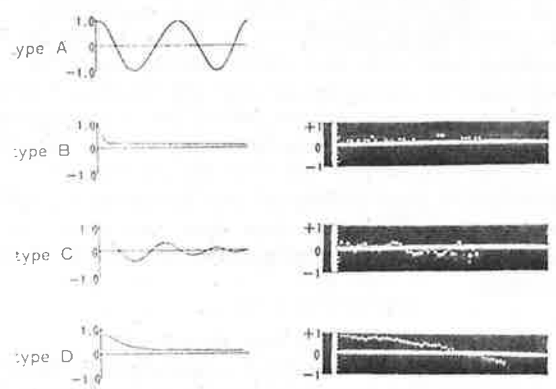

fig. 7

Type A represents a complete periodic changes regarding to muscle action potentials. Type B and D represents no correlation between the former and the later action potentials and type C represents a some periodic changes with some decrement in action potentials in the consecutive discharges.

Results

Much cases 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 cases that type C were obtained in 66.7% at the gluteus maximus, in 68.6% at the tibialis anterior and 17.1% of type D and 14.3% of type B were obtained at the tibialis anterior. As the gastrocnemius, type C were shown in 51.9% and 25.9% of type B and 22.2% of type D were obtained. As to the correlograms collectively, type C were shown in 62.8%, type D were in 22.1% and type B were in 15.1% (table 2).

Correlogram patterns in normal persons

Correlogram patterns / Muscles	type B	type C	type D	Total
Gluteus maximus	1	16	7	24
Tibialis anterior	5	24	6	35
Gastrocnemius	7	14	6	27
Total	13	54	19	86

table 2

Regarding to the analysis of the amplitude variations of the consecutive discharges, much cases showed small variations of the amplitudes in the diagrams in normal persons. The range of the variations of the amplitudes were classified into three classes, that is, (-) variation was below 100 μv of the amplitude, (±) was between 100 and 200 μv, and (+) was over 200 μv. The results were shown as follws; (-) variations were shown in 67.4%, (±) were in 18.6% and (+) were 14.0%. As to the diagrams of the amplitude variations collectively, the gluteus maximus showed (-) in 54.2%, (±) in 25% and (+) in 20.8%. The tibialis anterior showed both (-) and (+) in 91.4% and the gastrocnemius showed both (-) and (±) in 85.1% (fig. 8).

Amplitude variations
Normal persons

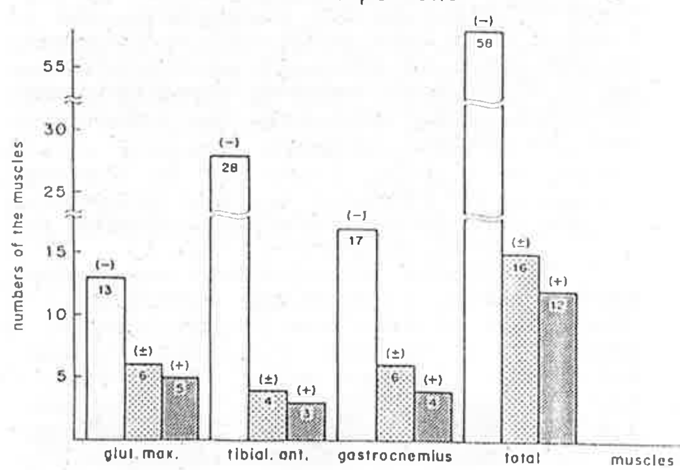

fig. 8

Much cases of normal persons showed type C and D with (-) variations of the amplitude variations, as shown on fig. 9.

Correlogram patterns and Amplitude variations
Normal persons

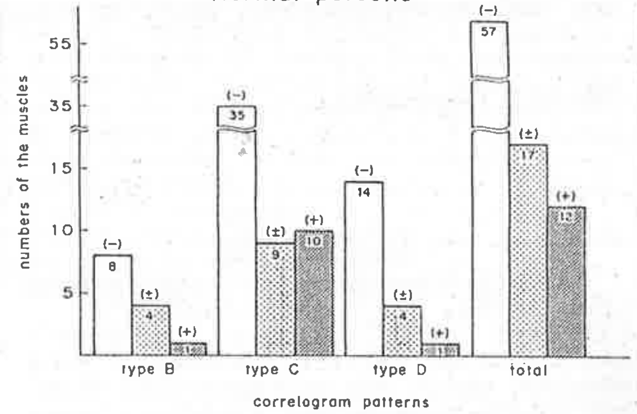

fig. 9

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

Correlogram patterns and Interval variations
Normal persons

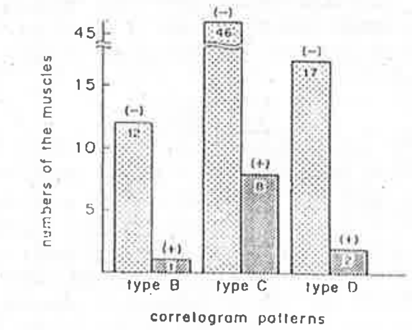

fig. 10

On the other hand, as to the types of the amplitude correlograms in patients with intervertebral disc herniations of L4/5 level, collectively, type C were shown in 17.9%, type B were in 44.6% and type D were in 37.5% (table 3).

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ELECTROKINESIOLOGICAL TESTING IN LUMBOSACRAL SPINAL DISORDERS

Ana Bobinac-Georgievski, Dept. of Rehab. Med. Hosp. Dr J. Kajfeš,

Zagreb, Yugoslavia.

INTRODUCTION

Electrokinesiological studies from Allen(1948), Floyd and Silver(1951, 1955) and Pauly(1966) have confirmed neuromuscular activity pattern of back muscles, which was characterised with two phasic myoelectric activity, lower during trunk flexion and greater during extension. In position of maximal trunk flexion electrical silence was recorded. Basmajian(1978) reported that clinical spasm was not accompanied by increased myoelectic activity, in contrast lower activiti in back muscles was registrated. The present study was performed to investigate electrokinesiological parameters of lumbosacral spine in low back pain and sciatica patients with and without radiologic signs of degenerative disorders of lumbar intervertebral disc, and to evaluate the value of kinesiological testing as a method of objectivisation of functional spinal disorders.

EXPERIMENT

Eighty patients with low back pain and sciatica and forty normal subjects, grouped by age (25-45 yrs old), sex, working activity and radiologic findings od lumbar spine were investigated. All patients had typical history of acute low back pain or sciatica 3-4 weeks before testing for which they have been treated by bed rest. Half of the patients had radiologic findings of lumbar disc degeneration and other half had the normal X-ray of lumbar spine. Electromyographic equipment (Medelec MS6) with two ampifiers and polgon equipment (Medelec PG6) have been used to record surface sacrospinal muscles activita, sinchronously with body movements in saggital plane. The silver chloride disc electrodes have been attached bilaterally, 3cm from the lumbar 5 spinous process and reference electrode was placed 5cm proximally. Polgon sensors have been attached on the right tigh and at the level of body waist. Surface electromyographic (EMG) activity from over the sacrospinal muscel and forward-backward body movement have been continuously recorded on Kodak Linograph paper type 1895. Amplitudes of EMG activity have been measured in milimeters as well as vertical and horisontal distance from one point to the other on polgon diagram, to determine the range and the time of motion. Three repeated movements in saggital plane, performed in patients natural cadence and maximal range of movement which he tolerated, have been registrated.

DATA ANALYSIS

Polyfaktorial 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 phasic EMG activity was registrated during flexion and extension. Electrical silence was registrated in positions of maximal trunk flexion and extension. The mean value of the time for flexion movement was 2 seconds and 1,6 second for extension. The mean range of forward flexion was 110 (96-130) degrees and 28 (16-36) degrees for backward extension. No difference was noted between male and female subjects in control group.

2. Electrokinesiological finding in patients' group (Fig. 2)

EMG activity of sacrospinal muscles in patients with low back pain and sciatica was characterised by monophasic 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=0,1%). 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 58,5 (19-114) and for extension 12,2 (2-24) degrees. The mean time performance was significantly prolonged in patients' group (p=0,1%).

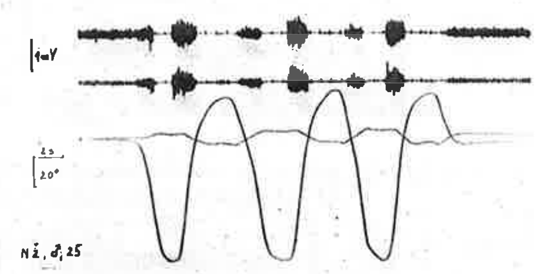

Fig. 1. Electrokinesiological finding in normal subject: EMG activity of the sacrospinal muscles (upper EMG represents right and lower left muscle) and polgon diagram of body positions, during three repeated movements in saggital plane. Sinusoid curve represents the movement of the sensor located on waist and nearly horisontal curve represents the simultaneous movement of right thigh.

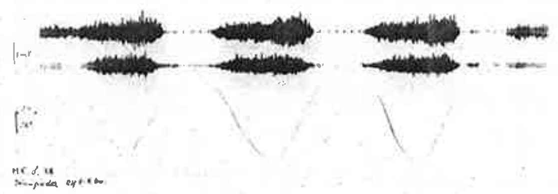

Fig. 2. Electrokinesiological finding from the patient with right sciatica

3. Effect of radiologic status of lumbar spine on EMG activity.

In both groups with and without positive radiologic findings of degenerative changes of lumbar disc, neuromuscular pattern of sacrospinal muscles EMG activity was significantly different from control group (Fig.3.). No difference was observed in EMG activity between patients' groups during forward flexion-extension movements, but difference appeared during backward movements, with significantly greater activity in patients with negative radiologic findings.

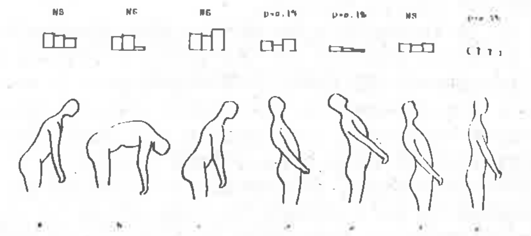

Fig. 3. Histograms of the mean amplitudes of EMG sacrospinal muscles activity in six different phases (a - f) of movements and during standing (g). From left to right: the first histogram-data from the patients with normal radiologic finding, histogram in the middle-data from the patients with manifested degeneration of lumbar disc, the thrid one-data from control subjects. Level of statistical significance of diference between patients' groups is indicated above.

CONCLUSION

It may be concluded that electrokinesiological testing of lumbosacral spine represents objective method which should be considerd in functional diagnosis of patients with low back pain and sciatica, with and without radiologically demonstrable degenerative changes of lumbar intervertebral disc.

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THE STUDY OF F WAVE IN THE PATIENTS WITH INTERVERTEBRAL DISC HERNIATIONS

Junichi Yamaguchi, Tadaatsu Ito, Yasumasa Shirai, Sumio Kota, Toshitsugu Takeuchi
Department of Orthopaedics, Nippon Medical School, Tokyo, JAPAN.

The measurement of the sensory and motor nerve conduction velocities is widely accepted to utilize the diagnostics of the peripheral nerve disorders and injuries. These examinations are useful to know the changes of the distal segments in the peripheral nerves out of the vertebral neural foramen.
But it is impossible to detect the changes of the proximal segment of the nerve fibers which are in the spinal canal. The peripheral nerve fibers are damaged in the spinal canal, lateral recessus or neural foramens in the patients with intervertebral disc herniations (I.D.H.).
On the other hand, F wave are considered to be generated by anti-dromic impulse alomg to the motor fibers and returning back along to the motor fibers again to the muscles. These natures of F wave are favourable to investigate the proximal segment of the nerve fobers upper than the involved area.
Much reports according to F wave were published recently.
In general, there are two kinds of F wave, such as F wave with the shortest latency and the longest latency.
Previous our report concerning to F wave in the patients with I.D.H. were revealed that F wave with the shortest latency had no significant correlation between normal subjects and patients with I.D.H.
In this paper, we will present the results of F wave with the longest latency obtained from normal subjects and patients with I.D.H. who were confirmed the existence of the prolapsed nucleus by the surgery.
And the purpose of this paper is to know the extent of the motor disfunction in patient with I.D.H. by means of measuring the latency of F wave and conduction velocity of the nerve responsible for producing F wave.
Method and Materials;
The subjects examined were 26 patients with L4-5 I.D.H., 14 patients with L5-S1 I.D.H., and 45 normal persons as control. The total cases examined were in 85 cases, which consisted of 51 male, 34 female and ranged from 15 to 58 years old in ages. All of these 85 subjects showed the typical F wave following the supramaximal stimulation delivered to the peroneal and tibial nerves using surface electrodes.
Another surface electrodes placed for detecting F wave on the abductor hallucis and extensor digitorum brevis muscles (Fig 1).

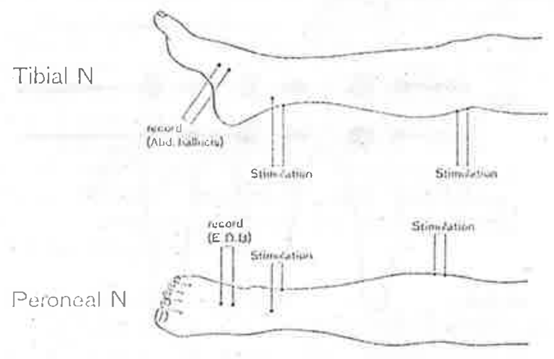

Fig. 1 Method of F-wave examination

Also, the conduction velocities of F wave were meaured by the stimuli induced to the popliteal area and output from the extensor digitorum brevis and abductor hallucis muscle.
The frequency and duration of the stimuli were once per second and 0.5 - 1.0 msec. of duration.
The responses were displayed on the cathode-ray oscilloscope and recorded with X-Y recorder.
This Figure shows, as a sample that fourty-eight F waves evoked consecutively were obtained from the abductor hallucis muscles by stimulating the tibial nerve, and No.1 indicates the onset of F wave with the shortest latency, No.2 indicates the onset of F wave with the longest latency. Moreover, we investigated the lag between the longest and shortest latency, and investigated duration and amplitudes of F wave also (Fig 2).

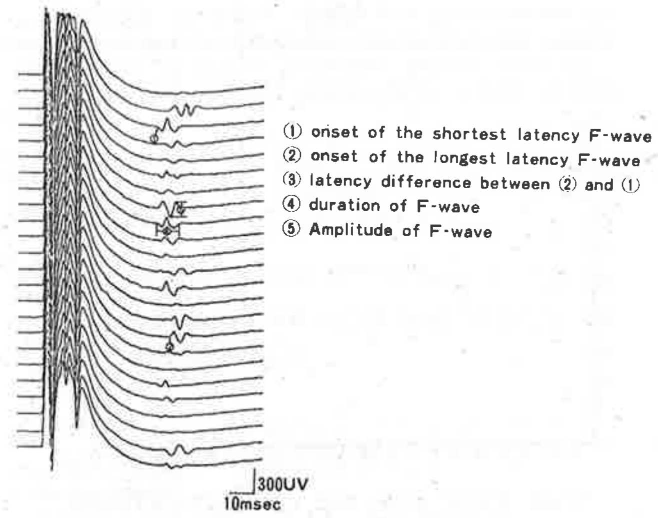

Fig2. 5 items of the examination

F wave conduction velocity (FWCV) was calculated by Kimura's method (Fig 3).

$$\mathrm{FWCV\,(meters/sec)} = \frac{(\text{Distance from stimulus point to } T_{12}) \times 2\,(\mathrm{mm})}{(\text{F latency} - \text{M latency}) - 1\,(\mathrm{msec})}$$

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 F wave with the shortest latency by stimulating the tibial nerve and peroneal nerves are shown on the table 1.

	Tibial nerve (msec)	Peroneal nerve (msec)
Normal cases	36.54 ± 2.27	39.11 ± 3.19
I. D. H. Patients P.	36.22 ± 3.00 N. S.	39.12 ± 4.08 N. S.

Table 1 Mean of the shortest latency of F-waves

There were no significant difference in the mean value of these shortest latency of F wave between normal and patients with I.D.H..
We investigated the FWCV with the shortest latency and resulted in no statistical difference between normal and patients with I.D.H., as shown on the table 2.

	Tibial nerve (meters/sec)	Peroneal nerve (meters/sec)
Normal cases	53.14 ± 4.47	50.87 ± 3.54
I. D. H. Patients P.	53.87 ± 4.93 N. S.	50.51 ± 6.14 N. S.

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

Next, we compared with the FWCV with the shortest latency by stimulating the tibial nerve at the popliteal area between normal and involved legs in the same patient with I.D.H., and were plotted into a graph (Fig 4).

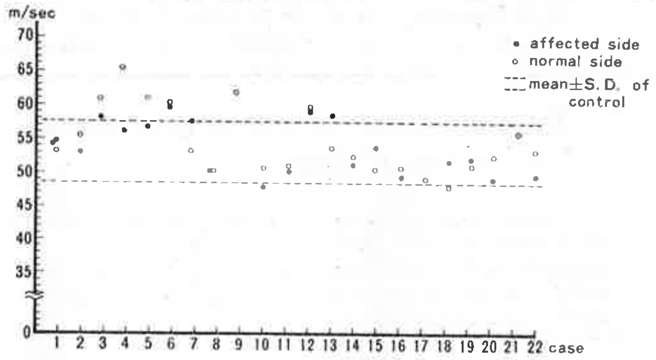

Fig4. FWCV OF THE SHORTEST LATENCY F-WAVE (Stimulation on the Tibial N. at the knee)

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.
Same investigations were done concerning the conduction velocity of F wave by stimulating the peroneal nerve in comparison with normal legs and involved legs in the same patients with I.D.H., even though two cases showed out of the normal range in the involved legs, we resulted in no significant difference between them (Fig 5).

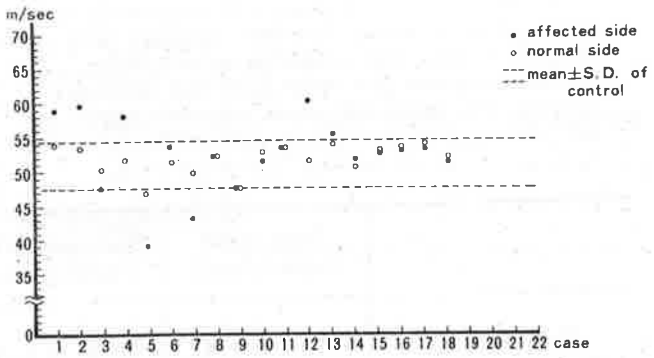

Fig5. FWCV WITH THE SHORTEST LATENCY F-WAVE (Stimulation on the Peroneal N. at the knee)

In F wave with the longest latency, that is, F wave with minimum conduction velocity, the mean value were shown on the table 3.

	Tibial nerve (msec)	Peroneal nerve (msec)
Normal cases	39.68 ± 2.52	42.08 ± 2.46
I. D. H. Patients P.	41.05 ± 2.95 <0.05	45.92 ± 5.75 <0.05

Table 3 Mean of the longest latency of F-waves

In comparison with normal, the longest latency of F wave in the patients with I.D.H. was significantly prolonged statistically and in FWCV , a delay is remarkable in the patients with I.D.H.(table 4).

	Tibial nerve (meters/sec)	Peroneal nerve (meters/sec)
Normal cases	47.32 ±4.38	45.74 ± 2.54
I. D. H. Patients P.	44.94 ± 3.58 <0.05	40.58 ± 5.67 <0.05

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

Comparing with the healthy and involved legs in the same patients with I.D.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 6.

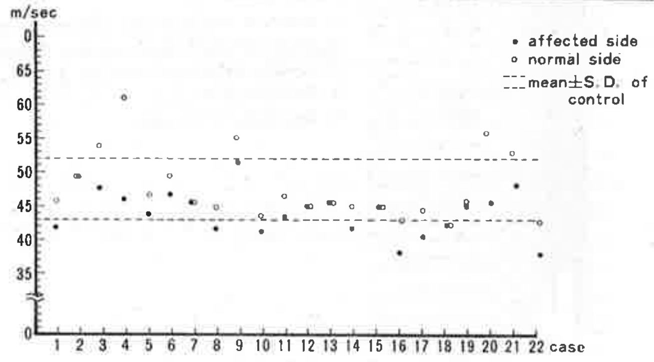

Fig6. FWCV WITH THE LONGEST LATENCY F-WAVE (Stimulation on the Tibial N. at the knee)

The conduction velocity by stimulating the peroneal nerve in the involved legs in the same patients with I.D.H. resulted in the same trend to the tibial nerve (Fig 7).

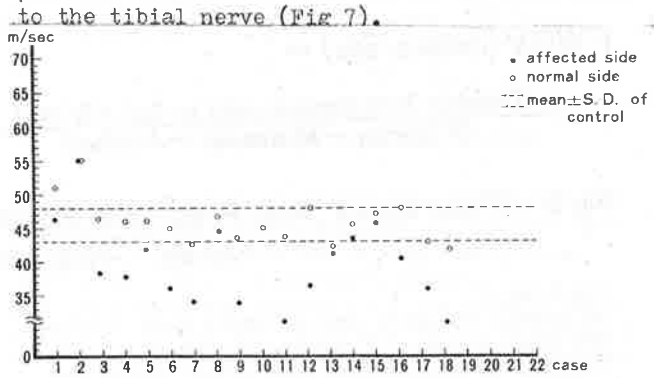

Fig7. FWCV WITH THE LONGEST LATENCY F-WAVE (Stimulation on the Peroneal N. at the knee)

Table 5 shows that the comparison with the lags of the longest and shortest latency of F wave between normal and patients with I.D.H. Whether the stimulation applied on the tibial nerve or peroneal nerve, the lag of these latencies in the patients with I.D.H. showed statistical difference in comparison with normal. The mean value of the duration of F wave were compared with normal and the patients with I.D.H., and resulted in no significant difference. But there are a trend to prolong of the duration in the patients with I.D.H.. As to investigation of M waves, there were no statistical difference in the motor nerve conduction velocity between normal and the patients with I.D.H. (Table 6).

	Tibial nerve (msec)	Peroneal nerve (msec)
Normal cases	3.16 ± 1.02	2.97 ± 1.44
I. D. H. Patients P.	4.82 ± 1.65 <0.05	6.80 ± 3.18 <0.05

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

	Tibial nerve	Peroneal nerve
Duration (msec)		
Normal cases	7.93 ± 2.20	6.54 ± 1.97
I. D. H. Patients	8.01 ± 2.21	7.54 ± 1.88
P.	N. S.	N. S.
M. C. V. (meters/sec)		
Normal cases	52.79 ± 4.02	51.93 ± 3.49
I. D. H. Patients	51.42 ± 5.39	50.34 ± 5.42
P.	N. S.	N. S.

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

Discussion;

In 1950, Magladery and McDougel 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 mixed peripheral nerve to the muscle was electorically stimulted.

This response was called F wave.

And then many investigators, such as Gassel and Wiesendanger, Mcleoad and Wray, Mayer and Feldman, and Kimura, have reported F wave in man.

F wave in man is due to the recurrent discharges of a few motoneurons produced by an anti-dromic volley.

Recently, some investigators have reported that the conduction velocity of F wave with the shortest latency were delayed in the patients with mono-root damage. However, the delayed conduction velocity of F 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 F wave of both normal and involved legs which were obtained from the extensor digitorum brevis by stimulating the peroneal nerve 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 L4. L5 and S1 nerve roots, that is, so called multisegmental innervated fibers.

At the damaged site of the sciatic nerve, the pathophysiological changes of the nerve fivers should be axonostenosis or axonochachexia as proposed by Bauence.

Some stimuli would pass through the normal fibers which would demonstrate a almost normal F wave with shorter latency and others would pass through the damaged fivers 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.

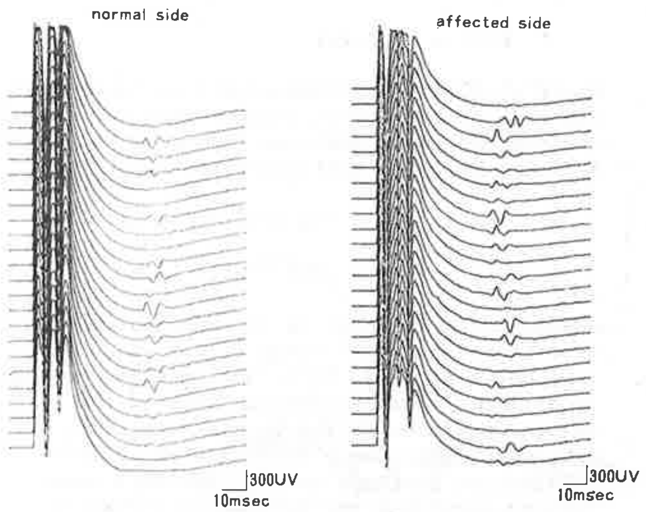

Fig8. F-waves derived following stimulation given to the Peroneal N. at the knee of the affected and the normal side of each same case with I. D. H.

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LOG NONLINEARITY IN PROPORTIONAL MYOELECTRIC CONTROL

P. Parker*, H. Evans** and R. Scott*
* 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 processors in the prosthesis 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 prosthesis 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 $\hat{s}(t)$ of s(t) and in the estimate $\hat{c}(t)$ 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

$$\hat{s}(t) = s(t) + e(t)$$

where the variance, σ_e^2, of the error e(t) is determined by the variance, σ_w^2, of w(t) and the Kalman filter parameters. Thus the variance σ_e^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 $\hat{s}(t)$ ($\sigma_e^2 = .03\ V^2$). An interesting property of the log nonlinearity is that σ_w^2 is independent of the value of σ_n^2 and depends on the form of the probability density of n(t).

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

$$\hat{c}(t) = \text{Alog}\ \hat{s}(t) = \text{Alog}\ [s(t) + e(t)]$$

$$= c(t) \exp\ [2.3e(t)]$$

and the estimate mean square error is

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

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

$$\left. + \exp\left(4.6e(t)\right)\right]$$

Since σ_e^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 e(t). Using a Gaussian approximation for this density it is straight forward 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 $\hat{c}(t)$ is proportional to c(t) and that

the percent error for any value of c(t) is constant. For a $k_0 = 0.2$ the percent error is 20%. This characteristic is evident in Figure 2 which shows c(t) and $\hat{c}(t)$.

OUTPUT VARIABLE

It is interesting to compare the error in a prosthesis output variable, θ, when controlled by $\hat{c}(t)$ or $\hat{s}(t)$. If c(t) has a normalized range of $[1, c_m/c_0]$ where c_m and c_0 are the maximum and minimum values respectively of c(t) then s(t) has a range of [0, log SNR] where $SNR = c_m/c_0$. Now with a range on θ of $[0, \theta_m]$ and with

$$\theta = k_1 s(t) \text{ or } \theta = k_2 c(t)$$

where k_1 and k_2 are control gains it is seen that

$$k_2 \simeq k_1\ SNR^{-1} \log SNR$$

The mean square errors α_1^2 and α_2^2 for these two systems are

$$\alpha_1^2 = k_1^2 E[\{s(t) - \hat{s}(t)\}^2] = k_1^2 \sigma_e^2$$

$$\text{and } \alpha_2^2 = k_2^2 E[\{c(t) - \hat{c}(t)\}^2] = k_0 k_2^2 E[c^2(t)]$$

Now consider the ratio R,

$$R = \alpha_1^2/\alpha_2^2$$

$$= k_1^2 \sigma_e^2 / k_0 k_2^2 E[c^2(t)]$$

$$= \sigma_e^2 SNR^2 / \left(k_0 E[c^2(t)] \log^2 SNR\right)$$

If we let c(t) have a uniform probability density then R becomes

$$R = 3\sigma_e^2/(k_0 \log^2 SNR)$$

with $\sigma_e^2 = 0.03\ V^2$, SNR = 50 and $k_0 = 0.2$ (Gaussian density for antilog e(t)) gives R = 0.16. Measurements show k_0 to be approximately 0.1 giving R = 0.31.

CONCLUSIONS

The absolute value of the error in the estimate of log c(t) is independent of c(t) and the percent error is inversely proportional to c(t). The absolute error in the estimate of c(t) is proportional to c(t) and the percent error is constant. In the control of an output variable the use of $\hat{s}(t)$ shows a somewhat lower output error than does the use of $\hat{c}(t)$. However the desirability of a log characteristic for the output variable has not been considered.

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ACKNOWLEDGEMENT

This work was supported in part by National Science and Engineering Research Council Grant A4445.

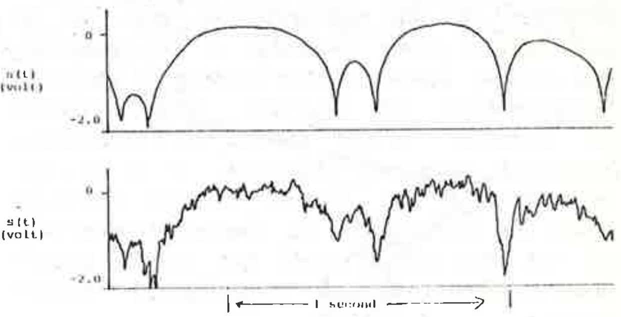

Figure 1

s(t) and its corresponding Kalman filter estimate, ŝ(t)

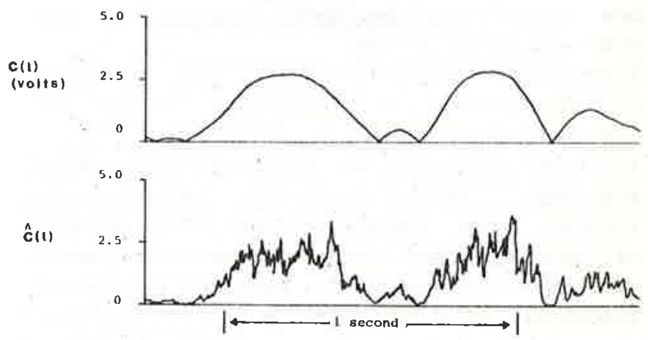

Figure 2

Typical c(t) and its Antilog Kalman filter estimate

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THE USE OF ORTHOPAEDIC APPLIANCE FOR CORRECTING FRONTAL BODY SWINGING OF PATIENTS WITH INFANTILE CEREBRAL PARALYSIS (ABST)

SEMENOVA K.A., ZSHURAVLEV A.M. ZAREZANKOV V.G., Pe PERHUROVA I.S.

Functional electrostimulation of mucles 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 case of infantile cerebral paralysis are manifested by discoordination of lag movements as well as by pathological body swinging on the sagittal and frontal planes in the procese of walking. Normalization of sagittal body swinging is possible as a result of treatment, including surgical treatment. It is by more difficult toc correct frontal swinging which is the result of complex interaction of the muscles of body and pelvic gurdle under conditions of pathological muscle tonus.
In case of normal walking body swinging on the frontal plane does not exceed $\pm 5^{o}$, while in many patients with infantile cerebral paralysis it is increased reaching in difficult cases $\pm 30^{o}$. These movements are conditioned by muscle distony in the body and pelvic gurdle.
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, bracings and wires.
The electrical stimulator with the battery weighs 250 gram, has the size of 33x75xx158 mm and produces series of electrical 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 mcs correspondingly. The latent period and duration of stimulation are equalto 0 - 0,5 and 0,2 - 2,0 s correspondingly.
The electrodes have the size from 20x20 mm2 to 40x250 mm2, 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 contactors that react to swinging relative to the horizontal line.
The sensor is placed on the patint` 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, incrense of stability, ameliortion of biomechanical parameters of walking, facilitation of walking and subjective feeling of motor comfort in patients.

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THE BIOMECHANICAL MODEL OF THE MAN'S LOWER LIMB SKELETON

Dejan Popović, Faculty of Electrical Engineering, Belgrade

Introduction

Gait analyses of the biped is an interesting problem for the kinesiology as well as for the robotics. The computer-aided methods are based on the multi-link rigid body dynamical model of the antropomorphic mechanism /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 e 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.

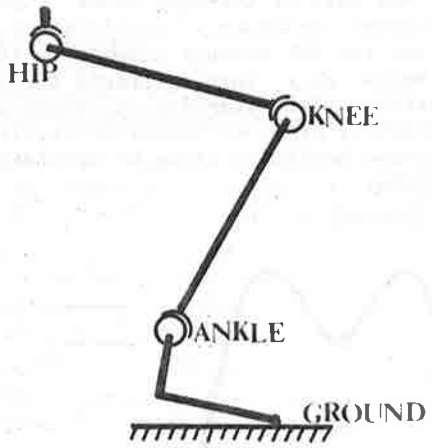

Fig. 1 The dynamical model of the man's leg (rigid body)

Basically the theory of machine and mechanisms is developed on elements which are not the biological ones, i.e. the wheel is one of the principle elements of motion, sliding curve mechanism is the generator of translatory movements from rotation and none of these are to be found in the motion of animals or men.

The robotics rises up from multi-link rigid body mechanics, theory of machine and mechanisms, control theory and it implies the development of medical robotics, gait analyses, rehabilitation engineering etc. The computer-aid cannot reach the problem of real identification of the gait parameters so far.

The prosthetic and orthotic requirments brought in the idea to synthetize the useful, not complicate model of the biological structure. The author believes that the role of the foot and the double ankle joint as well as the polycentricity of the knee and hip joint must be built in the model for kinesiologic investigation.

The biomechanical model of the man's leg

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

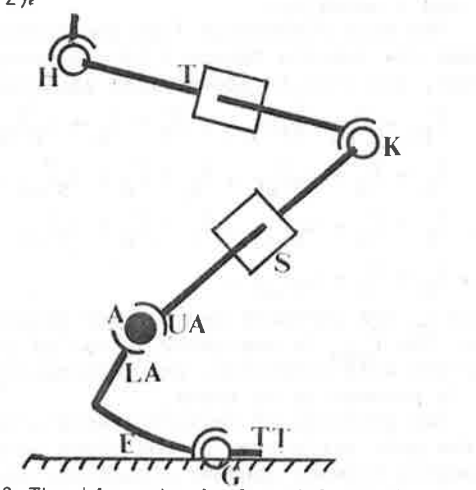

Fig. 2 The biomechanical model of the man's model

Such an mechanical structure is assumed because of the areal ground contact (not just the point foot-ground contact); the movement in the ankle belongs to two joints (articulario talocruralis and articulario subtalaris et articulario talocalcaneovanicularis); 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:

$$m_T \ddot{\vec{r}}_T = -\vec{R}_H + \vec{R}_K + m_T \vec{g}$$

$$\frac{d}{dt}([J_T]\vec{\omega}_T) = -\vec{M}_H + \vec{M}_K$$

$$m_S\ddot{\vec{r}}_S = -\vec{R}_K + \vec{R}_{UA} + m_S\vec{g}$$

$$\frac{d}{dt}([J_S]\vec{\omega}_S) = -\vec{M}_K + \vec{M}_{UA}$$

$$m_A\ddot{\vec{r}}_A = -\vec{R}_{UA} + \vec{R}_{LA} + m_A\vec{g}$$

$$\frac{d}{dt}([J_A]\vec{\omega}_A) = -\vec{M}_{UA} + \vec{M}_{LA}$$

$$m_F\ddot{\vec{r}}_F = -\vec{R}_{LA} + \vec{R}_G{}^F + \vec{R}_{TT} + m_F\vec{g}$$

$$\frac{d}{dt}(m_i{}^F(\vec{r}_i{}^F \times \vec{v}_i{}^F) = -\vec{M}_{LA} + \vec{M}_{TT}$$

$$m_{TT}\ddot{\vec{r}}_{TT} = -\vec{R}_{TT} + \vec{R}_G{}^{TT} + m_{TT}\vec{g}$$

$$\frac{d}{dt}([J_{TT}]\vec{\omega}_{TT}) = -\vec{M}_{TT}$$

where indicies are: T-thigh, S- shank, F- foot, TT - tiptoe, G - ground, A - ankle, K - knee, H - hip, L - lower, U - upper. The notation $\vec{r}$ is for the position vector, J_{QQ} inertia tensor, $\vec{\omega}$ - angular rate, $\vec{R}$ - forces, $\vec{M}$ - muscle couples, m - masses and $\vec{v}$ velocity.

The main difference from the hitherto model is that the inertia tensor J is a nonconstant parameter, and that in geometrical equations:

$$\vec{r}_{TT} = \vec{r}_G + \ell_{TT}\vec{r}_{oTT}, \quad \vec{r}_F = \vec{r}_G + \ell_F\vec{r}_{OF},$$

$$\vec{r}_A = \vec{r}_G + L_F\vec{r}_{oF}, \quad \vec{r}_S = \vec{r}_A + \ell_S\vec{r}_{oS},$$

$$\vec{r}_K = \vec{r}_A + L_S\vec{r}_{oS}, \quad \vec{r}_T = \vec{r}_K + \ell_T\vec{r}_{oT},$$

$$\vec{r}_H = \vec{r}_K + L_T\vec{r}_{oT},$$

ℓ_Q and L_Q are variable appropriate lengths of the links. The $\vec{r}_{oQQ}$ is the unity vector of the segment principle axis direction. The deformation of the foot is assumed to be known.

Two problems of mechanics were solved. First, the gait activities of the biped were given by the angles between segment directions and orthogonal nonmoving coordinate axes, and ground reaction (GR), joint torques and centre of pressure (CP) displacement from the zero moment point (ZMP) were determined. Secondly, the kinematics of the gait were analysed starting with the known muscle couples, and other necessary data. The upper part of the body was adopted in the form of two segments connected with the ball joints.

The movings of the leg, even in the simple locomotion activities (such as ground level walking at the average speed) are very complex.

The dynamic analysis was done on different locomotor functions in order to understand the advantages and drawback of the introduced lower limb model.

Results and Discussion

The normal gait dynamics was chosen for the presentation. The detail Selspot and photo-cammera measurments were taken with instantaneous application of the tenzo-platform and on a line computer /3/. The data acquisition is prepared (lengts changes, inertia tensor, the knee joint moment pole of rotation position, the foot deformation etc.), and computer-aided analyses of dynamics realised.

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

The comparative analysisof 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 belongs 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 part of the body moves in order to bring the bipedal structure in stable state. Just by slow motion the ZMP belongs allways to the support area. Normal gait, turning around the vertical axes, sitting down, standing up, slope walking, running and plenty of other normal activities belong to the above mentioned class of nonstable-stable transitions.

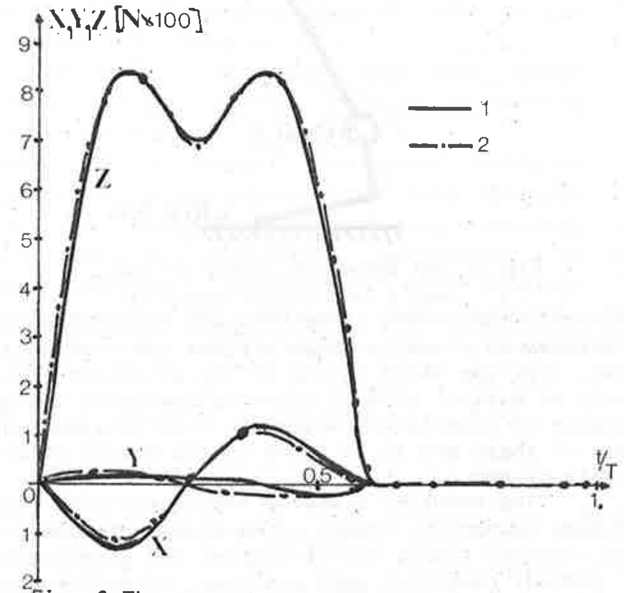

Fig. 3 The ground reaction forces
1-experimens, 2-numerical results

The complexity of the model, i.e. the number of links, many varyable parameters and necessity for numerical derivation is the basic drawback of the model presented.

Conclusion

The new modelling approach of the man's lower limbs is presented. The biomechanical model rises up from the bipedal functional anatomy and physiology of the locomotion activities. The gait analysis and functional rehabilitative assistive devices synthesis /4,5,2/ were the main goals for the investigation and introducten of the new dynamical model.

The biomechanical method involves the biological mechanism of movement, not the simple mechanical one. The non-numerical control is to be applied to the eventual attempt of the gait reproduction of the man-machine system /6/.

The better accuracy and agreement with the experimental results is obtained due to introducing of the functional movements.

Thus, the preliminary presentation is the element in the new man's dynamical model which takes into account the visco-elastical properties of the skeletal constraints, polycentricity of the joints, the deformation and bending of the body links and some other biological characteristic.

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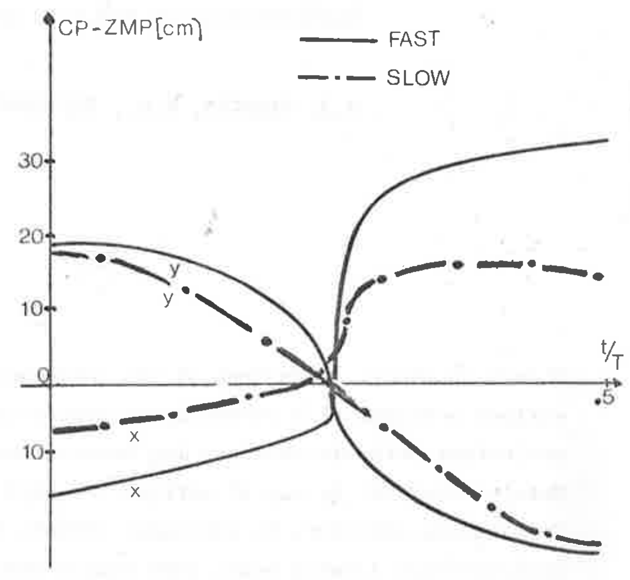

Fig. 3 The CP-ZMP displacement during ground level walking

COMPUTERIZED EMG-ANALYSIS IN EVALUATION OF BELOW-KNEE ORTHOTICS

M.J. WICHERS, M.D., RESIDENT IN REHABILITATION MEDICINE

Object of study: Assessment of the influence of various settings of a conventional double-bar ankle-foot orthosis on lower leg muscle function. Muscles studied by way of surface EMG are:
M. tibialis anterior, M. peronaeus longus, M. gastrocnemius lateral head, same muscle medial head and M. soleus.
If on a significant number of subjects a certain setting of the ankle joint would produce the same type of change in EMG of the muscles recorded, this might elicit the way orthosis influence the gait.

Methods: A subject is equipped with surface EMG electrodes on the muscles mentioned on both lower legs. He or she wears sports shoes, provided with heel and toe contacts, in order to identify stand- and swing phase left and right.
Over the shoe a conventional double-bar below-knee orthosis on a foot-plate is firmly attached. As the ankle joints of the orthosis allows for inserting at will of dowels or springs to produce stops or assists, the action of a conventional brace is achieved. The position of both ankle joints can be adjusted horizontally and vertically. The subject walks on a treadmill. Eight runs of over 60 steps are registered, the first and the last run being without orthosis, run two through seven representing different settings in random order.

The registration is recorded on magnetic tape and re-recorded to be processed in a computer. The computerized processing permits calculation of integrated EMG values per time unit (i.e. per stand phase, swing phase or per second during these phase).
Also exact time measurements can be processed in order to study changes in walking rhythm related to the setting of the orthosis. Furthermore the number of zero-passages of the EMG-signal is computed.
This computer programme allows for interrelating various aspects of muscle function in orthotic gait. Tables and graphs need no longer to be drawn by hand. Quantification of the EMG-signal obtained is nog done by subjectively overviewing a picture on a monitor or an ultra-violet print. These advantages of computerized processing however do present an immense challenge to record the experiments with the absolute minimum of signal noise, hum and movement artefacts.

Results obtained: A series of 4 healthy subjects was recorded in the summer of 1980. This limited pilot study showed that indeed a change of the orthosis did produce some change in integrated EMG values. Processing was done by a simple electronic integrator and computing was painstakingly done by hand. After that computer processing was initiated. We recorded a new small series of healthy persons which is currently being processed.
Regardless of the results of the healthy group

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.

Author's address:
Rehabilitation Dept., building 5,
State University Hospital
Rijnsburgerweg 10,
2333 AA Leiden, The Netherlands.

Experiments performed in:
Laboratorium voor Bewegingsanalyse
(Movement Analysis Laboratory),
Mezenstraat 2A, 2333 VT Leiden.

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EFFECTS OF JOINT IMMOBILIZATION ON MUSCLE ELECTRICAL ACTIVITY DURING WALKING

A.S. VITTENZON, K.A.PETRUSHANSKAYA, USSR, CRIP

By means of a specially designed orthopaedic devices we have studied effects of immobilization 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 immobilization leads to vivid decrease and redistribution of electrical activity during the locomotor cycle. This activitiy decrease is more vivid in case of one-joint muscles rather that 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 smoothened. We suppose that the decrease of muscle activity during walking under conditions of joint immobilization results from a weaker afferental 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 looking 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.

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EMG PATTERNS IN NATURAL AND SLOW SPEED TREADMILL WALKING

David A. Winter, Ph.D.,P.Eng
Jaynie F. Yang, B.Sc. (PT)
Department of Kinesiology
University of Waterloo
Waterloo, Ontario, Canada N2L 3G1

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.

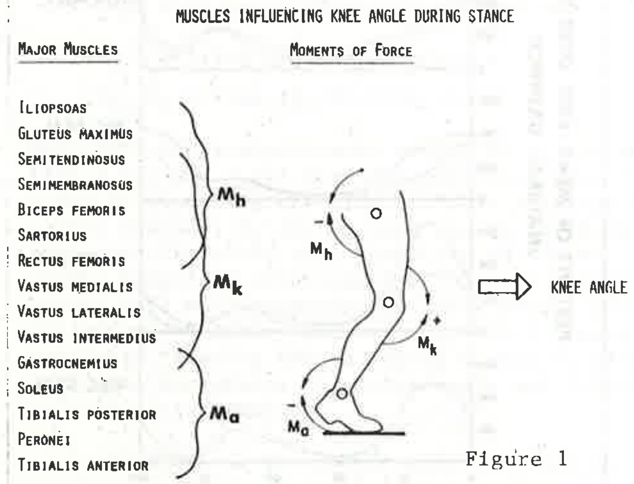

Figure 1

A processed version of the raw EMG that more closely represents the force pattern is desired, and this is found in the linear envelope. Regardless of which EMG signal (raw or linear envelope) is used there appears to be several major shortcomings in the diagnostic process:

(i) Many clinicians compare only the phasic activity of the muscle and ignore the level of activity over the contraction period.

(ii) Comparisons are usually made with those patterns obtained from adults walking at their natural cadence rather than with a population walking at a much reduced cadence more representative of the patient population.

(iii) No effort has been made to account for the large between-subject variability.

The purpose of this paper is to present techniques to overcome these shortcomings and improve the diagnostic process.

METHODS

Eight normal subjects were assessed using surface electromyography on five muscles during treadmill walking at natural cadences plus slow speed (1.75 miles/hr) and very slow speed (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 average 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 average of the patterns for the 8 subjects was then calculated for each muscle and speed.

RESULTS AND DISCUSSION

Figure 2 is 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

Table 1

Coefficient of Variation of Ensemble Average

Muscle \ Speed	Natural	Slow	V. Slow
Soleus	29	30	32
Tib. Ant.	51	43	48
Bic. Fem.	53	63	73
Vast. Lat.	53	74	87
Rect. Fem.	44	57	78

decreases its activity by about 30% as the subjects slowed down but the pattern remains essentially unchanged. The tibialis anterior also underwent a similar reduction at weight acceptance presumeably 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 70% 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 (adaptability) 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 proportional to the mean. Such variability is indicative of the tremendous flexibility of 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 Guy.

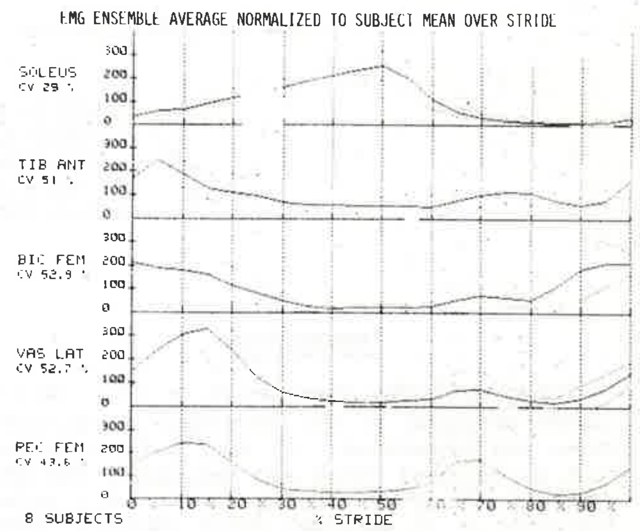

Figure 2

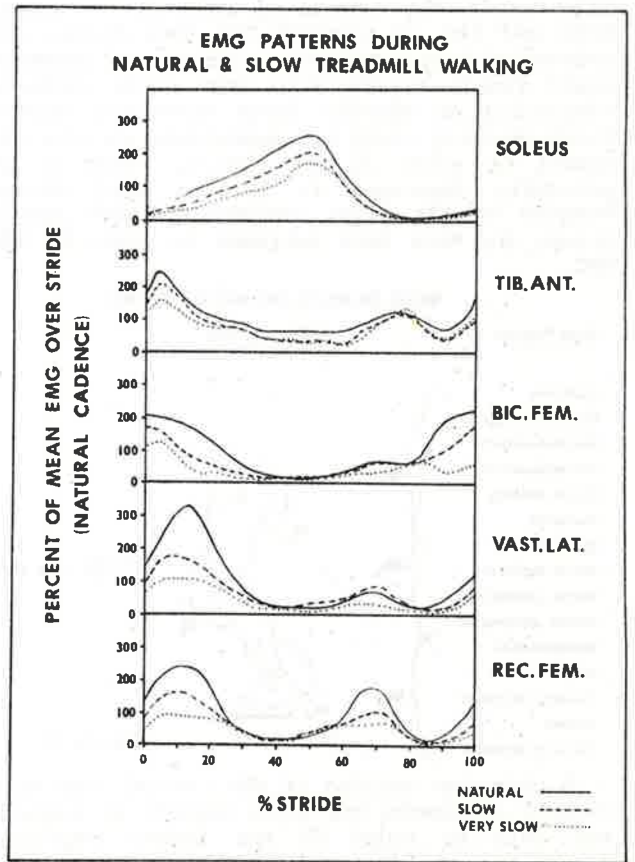

Figure 3

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

Sandra J. Olney, School of Rehabilitation Therapy, Queen's University, Kingston, Ontario, Canada

David A. Winter, Department of Kinesiology, University of Waterloo, Ontario, Canada

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 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:

$$M(t)=k_{11}E_1(t)(1+k_2[\theta(t)-\theta_1]-k_3\omega(t))$$
$$-[k_{12}E_2(t)(1+k_2[\theta_2-\theta(t)]+k_3\omega(t))]$$

when M(t) was instantaneous moment about the ankle; $E_1(t)$ and $E_2(t)$ were instantaneous amplitudes of processed EMG from tibialis anterior and soleus muscles; θ_1 and θ_2 were angles at which static calibrations were conducted to derive static force/EMG relationships for each muscle group; $\theta(t)$ was instantaneous ankle angle; $\omega(t)$, instantaneous ankle velocity; k_{11} and k_{12} were constants relating EMG to force; k_2 and k_3 were constants accounting for force/angle and force/velocity relationships.

In Model 1, the static model, k_2 and k_3 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_3 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 single pass of a second order Butterworth low-pass digital filter varying the cut-off frequency until a correlation with the force signal yielded optimal 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_{11} and k_{12}.

Walking trials included simultaneous collection of 16mm cine film, force plate and EMG data. The film and force plate data were processed and analysed 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_{11} and k_{12} into each model, k_2 and k_3 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 1.6 Hz and the highest 2.0 Hz. Optimal force/EMG relationships derived from the isometric calibration contractions ranged between .92 and .98 with a mean of .97.

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 are shown in Figure 1. The effectiveness in predicting the resultant moment when using 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.

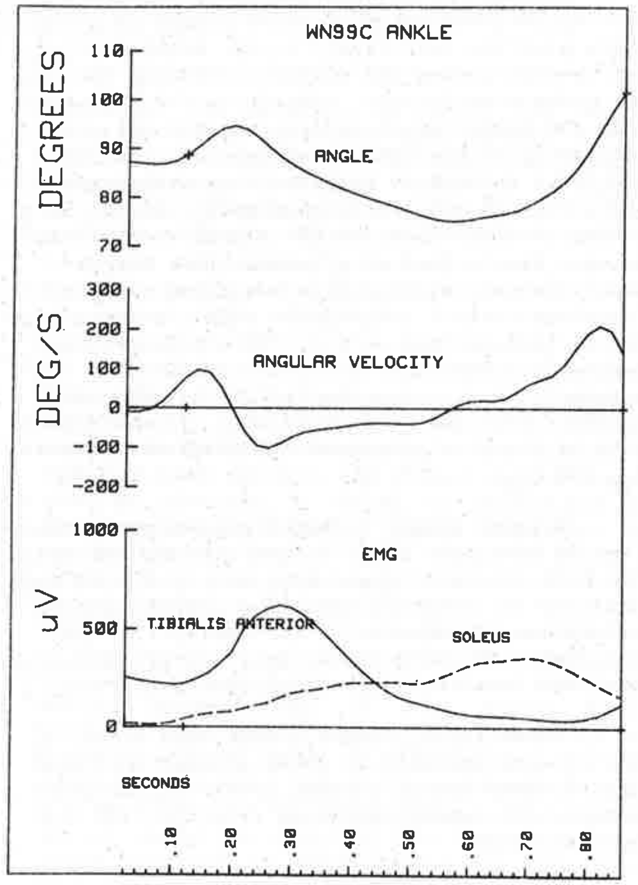

Figure 1. Ankle angle, angular velocity and processed.EMG obtained during stance phase of gait and used in models, Figure 2.

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 1 demonstrates the errors if EMG alone is used; moments 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.

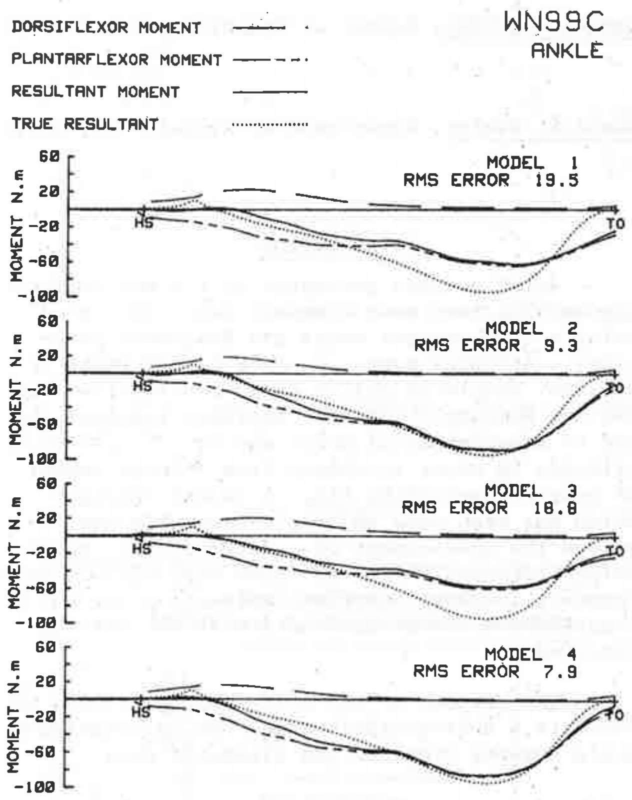

Figure 2. Predicted moments and true resultant moments obtained from each model.

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

SUBJECT	TRIAL	K_2	K_3	RMS ERROR,N.m
WN99	B	.041	.0022	7.5
WN99	C	.033	.0022	7.9
WN79	F	.062	.0023	3.2
WN79	H	.038	.0018	9.5
WM01	M	.020	.0030	6.2
WM01	N	.014	.0037	8.1
WM02	A	.006	.0026	5.6
WM02	B	.023	.0010	7.0

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ACTIVITY OF THE MUSCLES OF THE LEG AND FOOT-FLOOR CONTACT PATTERNS DURING NORMAL WALKING

Joop H.M. van der Straaten, M.D., Ph.D., Department of Anatomy, Katholieke Universiteit Nijmegen, Nijmegen, the Netherlands.

ABSTRACT

Using surface electrode-amplifier modules (van der Lochts et al., 1980) the electromyogram of gastrocnemius, soleus, anterior tibialis, extensor digitorum longus and peroneus muscles of both legs are recorded during gait.

Simultaneously the foot-floor contact patterns are recorded (van der Straaten, 1980).

Twenty normal adult subjects were tested, while wearing their own shoes, shoes with a "negative heel" and shoes of a classical model.

Special attention has been paid to the heel contact time and sole contact time in relation to muscular activity in the three different "shoe-situations".

The results indicate that there is no statistical significance in the rations of heel contact time versus sole contact between the three situations. When these rations are combined with the electromyographic activity of certain leg muscles, some differences can be observed, although statistical significance is only at 1 low level.

The results contradict the frequent stated thesis, that small differences in height of the heels will influence the actual forces in statics and dynamics of the low back. The conslusion of this pilot study is that, if there is any influence on the actual forces, this will be absorbed in the kinematic chain formed by the foot, leg and thigh, so that the low back will not be affected by these minor differences in normal subjects.

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

A.L. Hof and Jw. van den Berg, Lab. for Medical Physics, State University Groningen, Bloemsingel 10, The Netherlands.

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 (3). 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.o. Cavagna and Margaria (1) 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 32/23 yr, mass 91/69 kg, stature 1.92/1.77 m, leg length 1.00/0.94 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 steplength was left free. It turned out not to vary much with slope; subj. A: 0.70-0.76 m, subj. B 0.56-0.63 m. EMG was recorded by means of surface electrodes, ankle position by means of an electrogoniometer. Moment and work of the calf muscles were determined by EMG to force processing (2). For normalization 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 ca. 0.1 J/kg at +20%.

The moment has a maximum, M_{max}, at the onset of push-off, which increases from $M_{max} \approx M_{ref}$ (no peak) at steeply downward slopes to a pronounced peak with M_{max}/M_{ref} = 1.5 to 2.0 at a slope of +20%. The value for 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 8 to 44J (at 0% and +20% respectively); the greater work is largely due to a longer duration of the effective push-off.

DISCUSSION

In spite of their 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.

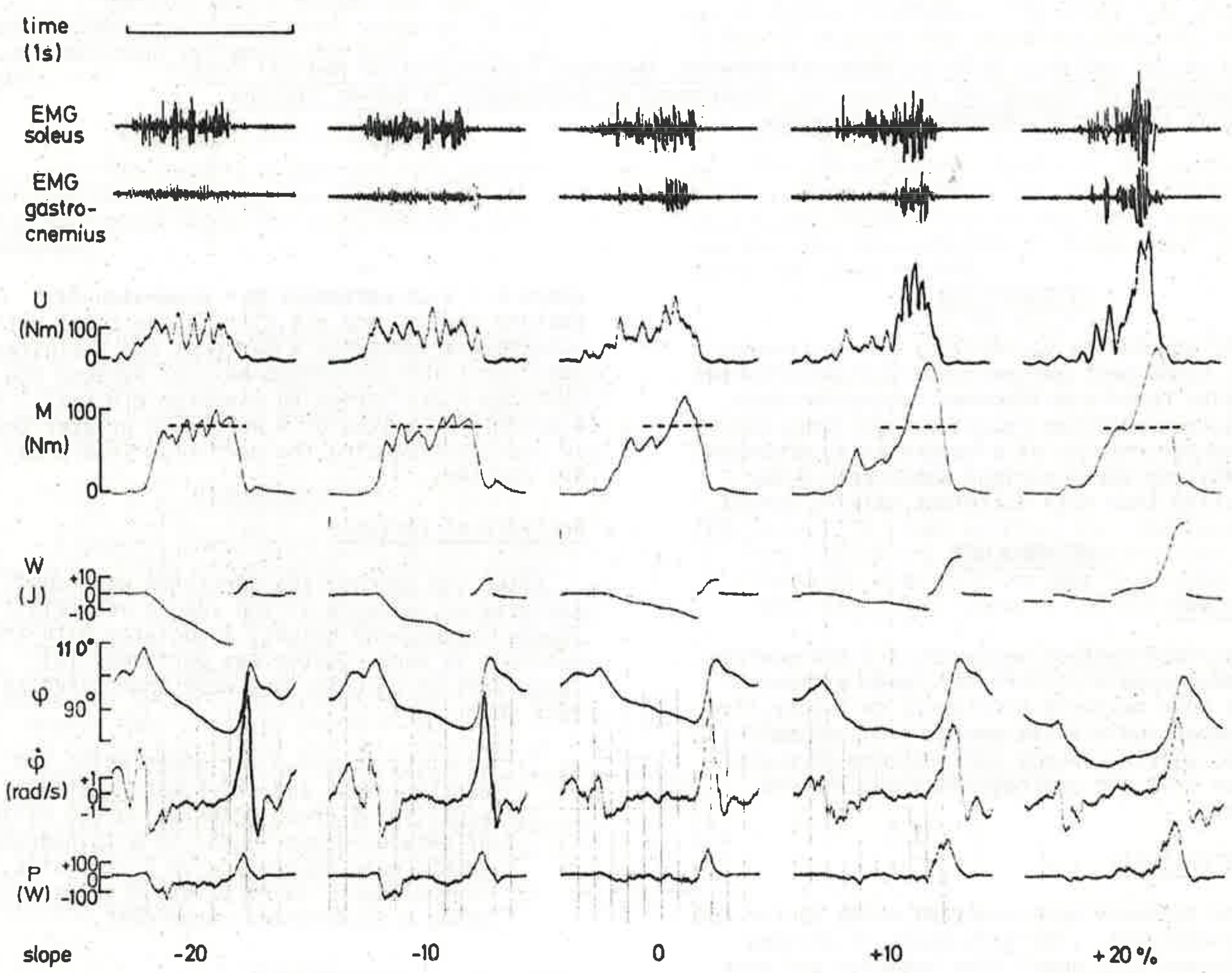

Figure 1. *Recordings at various slopes, subject B. From top to bottom: raw EMGs; U, weighted sum of both rectified and smoothed EMGs (scale in Nm of isometric moment); M, moment obtained by EMG to force processing; W, muscle work = $\int M\dot{\phi}dt$; ϕ, ankle angle (plantarflexion positive); $\dot{\phi}$, angular velocity of ankle; P, muscle power = $M\dot{\phi}$.*

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CHANGES IN ELECTROMYOGRAPHIC PATTERN OF GASTROCNEMIUS MUSCLE WITH GAIT WALKING SPEED

Richard Shiavi and Paul Griffin, Research Service, Veterans Administration Medical Center and Department of Biomedical Engineering, Department of Orthopedics & Rehabilitation, Vanderbilt University, Nashville, Tennessee

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 K+1 instead of K partitions (clusters). The ratio is

$$R = \left[\frac{e(N,K)}{e(N,K+1)} - 1 \right] (N-K+1) \approx F_{M,\ (N-K-1)M} ,$$

where N = # of patterns, M = dimension of feature vector, and e(N,K) = is the total within-cluster variance for N patterns and K clusters. The F distribution cannot be used because the partitions are formed to minimize e(N,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 its second peak over 10. The second peak occurred at K = 7, which means that a partition of 8 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 algorithms. Listed also are the analysis of variance statistics.

1) Equity of Group Variances.

The chi-square values are much less than the 95% confidence level value of 14.07; thus, the group variances for each pattern are equal.

2) Equality of Group Means.

The F-ratio values are much greater than the 95% confidence level value of 2.1; thus, the group means for each pattern are different.

3) Inequality of Group Means

The Studentized range for several pairs of group means encompassed zero at the 95% confidence level. These pairs are indicated by vertical lines paralleling the respective group numbers.

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 60% as speed increases. In fact, at faster speeds the activity commences during late swing. Examine the patterns of Groups 4, 8, 3, 6, and 7. Notice that the main phase of activity always ceases before the end of stance and at the end of midstance (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.

An important exception to this trend is the activity of Group 1; it displays a short burst during midstance. The other two groups (2 and 5) account for anomalous patterns which occur very seldom.

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.

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```
               SPEED        REL  SPEED                                    PERCENT
GROUP  AVE      VAR         AVE    VAR        EMG PATTERNS

  4 |  0.48   0.04        0.30   0.02     |   ********          |        12.63 %
  2 |  0.50   0.05        0.28   0.02     |   *      ***        |         3.16 %
  8   |0.70   0.08        0.42   0.03     |    ******           |        16.84 %
  5 | |0.84   0.14        0.50   0.05     |  ************       |         3.16 %
  3 |  0.88   0.12        0.52   0.04     |  *******            |        32.63 %
  1 |  0.92   0.10        0.57   0.04     |      ***            |        20.00 %
  6    1.27   0.06        0.74   0.03     |********           * |         6.32 %
  7    1.59   0.06        0.92   0.01     |******** **        * |         5.26 %

TEST EQUALITY OF GROUP MEANS

FRATIO         9.06              7.99     DF:K-1=  7,N-K= 87
BGMS         0.8909            0.2949
WGMS         0.0984            0.0369

TEST EQUALITY OF GROUP VARIANCES

CHI SQ         4.96              4.87
```

Figure 1

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REPRODUCIBILITY OF GAIT ANALYSIS

T. NORIMATSU, M. FUJITA, N. MATSUSAKA, R. SUZUKI, H. OKUMURA* AND T. OKAJIMA*.

Department of Orthopaedic Surgery, Nagasaki University School of Medicine
*Department of Radiation Biophysics Atomic Disease Institute, Nagasaki Univ.Nagasaki,Japan

Introduction

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 well 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., 1/36 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 (Fig. 1).
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.

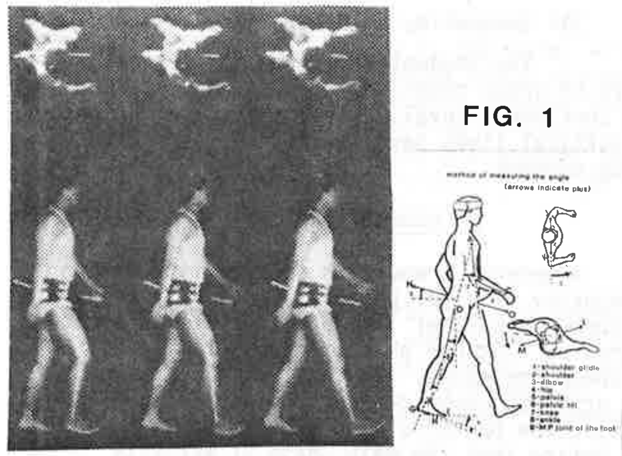


FIG. 1

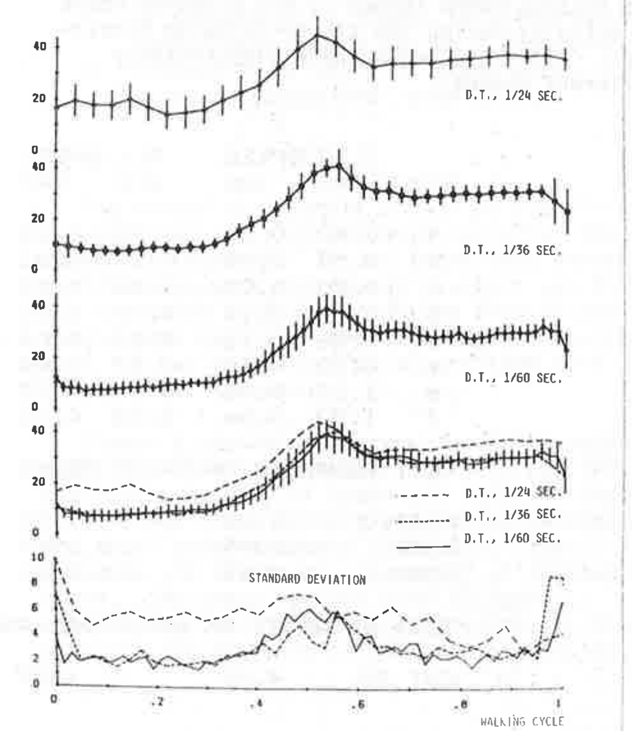


FIG. 2 ANGULAR CHANGES OF M.P. JOINT OF THE FOOT (MEAN PATTERN AND S.D.)

T Norimatsu et al Reproducibility of Gait Analysis

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 durations (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 beak 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.

FIG. 3 INTERPOLATION (simulation) M.P. JOINT OF THE FOOT

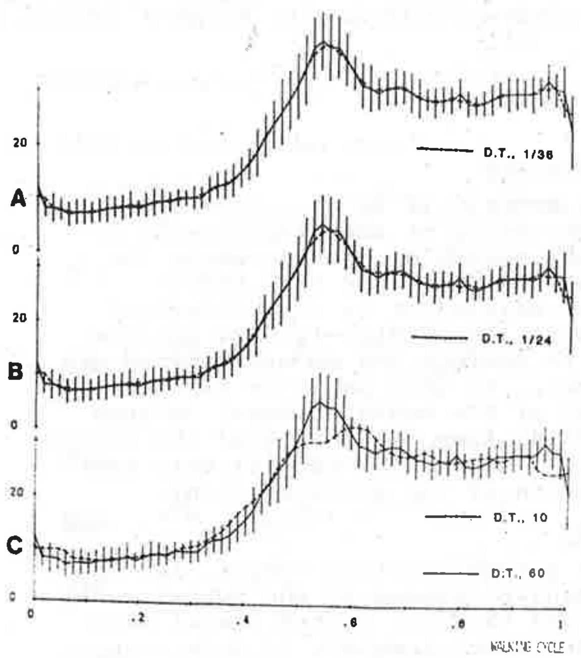
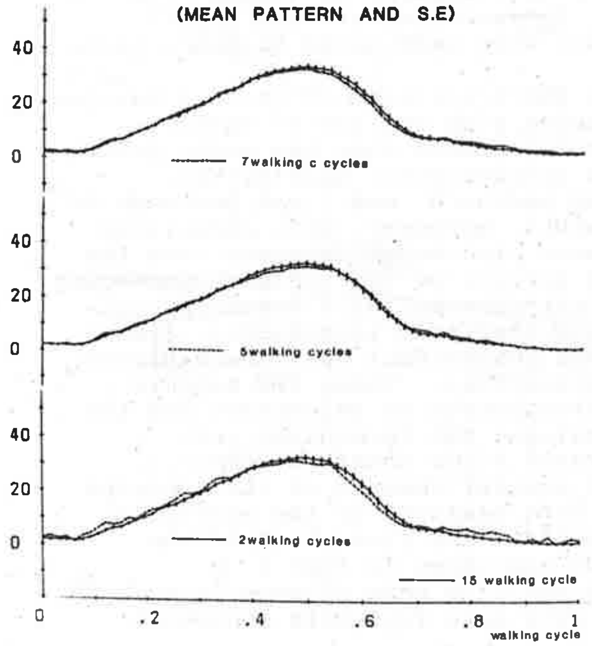


FIG. 4 ANGULAR CHANGES OF ELBOW (MEAN PATTERN AND S.E)



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RELATIONSHIP BETWEEN THE SUBTALAR JOINT AND THE CONTROL OF MEDIAL-LATERAL BALANCE IN WALKING

NOBUOU MATSUSAKA, M. FUJITA, G. CHIBA*, T. NORIMATSU AND R. SUZUKI.

Department of Orthopaedic Surgery, Nagasaki University School of Medicine, Nagasaki, Japan
*Department of Orthopaedic Surgery, Ohmura Municipal Hospital, Nagasaki, Japan

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 singal 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 Matake's force plate.

The EMG activities of various muscles were noted with the use of surface electrodes placed over the motor points of the paravertebral muscle(PVM), gluteus medius(G. med.) and peroneus longus(PL), moreover, with indwelling fine-were electrodes inserted into the muscle bellies of the tibiales 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 90msec.).

The angular changes of the subtalar joint were measured by the electrogoniometer. The flow chart of the methods was shown in Fig. 1.

The subjects were allowed to walk 30 steps with bare feet with customary gait.

Results

In the ground reaction forces the medial-lateral component was deficient in reproducibility of the pattern as compared with the vertical and forward-backward components. In order to investigate further the medial-lateral component the points shown in Fig. 2 were measured. 'T' test was applied to the coefficient of correlation between these points; FL1 was directly proportional to FL2.

When the medial-lateral component had large FL1 and FL2 (Fig. 3A), the EMG of the PL appeared from the early period of the stance phase and the amplitude of the EMG in the TP was lower than that of the type B, and then the EMG of the FDL appeared after heel-off and that of the EHL was hardly found in the stance phase.

On the other hand, when the medial-lateral component had small FL1 and FL2 (Fig. 3B), the EMG of the PL appeared after heel-off and the amplitude of the EMG in the TP was higher than that of the type A, and then the FDL acted from the early period of the stance phase and the EHL acted from the swing phase to the last half of the next stance phase.

In the angular changes of the subtalar joint, the period of the eversion in large medial-lateral component was longer than that in the small component during the stance phase.

Discussion

In the human gait the enter of gravity displaces to the suspending side, and is confined to the inside of the dynamic point of the stance foot (C). (Fig. 4). With the alterating of the distance (D) from the locus of the center of gravity of C, the control of the medial-materal balance in walking should be converted.

When D is longer, in order to

prevent the medial inclination of the body, it is necessary that PL acts from the early period of the stance phase and TP mades its activity weaker after heel contact.

On the contrary, when D is shorter, the medial-lateral component is smaller, and TP, FDL and EHL continue their activities throughout the periods 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.

Fig. 1.

Flow Chart of the Methods

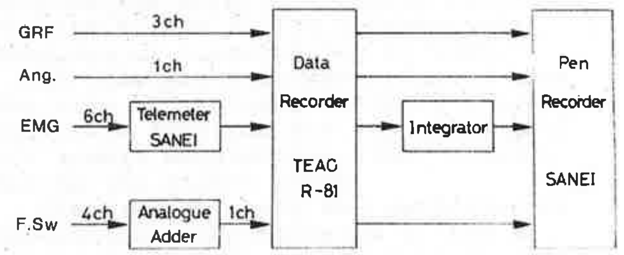


Fig. 2.

Points of Measurement in F.R.F. & F.Sw.

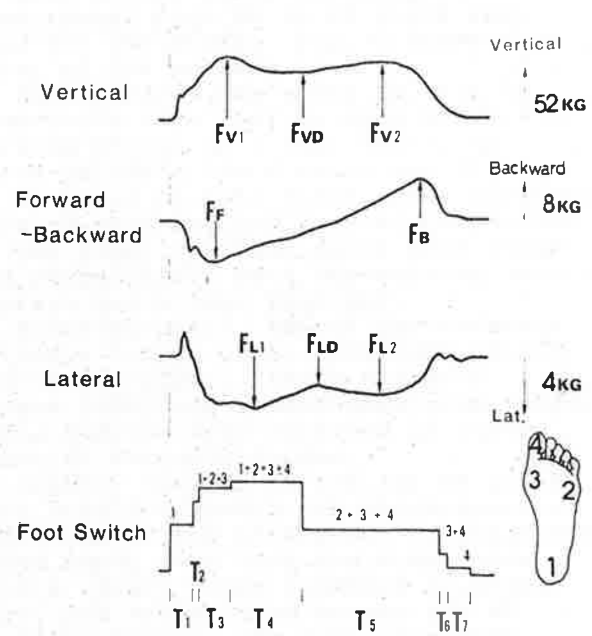


Fig. 3.

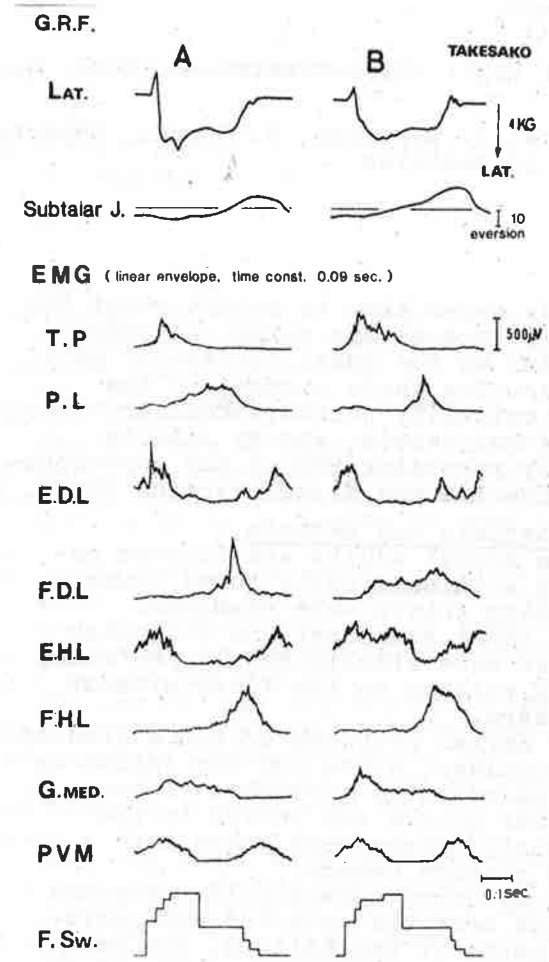
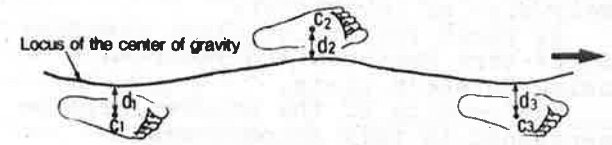


Fig. 4.



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ARM SWING AND BODY ROTATION IN WALKING

Hiroaki Fujii, N. Norimatsu, M. Etoh, N. Matsusaka, M.Fujita, S. Furuta, G. Chiba,

H.Mimura, T. Teramoto, R. Suzuki, Department of Orthopaedic Surgery, Nagasaki University School of Medicine

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 footwears.

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

2) Electromyographically examined muscles were the anterior and posterior parts of the deltoid, the pectoralis major, the upper part of the trapezius, the biceps brachii, the triceps brachii and the lumbar paravertebral muscles. These action potentials were integrated respectively with an integrater.

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

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

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MOTION AND ROLE OF THE MP JOINT OF THE FOOT IN LEVEL WALKING

M. FUJITA, M. MATSUSAKA, T. CHIBA*, T. NORIMATSU, T. HATASHI AND R. SUZUKI.

Department of Orthopaedic Surgery, Nagasaki University School of Medicine Nagasaki, Japan *Ohmura Municipal Hospital, Nagasaki, Japan

Introduction

The function of the metaparsaphalangeal 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 was proposed that the role of the toes was only accessary and their function was to act as roll off agents maintaining floor contact during the stance phase.

The present study was performed to inverstigate the function of the MP joints in various types of level walking which were composed of customary, slow, fast and longer stride styles of walking.

Materials and Methods

Five normal adult volunteers, whose ages ranged from 20 to 35 years were used for the study. Fig. 1 shows flow chart of the methods.

Each subject was asked to walk 10 times with bare feet in each of the walking styles on a flat table, in which Matade's force plate was fixed. By stepping on this force plate vertical, forward-backward and lateral components of the floor reaction force were recorded respectively by a pen-writing oscillograph and a tape recorder.

Simultaneously, EMG of the extensor hallucis longus (EHL), extensor digitorum longus (EDL), flexor hallucis longus (FHL) and flexor digitorum longus (FDL) muscles were recorded by intermuscular wire-electrodes.

Angular changes of the Ist MP joint were measured by our new electrogoniometer consisting of a piece of stainless steel about 10cm long and 0.6cm wide with a strain gauge attached to middle part, 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 simulataneous records of EMGs, MP angle and forward-backward component of floor reaction forces.

The Ist 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 Ist MP joint.

From the above data it would seem that the toes do not participate in the action of push off in customary malking. During 10% of the stance phase the contralateral 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 toes in slow, fast and longer stride styles of walking. Main points were measured to compare with customary walking and other styles of walking (Fig. 3).

T5 and T6-7 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 Ist MP joint. F-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 forces. T5, T6-7, MP and F-B of each walking style were increased significantly according to velocity with the exception of T5 in the longer stride style of walking. T-FHL and T-FDL were increased significantly compared with the customary walking style with the exception of T-FHL in the fast walking style.

In the fast walking styles, T5, T6-7, T-FDL, MP and F-B were increased significantly, it was supposed that the toes would be required to exert propulsion forces during the end of the stance phase.

In the slow walking styles T-FHL and T-FDL were increased but T5, T6-7, MP and F-B were decreased significantly. The above data suggested that the toes would not be required as progulsion forces but stabilizing forces to maintain floor contact.

In the longer stride walking styles T6-7, T-FHL, T. FDL, MP and F-B were increased significantly. It was supposed 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 accesory 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 propulstion forces.

In slow walking styles the toes are not required as propulstion forces but as stabilizing forces.

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Fig. 1.

Flow Chart of the Methods

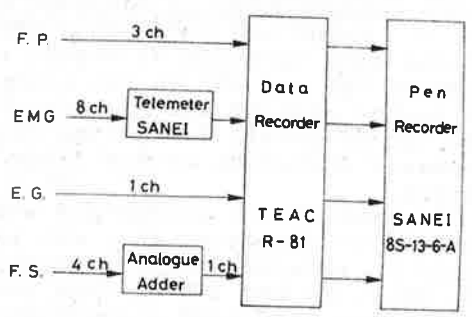

Fig. 2.

Simultaneous Records of EMGs, MP angle & Floor R.F.

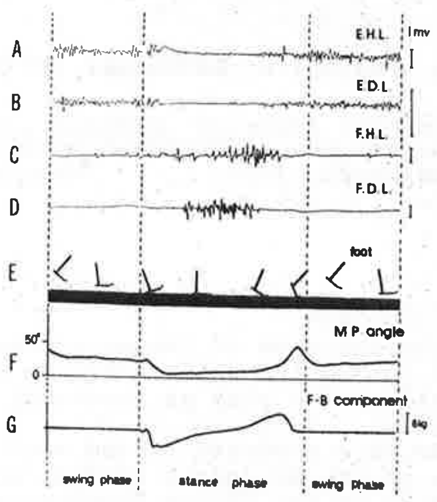

Fig. 3.

Points of Measurement in EMG, F.Sw., MP angle and Floor R.F.

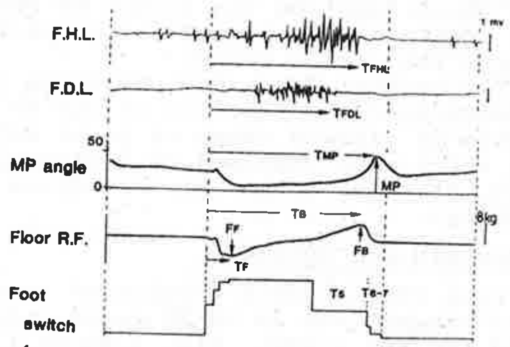

Fig. 4.

Comparison with Customary Walking and Other Walkings

	Fast		Slow		Longer s.	
		α		α		α
T_5	♦	++++	♦	++++	♦ ♦	—
T_{6-7}	♦	++++	♦	++++	♦	++++
T_{FHL}	♦ ♦	—	♦	++++	♦	++
T_{FDL}	♦	++	♦	++++	♦	++++
MP	♦	++	♦	++++	♦	++++
T_{MP}	♦	+	♦	+	♦	++++
F_B	♦	++++	♦	++++	♦	++++
T_B	♦	+	♦	++++	♦	++++
T_B-T_{FHL}	♦ ♦	—	♦	++++	♦	++++
T_B-T_{FDL}	♦	++	♦	++++	♦	++++
$T_{MP}-T_B$	♦ ♦	—	♦	++++	♦ ♦	—

$\alpha \geqq 0.1$ — $0.1 > \alpha \geqq 0.05$ +
$0.05 > \alpha \geqq 0.01$ ++ $0.01 > \alpha \geqq 0.005$ ++
$0.005 > \alpha \geqq 0.001$ ++++

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ANALYSIS OF TIPTOE GAIT

R. SUZUKI, N. MATSUSAKA, M. FUJITA, S. FURUTA, J. KAYANO AND T. NORIMATSU.

Department of Orthopaedic Surgery, Nagasaki University School of Medicine
Nagasaki, Japan

Introduction

Walking on tiptoe is used for training of the limb muscles in orthopaedics 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 stick picture camera which can photograph 120 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 tubercle 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 Matake's force plate which was embedded in the platform for walking examination.

5. Action potentials of following sixteen muscles were recorded on a pen writing oscillopaper using surface and/or fine wire electrodes and linear envelops were made.

1) M. sacrospinalis
2) M. rect. abdom.
3) M. glut. max.
4) M. glut. med.
5) M. rect. fem.
6) M. biceps fem.
7) M. tib. ant.
8) M. gastrocn.
9) M. peron. long.
10) M. tib. post.
11) M. flex. dig. long.
12) M. flex. hall. long.
13) M. ext. dig. long.
14) M. ext. hall. long.
15) M. flex. dig. brev.
16) M. abd. hall.

Results

1. Range of motion of the ankle joint during the stance phase was as follows:

Type A : 10° of dorsifles. — 20° of plant. flex.
Type B : 0°— 25° of plant. flex.
Type C : 25° — 40° of plant. flex.

Among three types of gait, remarkable differences of hip and knee motions during walking were not found.

2. Breadth of the forefoot fairly increased during stance phase in every type of gait. However, in the type A, the maximum peak existed at the thrust phase, while in the remaining two types, peak of the reception phase was higher than that of the thrust 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

R Suzuki Analysis of Tiptoe Gait

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, sacrospinalis, glut. max., biceps fem. and abd. hall. 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 gastrocn. 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 peron. 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 seemed 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 these musclular activities. However, the length and breadth of the foot were scarcely influenced by walking types in the normal adults owing to rigidity of the arches.

The results of the above mentioned experiments are considered to be helpful for evaluation of equinus deformity of the foot and planning of muscular training.

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ANALYSIS OF THE GAIT IN BOYS AND GIRLS WITH THE VECTOGRAM TECHNIQUE: FIRST RESULTS

Rodano Renato° and Tosoni Anita°°; °Centro di Bioingegneria, Politecnico di Milano-Fondazione Pro Juventute Don Gnocchi, Milano; °°Scuola Media Rinascita Livi, Milano, Italy

INTRODUCTION

The human gait has been investigated by many authors with different techniques. Many have been the goals of these analysisis: to set up appropriate instrumentations, to investigate and differentiate normal and pathological patterns, to know the mechanisms of motor coordination. An important field of study is the gait of children as the knowledge of the biomechanical modification produced by the growt, play an important rule for the prevention of pathologies of the muscolo-skeletral apparatus. The aim of this study is to fournish a contribution in this area throught 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 declaired light impairments to the lower limbs. The children were requested to walk at the fixed cadence of 108 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 (Boccardi et al., 1977; Cova et. al., 1980). The main characteristics of normality are so defined: a) high repeatibility of the patterns of each subject; b) two equilibrated maxima localized during the impact and the pushing phases, and a minimum between them; c) a continous 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 simmetry 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 repeatibility 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 conduced without considering the physical conditions. Fig. 1b shows those of a boy affected from bilateral talalgia; the uncorrect 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, Fig. 1c is an example of two of them.

The children with normal vectograms in both the legs were eight and two of them showed an appreciable simmetry of gait while six were characterized by various asimmetris due to the

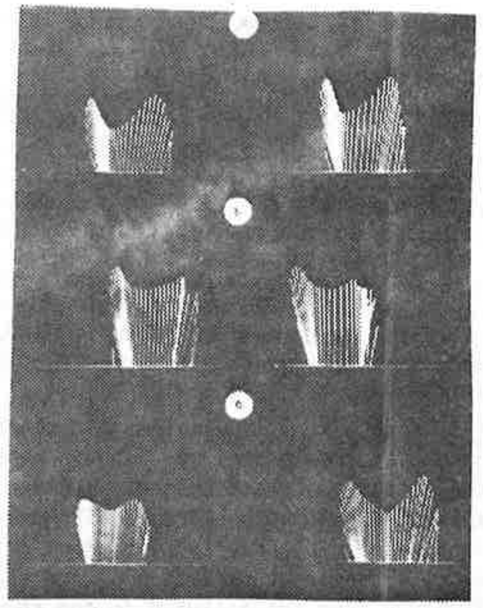

Fig. 1 Vectograms obtained in the sagittal plane. Cadence of walk 108 st/min. Each vector every 20 msec. a) normal b) pathological c) border line.

amplitude, inclination and distribution of the vectors.

A statistical approach has been attempt in order to define by a mathematical point of view the information contained in the vectograms and to well identify the meaning of the border-line patterns. The method utilized was, in the ambit of factorial analysis, those of the principal components.

In our analysis the considered variables were those easier to quantify and, at the same time, characterizing the vectograms: amplitude of the maxima and minimum (normalized with the body weight), time duration of the stance phase.

For an easy screening was utilized only the first equation of the principal components explaining the 44% of the total variability. The numerical values obtained were not able to discriminate normal from pathological patterns, contrarily to the vectogram's moreover no further information were introduced on the border line.

CONCLUSIONS

The vectogram's techniques has been demonstrated useful instrument for a mass analysis of human gait. The method fournish 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 simmetry 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 continous change of the muscolo-skeletal apparatus.

By observing the relatively low percentage of vectograms skeduled as normal (50%) and the consistency of the groups defined border-line (34%) rises a question: are the border-line indicating unknown dynamics of normal gait or earily advices 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 conduced without significative results. Three are the possible explanations: 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.

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QUANTITATIVE CHARACTERISTICS OF MUSCLE WORK IN CASE OF PATHOLOGICAL GAIT

A.S. Vittenzon, A.V. Sarantsev, USSR, CRIP

At present alongside the long-known methods of studying pathological gait a new one may be suggested that defines the dependance of a number of parameters of integrated muscle electrical activity from the gait speed (tempo). Such parameters include the summary electrical activity during the cycle, which reflects the relative muscle work in the process of walking, its variability as to step sequence, which shows the peculiarities of controlling the motor process, and the relation between periods of electrical activity and rest, which allows to evaluate the effectiveness of work of muscle forces during the locomotor cycle.

All the stated dependances in case of normal gait may be described by a conjuction of functions with a minimum at the arbitrary (medium) tempo.

In case of pathological gait in patients with injuries of supportive-locomotor system (patients with above-knee and below-knee prostheses, with infantile cerebral paralysis) the character of these dependances change to a great extent. In this case we observe decrease of values of those coefficients that define the steepness of curves and the position of their minimums relative to the axis of coordinates.

By analyzing these changes we may obtain quantitative characteristics of muscle work for pathological gait.

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BALANCE TRAINING BY MEANS OF BIOFEEDBACK

Rajko Turk, Tadej Bajd*, Helena Benko, and Janez Šega*, Rehabilitation Institute and *Faculty of Electrical Engineering, Edvard Kardelj University, Ljubljana, Yugoslavia

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 simple walking pattern. First experiments of paraplegic patients walking were already performed, most of them in the parallel-bars. 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 proprioreceptors 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.

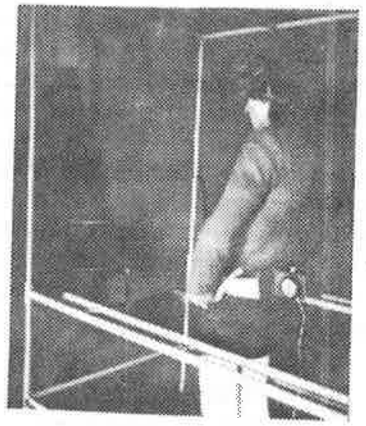

Fig.1

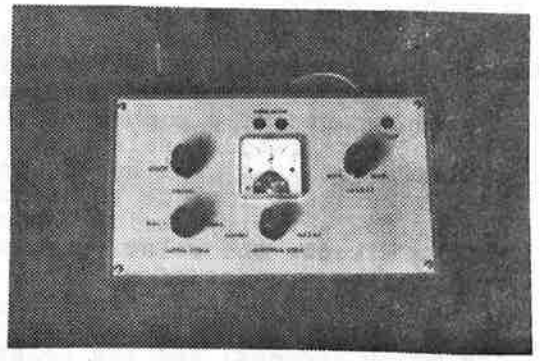

Fig.2

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 of 2 kΩ, range of motion ± 45°,accuracy ± 1% and frequency 3,2Hz. 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 denoting either leaning forward or backward.

METHODS

During biofeedback training the patient is standing in a special supporting frame providing the arm support. His knees are locked either by stimulation or mechanical braces. The precision pendulum is attached to the sacral part of the body. One loud-speaker is placed in front of him while the other behind him. When the patient is leaning forward the first loud-speaker is loud, when he is leaning backward the other is activated. The frequency of the tone is proportional to the angle from the central position of the body.

RESULTS

Four paraplegic patients participated in the program of the balance training by the biofeedback: patient R.Š. with complete paraplegia at T-9 level, patient F.D. having complete T-9,10 spinal cord lesion, patient M.Č. 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 occured 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 loudspeakers were silent.

The patient R.Š. 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 5 s. This patient didn't succeed to stand, without the arm support. The second patient F.D. started with 13 s with one arm support and ended with 6 min 35 s after twenty days of

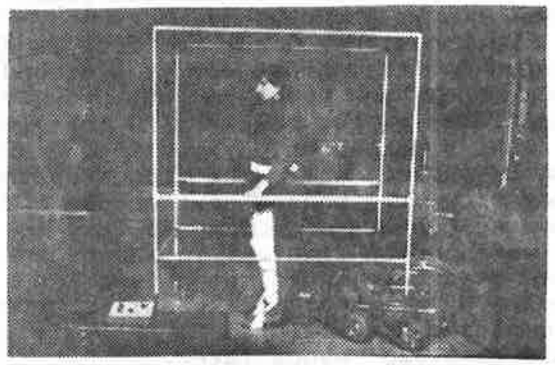

Fig.3

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.Č. 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 40 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 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.

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COORDINATION OF THE PERONEUS LONGUS AND FLEXOR HALLUCIS LONGUS MUSCLES DURING PUSH OFF.

Finn Bojsen-Møller, Anatomy Department C, University of Copenhagen, Denmark.

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 metatarsal bones, 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 (Bojsen--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 tuning 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 registered with an accuracy of 5-10 msecs (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 strobo-flash 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 retrieved 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 fleshy 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 msecs, while the strobe light will show the position of the leg and the syncronism between the four light tracks each 200 msecs.

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 show the phases of the push-off 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 82o msecs later.

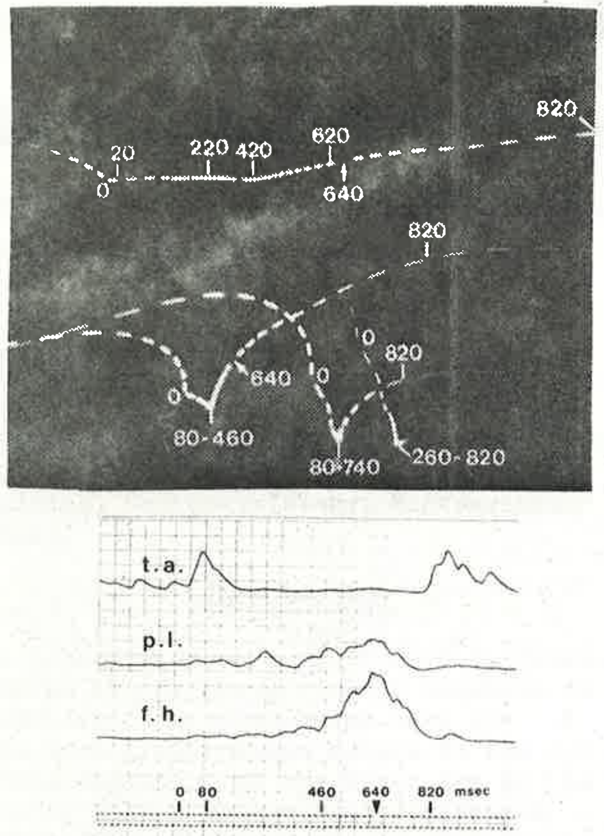

Fig.1. Still photo with four light tracks and corresponding electromyogram showing the stance phase of a left foot. The light tracks are formed by diodes which are supplied by the 5oHz signal shown on channel 4 on the EMG. Time O is set at the first deflection of the ligntttracks caused by heel strike. The five Hz stroboflash shows the position of the leg and foot at 20, 220, 420, 620 and 820 msecs. The missing signal which in this case is found at 640 msecs 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 460 and 740 msecs, 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).

With the contact phase normalized into 100 time-units, heel rise occurred in eversion at 47% (fig. 2). Activity of the peroneus longus was registered from 18% - 91%, and of the flexor hallucis longus from 35% with an increase at 71%, which is approximately 140 msecs before the metatarsophalangeal joints are lifted and before the great toe starts to return from its dorsiflexed position. In inversion heel rise occurred earlier (at 35%) while activity of the peroneus longus and flexor hallucis longus muscles were found later (from 38% - 91%, and from 61% - 93% respectively). The evertor shows an early, transient activity.

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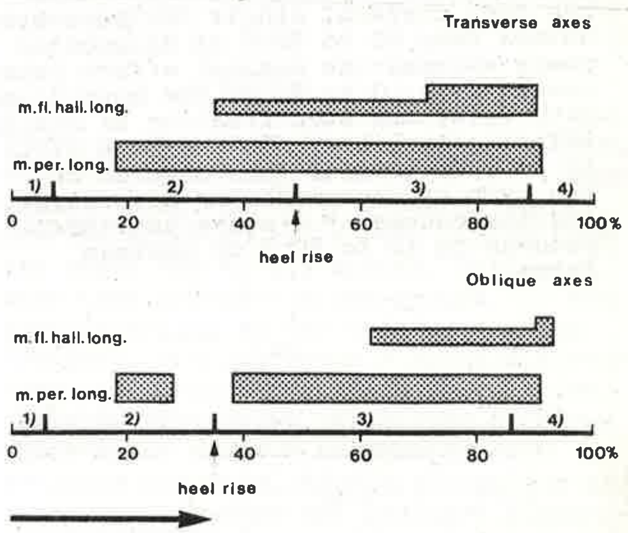

Fig.2. Diagram showing the coordination of the flexor hallucis longus and the peroneus longus muscles during ten push offs in eversion (transverse axis) and ten push offs in inversion (oblique axis). Same individual. The contact phase is normalized into 100 time units.

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AUTOMATIC ANALYSIS OF EMG IN CLINICAL **DIAGNOSIS**

J.Kopeć, I. Hausmanowa - Petrusewicz, Department of Neurology Medical School, Warsaw

Since four years automatic analysis of ANOPS computer has been used in our Department. The analysis is given in a form of 5 histograms describing a complete EMG examination. Three paraments evaluate single motor unit potential (MUP) and two other maximal effort pattern. Using different fraction 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 fraction of maximal effort at which single MUP are still to be identifield. This depends on the type of neuromuscular disorders and the course of disease and corespondens to 10 to 50 % of maximum force.

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EMG FINDINGS AND AIDS IN THE PREVENTION OF PERIPHERIC COMPLICATIONS IN THE HYPOTONIC EVOLUTION OF VASCULAR HEMISYNDROMES

ROBERTO CASALE,WORK MEDICINE FOUNDATION OF PAVIA,REHABILITATION CENTER,SERVICE OF CLINICAL NEUROPHYSIOLOGY,MONTESCANO,ITALY.

EMG FINDINGS AND AIDS IN THE PREVENTION OF PERIPHERIC COMPLICATIONS IN THE HYPOTONIC EVOLUTION OF VASCULAR HEMISINDROMES.Roberto Casale,Work Medicine Foundation,Rehabilitation Center,Service of Clinical Neurophysiology, Montescano, Italy.

INTRODUCTION

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 hypotonic evolution of the upper arm in patients with cerebrovascular diseases (CVD) featuring a so called drop shoulder (Colombo et al.,I970).Investigations have been already carried out by several authors in patients with CVD to spot the presence of "sick motoneurones"(McComas et al.,I973).Expecially American literature suggests a peripheral involvement (Moskowitz and Porter,I963;Goldkamp,I967; Bhala,I969;Kruger and Waylonis,I973;Zalis and al.,I976).On the other hand we can find in literature data suggesting a not complete aggreement upon the presence of lower motoneuron diseases (Isch,I968; Alpert,I97I,I973).A certain dishomogeneous criteria in the patients's choices 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 Caccia et al.(I976) gave notice about this but without taking any correlation with the electrophysiological data.Zalis and al.(I976),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 differents tonic evolutions following acute CVD.

METHOD

Subjects.The study involved 20 patients, male and female,aging from 40 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 (Soriani, I972):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 patients 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 (Buchthal and Pinelli,I95I,I953;Buchthal I957;Gassel,I964;Arrigo,I982). Axillary nerve and its supplied deltoid muscle were selected for their anatomical peculiarities.

RESULTS

Casale Preventionof Peripheric Complications in Hypotonic Evolution of Vascular

Data obtained for each sides,were collected in the following tabels:

PATIENTS		B.M.	G.S.	B.I.	P.B.	P.L.	N.M.	G.R.	P.E.	R.C.	B.B.
SPONT. ACTIVITY		FIB.	FIB.	ABS.	FIB.	ABS.	FIB.	ABS.	FIB.	FIB.	FIB.
MUAP	DUR.	ABS.	ABS.	20	ABS.	17	12/16	14	14	ABS.	ABS.
	AMP.	-	--	2	--	1	26	1	4	--	-
	SHAPE	--	--	POL.	--	POL.	POL.	POL.	POL.	POL.	-
MAX. VOL. EFFORT	AMP.	-	--	2	--	1	6	1	4	--	--
	SHAPE	-	--	1/2	--	1/2	2/3	1/2	1-	--	-
MOT. COND. MEAS.	DISTAL LAT	91	9	5.7	5.5	5.1	5.6	4.6	4.9	6.2	5.9
	AMP.	05	04	1	06	2	6	2	5	1	1
	SHAPE	IRR.	IRR.	POL.	IRR.	POL.	IRR.	IRR.	IRR.	IRR.	IRR.

HYPOTONIC EVOLUTION
PLEGIC SIDE

PATIENTS		S.M.	G.S.	B.I.	P.B.	P.L.	N.M.	G.R.	P.E.	R.C.	G.B.
SPONT. ACTIVITY		ABS.	ABS.	ABS.	ABS.	ABS.	ABS.	ABS.	ABS.	ABS.	ABS.
MUAP	DUR.	4\12	4\14	4\10	3\10	3\12	4\12	3\10	3\12	3\12	4\14
	AMP.	04\1.5	04\1.5	04\1	04\2	04\2	04\15	04\2	06\2	04\15	06\15
	SHAPE	S+P<12	S	S	S	S	S	S	S	S	S
MAX. VOL. EFFORT	AMP.	4	5	2	5	6	3	3	4	2	2
	SHAPE	AR	AR	AR	SI	AR	SI	AR	SI	AR	AR
MOT. COND. MEAS.	DISTAL LAT	4.8	5.1	4.1	3.8	3.7	4.1	3.8	3.7	4.9	4.5
	AMP.	6	5	4	4	6	10	10	8	5	4
	SHAPE	POL	POL	POL	S	POL	IRR	POL	POL	IRR	POL

HYPOTONIC EVOLUTION
HEALTHY SIDE

PATIENTS		R.S.	C.B.	F.L.	N.A.	R.M.	C.C.	F.Z.	E.M.	F.D.	M.C.
SPONT. ACTIVITY		ABS.	ABS.	ABS.	FIB.	ABS.	ABS.	FIB.	ABS.	ABS.	FIB.
MUAP	DUR.	4\14	3\12	4\14	4\16	3\12	3\12	4\14	3\12	3\12	4\14
	AMP.	08\2.5	06\15	04\15	08\25	04\15	04\15	08\2	04\15	04\2	04\2
	SHAPE	IRR	Pol	IRR	Pol	S	S	Pol	S	S	Pol
MAX. VOL. EFFORT	AMP.	4	2	4	6	2	3	3	2	4	1
	SHAPE	AR	SI	AR	AS	SI	SI	AR	SI	SI	AR
MOT. COND. MEAS.	DISTAL LAT	4.4	4.6	4.7	5.9	4.4	5.2	4.9	4.8	4.7	5.8
	AMP.	4	3	3	5	4	2	5	2	3	2
	SHAPE	IRR	IRR	IRR	POL	IRR	POL	IRR	POL	POL	IRR

HYPERTONIC EVOLUTION
PLEGIC SIDE

PATIENTS		R.S.	C.B.	F.L.	N.A.	R.M.	C.C.	F.Z.	E.M.	F.D.	M.C.
SPONT. ACTIVITY		ABS.	ABS.	ABS.	FIB.	ABS.	ABS.	ABS.	ABS.	ABS.	ABS.
MUAP	DUR.	4\12	4\12	3\12	3\14	3\12	4\12	3\12	5\14	3\12	3\12
	AMP.	04\15	04\2	04\1.5	04\25	04\1	06\2	04\1.5	04\2	06\2	04\2
	SHAPE	S	S	S	S	S	SM	S	S	SM	SM
MAX. VOL. EFFORT	AMP.	2	4	4	3	3	5	4	2	2	3
	SHAPE	AR	SI	SI	SI	AR	SI	SI	AR	AR	SI
MOT. COND. MEAS.	DISTAL LAT	4.1	3.8	3.6	4.7	4.1	4.9	4.8	3.9	4.2	4.7
	AMP.	6	10	4	10	12	8	6	4	3	2
	SHAPE	IRR	S	TRI	IRR	S	S	IRR	S	S	S

HYPERTONIC EVOLUTION
HEALTHY SIDE

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.I8+I.56 (Hypotonic side) against 4.23+0.55 (Healthy side) with "t" test=I7.I and p=<0.0005 was carried out.We had also data as much significant as the previous in the Hypertonic patients (II° Group) with a mean values of 4.92+0.5 (Plegic side) against mean values of 4.28+0.45 (Healthy side) with "t" test=5.3I and p=<0.0005Student "t" test was also calcolated between the mean values of the affected sides of both groups with "t" test=2.I9 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 sickening 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 30% of the hypertonic ones.No evidence of controlateral healthy side of both groups was noted. The presence of pathological data on the affected sides of both groups is according to data and theories present in literature as trophic factor,alteràtion in blood supply,transsynaptic changes.But these theories did not explain why alterations are more evident in the hypotonic evolution and,more, there is evidences 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 (Casale et al.I982).So EMG screening after acute CVD could be of great importance in order to evaluate and prevent this kind of complications, with early application of orthesis and appropriate choice of FKT strategy.

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CUTANEUS TEMPERATURE AND SKIN RESISTENCE IN DIFFERENTIAL DIAGNOSIS OF PERIPHERAL NERVE LESIONS (PRELIMINARY REPORT)

Kadoić, V., Žagar, M., Fronjek, N., Kolaj, M., Jušić, A.

Center for neuromuscular diseases, Zagreb, Yugoslavia

Some peripheral nerves contain more sympathetic fibres than others. This is particulary 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 vasoconstrictor paralysis. "Warmer phase" is lasting for some weeks. In incomplete irritative 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 sudomotors fibres are also involved, sweating is reduced and skin becomes dry.

In order to record loss of vasomotor function the temperature was measured by DISA Biological thermometar with skin and intramuscular thermosensors.

In order to record the function of sudomotor sympathetic fibres the skin resistence was measured. This parametar is significantly increased after dennervation. The measuring instrument Bio-R-Metar and recording devices were constructed by ing. M. Kolaj.

A total of 50 patients had skin temperature and skin resistence measurements. 6 patients had median nerve, 9 had ulnar nerve and 4 patients had plexus brachialis lesion. Two patients had reflectory sympathetic dystrophy. Skin temperature and skin resistence 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 2,2°C more. The skin resistence is higher on affected side. In case with motor fibres dammage with nerve potential preserved, differences in temperature and skin resistence were not significant.

Representative case: Four months ago trauma in fossa cubiti, now paresthesie 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 resistence for 20 kO higher.

II - Patients with ulnar nerve lesions

The findings of temperature and skin resistence are different in ulnar nerve lesions. In complete denervation skin temperature and intramuscular temperature, as well as skin resistence may be normal. In incomplete lesions the skin is warmer for months.

Representative case: Tingting and atrophy of interosseal muscles due to cubital channel syndrom. EMG: single abortive motor unit potential and fibrillations. Nerve potential absent. Temperature of hypothenar and thenar was for 0,5 - 0,8°C higher than on contralateral normal side. Skin resistence was for 2-10 kO lower on affected side. No visible skin changes.

III - Patients with plexus brachialis lesions

Case: Weakness of hand after Ca mammae 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, hypothenar temperature for 2,8°C lower. Skin resistence for 16 kO higher. Skin of the hand was dry, livid and atrophic.

IV - Reflex sympathetic dystrophy

Cutaneus temperature and skin resistence measurment are useful in differential diagnosis of organic causalgia against psychogenic symptoms.

Case 1.: Three months ago third finger lesions. Nowaday paresthaesie and pain at hypothenar, extending to axila. Skin of hypothenar is cold and wet. EMG and nerve conduction normal. Skin temperature of thenar for 1,5°C higher, of hypothenar for 1°C higher on affected side. Temperature of m. abductor digiti V for 2°C increased. Skin resistence over the hypothenar for 9 kO lower, on thenar for 17 kO lower, than on oposite side.

Case 2.: Cubital tunnel syndroma, operated twice, second time with postoperative suppuration. 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 2,2°C, on hypothenar for 1,2°C decreased against the healthy side. Skin resistence in thenar and hypothenar for 20-30 kO 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 prolongated bleeding was observed during the EMG analysis.

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EVOKED ELECTROMYOGRAPHIC OBSERVATION OF INTERFASCICULAR NERVE GRAFTING

Yoshiki Yamano, M.D., Assistant Professor, Dept. Orthop. Surg., Kawasaki Medical School, Kurashiki, Okayama, Japan

Specific object of study

To elucidate nerve, especially motor nerve, regeneration and nerve recovery after interfascicular grafting, I 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 (M 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 electrophysiological 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 peroneal nerves of 40 adult rabbits were exposed aseptically under anesthesia, a section of peroneal nerve 2.0 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 peroneal 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 peroneal nerves were taken out, and strained with myelin stain. The axon size, the thickness of myelin as well as the number of regenerating nerve fibers were compared in the graft site and distal portion of the peroneal 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, M waves show mostly a polyphasic tendency with a MCV of less than 30 m/sec. In the case where a simple 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 average 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 pattern was mostly polyphasic and in general of a low amplitude, and distal motor latency was prolonged. There 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±3.8 m/sec in average. By 12 months the average MCV in the 12 cases increased to 44.3±3.4 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 ±4.5 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 high amplitude wave has a low 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 micron and 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 myelinization is fair.

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

Conclusion

Study of the M wave yields useful information on the regeneration of the motor nerve. Picking up the M wave by concentric bipolar needle electrodes enables more precise analysis of the wave.

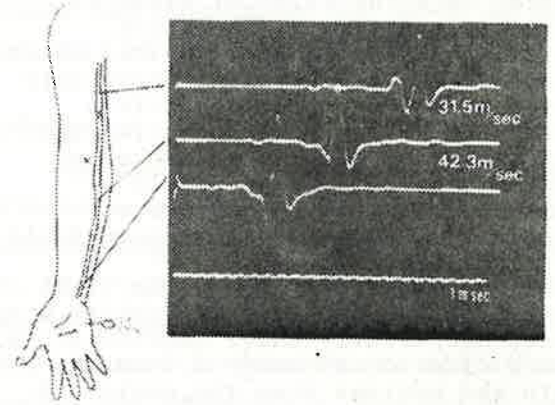
The M wave recorded in this manner is of short duration, low amplitude, and a so-called nascent unit like wave in the initial stage of nerve regeneration. With recovery it undergoes change to a polyphasic wave, and then to a still simpler polyphasic wave, and when the nerve matures to a certain extent,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

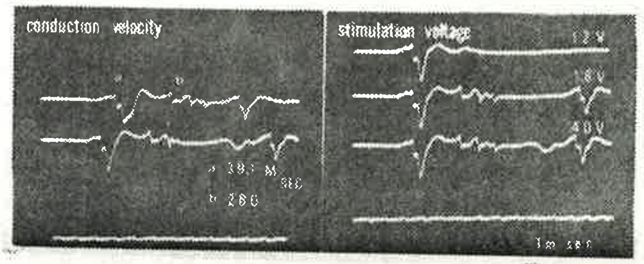
The MCV of high amplitude waves is faster than that of polyphasic waves and so regenerating fibers showing high amplitude waves would seem to be more mature nerve fibers. Recovery of MCV after nerve grafting was 65-70% of normal.

In the present study, the number of large fibers in the normal peroneal nerve is 35.0% and the percentage of large fibers in the graft side at 24 postoperative months is 24.3%, which corresponds to 69.0% 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.





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DEVELOPMENT OF A GRADING SYSTEM OF CLINICAL AND EMG AMELIORATION DURING REHABILITATION THERAPY IN PATIENTS WITH LESIONS OF THE PERIPHERIC NERVOUS SYSTEM (PNS).

R. CASALE; Rehabilitation Medical Center, Service of Neurophysiopathology-27040

E.Alfonsi; Center of Neurophysiology of Pavia University,Via Palestro N°3,27100 Pavia

In this study, on the ground clinical and EMG data, the Autors 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 PNS lesions.

The patients were examined at the beginning and at the end of the FKTherapic treatment and evaluated both clinically and with the aid of EMG testing. In the clinical examination the parameters of strength, sensitivity and functionally were issued a score. This was given setting score by the motricity 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 EMG 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 thew recovery atteined in the follow-up of various pathologic condition affecting the PNS.

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NORMAL SENSORY NERVE CONDUCTION IN THE FEET NERVES. New method. (Preliminary report)

A. Jušić, M. Vujić and N. Fronjek

Center for neuromuscular diseases, Zagreb, Yugoslavia

Behse and Buchthal 1971 described and standardized the method of conduction velocity measurement along proximal and distal segments of the superficial peroneal, posterior tibial and sural nerves. The stimulated receptor region was the root of the big toe for the first two, and the region near to the metatarsophalangeal joint five, for the sural nerve.

The starting point of our method were following considerations: The amplitude and the surface of nerve potential is first of all dependent on a number of nerve fibers excited. With the stimulation of wider surface more receptors and consecutively more fibers will be excited. We tried therefore to involve with our stimulation as waste receptor region as possible.

SUBJECTS

The probands were healthy young men accepted for regular military service.

The age of examinees was $\bar{x}$: 18,19 years with s: 0,86. 68% of them were high school students, 29% workers, and 2% were dealing with agriculture. The clinical local finding in the feet was following: 90% had cold and humide, 5% had warm and humide, and 5% warm and dry skin in the feet.

METHOD

The cutaneous sensory innervation map, used as point of reference was taken from "Aids to the investigation of peripheral nerve injuries", Medical research Council war memorandum No 7, London Her Majesty's stationary office, 1943: reprinted 1963.

Stimulation points. For superficial peroneal nerve it was dorsum of the foot, just proximaly to the metatarsophalangeal joints, from medial to the lateral border of the foot. For tibial nerve the stimulation was applied over the plantar surface of the foot just proximaly to metatarsophalangeal joints, again transversely from medial to the lateral border of the foot. In sural nerve stimulation the region nearer to the lateral maleolus point was stimulated. In all probands the nerve potentials were elicited with rather low stimulation intensities and amplitudes even as big as 20 µV were obtained.

The stimulation electrodes were adhesive strips electrodes constructed especially by Miroslav Kolaj for this purpose, made of a band one cm wide with metalic core 13 cm long, and 1,5 cm wide.

The intensities of the stimulus used were well tolerated (range 10-30 V); the muscle relaxation necessary for recording small amplitude nerve action potentials complete. The 16-32 stimuli were given during the time of analysis of

50 ms. The square waves were of 0,2 ms duration.

Recording points were following. For superficial peroneal nerve the initial position was that on peroneus profundus, typical for evoking the extensor digitorum brevis direct (M) muscle answer, 4 cm proximally to the middle of the bimaleolar line. The stimulation electrode was removed many times to obtain the potential with lowest stimulation intensities. The distal needle of bipolar electrodes (notisolated) was than placed lataly at 2 cm interelectrode distance as indifferent electrode, and nerve stimulated.

As tibial nerve recording point, the distal stimulation point for evoked flexor hallucis brevis muscle M-potential was used without position change. The notisolated 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 rearangements along the nerves or perpendicular to it, with indiferent 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 Buchtal (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 edge of the examination desk. In sural nerve analysis the patient is lying halfway on the side oposite 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 milimeter distance between the stimulation and recording side was read from the middle of the space inbetween 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 chanels and digital averager DISA.

RESULTS

Table 1

SUPERFICIAL PERONEAL NERVE conduction velocities and nerve potentials features

No		Mean	SD	CV
36	cond. vel. m/s	41,06 m/s	7,18 m/s	17,49%
	ampl. μV	3,28 μV	1,72 μV	52,44%
	width ms	1,50 ms	0,60 ms	40,00%
	S_o T^o	29,55°C	1,28°C	4,33%
	R_1 T^o	30,05°C	1,50°C	4,98%

Table 2

TIBIAL NERVE conduction velocities and nerve potentials features

No		Mean	SD	CV
36	cond. vel. m/s	43,28 m/s	7,26 m/s	16,78%
	ampl. μV	4,55 μV	4,14 μV	90,98%
	width ms	1,53 ms	0,49 ms	32,03%
	S_o T^o	28,33°C	1,23°C	4,36%
	R_1 T^o	29,86°C	0,99°C	3,31%

Table 3

SURAL NERVE conduction velocities and nerve potentials features

No		Mean	SD	CV
30	cond. vel. m/s	40,62 m/s	6,15 m/s	15,15%
	ampl. μV	2,83 μV	1,24 μV	43,83%
	width ms	1,65 ms	0,4 ms	24,24%
	S_o T^o	29,17°C	1,13°C	3,88%
	R_1 T^o	29,37°C	2,06°C	7,02%

Table 4

PERONEAL PROFUNDUS terminal latency quotient

No		Mean	SD	CV
41	Terminal latency quotient cm/ms	1,58	0,32	20,25%

CHANGES IN THE MEDIAN FREQUENCY OF THE MYOELECTRIC SIGNAL IN MUSCLE DYSTROPHY

M.A. Sabbahi, R.G. Rosenthal and C.J. De Luca, NeuroMuscular Research Laboratory,

Children's Hospital Medical Center, Harvard Medical School; and Liberty Mutual Research Center, Boston, Massachusetts, U.S.A.

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 performand 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.

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ELECTROMYONEUROGRAPHY, CT AND IMMUNOBIOCHEMISTRY IN NEUROMYOTONIA CASE ANALYSIS
(Preliminary report)

A. Jušić, Z. Mitrović, Z. Brzović

Center for neuromuscular diseases, Zagreb, Yugoslavia

The variety of names given to the syndrome in question reflects the disagreement about the nature of the abnormal muscle 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 finding. Generalized fasciculations and myokymias over trunk and extremities sparing the head. The permanent muscular contraction with consecutive stiffnes 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 stretsch reflexes, beside 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 digit. communis, deltoid, quadriceps femoris, anterior tibial, gastrocnemius, flexor hallucis muscles) during the rest spontaneous sequencies 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 "bizzare 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 typicaly normal for the muscle analysed.

Ischemic EMG 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 EMG 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 oposition 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; F 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 F 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 ccm of 2% Xylocaine. The canule was used as different stimulating electrode and displaced carefuly in order to obtain the M-potential with as low stimulus intensity as possible. Inspite of extreme M-potential reduction, and deep hypoesthesia 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 ccm of 2% Xylocain. The results were identical both sides. Total repetitive activity joining the M-potential disappeared first. M-potential disappeared with thumb oposition analysis.

Direct muscle infiltration (gastrocnemius muscle) in the myokimia 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 cource, caused by infiltration.

OTHER LABORATORY FINDINGS

Computerized cerebral tomography: difuse atrophic changes. Cerebrospinal fluid: total proteins (Rieder) and gamma globulins percentage increased. IgG increased. Phosphates in urine increased (4 times normal values). Calcium in urine normal. Calcium, phosphates, and alcalinic phosphates in serum normal. Parathormon normal.

DISCUSSION

In our case besides the signs and symptoms overtly due to peripheral nerve distal nerve involvement the cerebral atrophy and IgG increase in cerebrospinal fluid was found. The lively quadriceps reflexes was a finding in disagreement with other authors finding. After the Tegretal supression of continous activity the quadriceps clonus even was elicitable for some days. The cerebral atrophy and cerebrospinal fluid IgG increase was reported in a stiff-man case by Maida et al. 1980.

Schould our case represent a combination of two syndroms stiff-man and Isaacs-Mertens syndrom?

Our electromyographical and electroneurographic findings are very similar to those recorded in our three cases of untreated tetania parathyreopriva! Mertens and Zschocke 1965. already pointed to electromyographical similarity with spasmophilia!

ELECTROMYONEUROGRAPHIC FINDINGS IN NEUROMUSCULAR DISEASE OF HYPOPARATHYREOIDISMUS

A. Jušić, A. Bogunović, M. Žagar

Center for neuromuscular diseases, Zagreb, Yugoslavia

In hypoparathyreoidismus cuff ischemia and hyperventilation provoke the abundant discharging of multiplets. We had the opportunity to observe in four such cases spontaneous generalised appearence of multiplets and serial repetitive frequent discharges with other electrophysiological 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. Dj., 50 years old, male worker.

Anamnesis: Since the age of thirty sight difficulties. With 34 years cataracta 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 finding: The general appearence was of akinesia and bradikinesia with unsteady gait and with some resistence on passive movements pointing to muscle hypertonia. There was no possibility to elicit the proprioceptive 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 paresthesias.

Electromyographical finding (before starting the Calciferol treatment): Quadriceps 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 hith 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.

Neurographic results. Median nerve: motor conduction 50 m/s distal latency quotient 1,2 (5,8 cm/4,8 ms). Sensory conduction (II 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 fibrillary 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 appearence of multiplets and high frequency serial discharges, the quantity of which increase on ischemia and hyperventilation.

In quadriceps muscle electromyographic finding 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 calcifications projecting into the caput nuclei caudati. Abundant calcifications projecting into the posterior thalamus nuclei. Abundant paraventricular calcifications also. Even at both cerebellar hemispheres significant region of increased apsorbtion coefficient.

Cerebrospinal fluid: Cells 28/3, Ly 10, polymorphonuclear leucocytes 3, proteins 0,45 g/l, suggar 4,05 mmol/l. Parathyroid hormon (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, Alcaline phosphatase: 51-119 U/l, Magnesium at serum: 0,3-0,5.

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THE USE OF SURFACE EMG (PART II): THE CLINICAL USE

Hermie J.Hermens, Kasper L.Boon, Gerrit Zilvold
Rehabilitation Centre't Roessingh, Technical University Twente, Enschede, the Netherlands.

INTRODUCTION

In part I it is described how surface EMG can be analysed 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.

METHODS

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 20-50% of the mean maximum force (for instance 20-30-40-50% Fmax). 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 poliomyelitis. The individual action potentials are clearly visible.
In figure 1b a single motor unit pattern is shown of a subject with a plexus brachialis laesion. These kind of registrations cause alternations 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 skewness. Furthermore the number of peaks per second decreases, even to frequencies that coincide with Ffire.

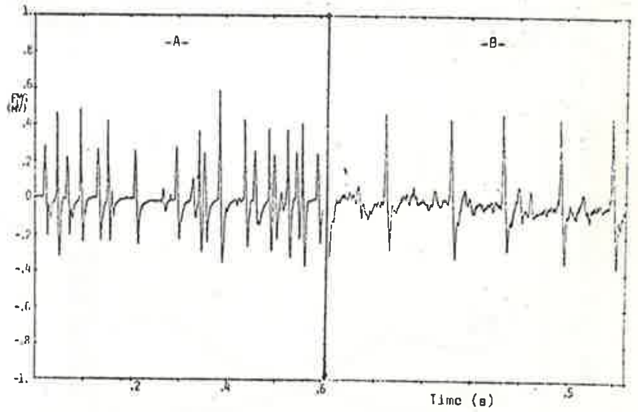

Figure 1. Examples of deviating EMG patterns of the M.Biceps.
A: Poliomyelytis; B: Plexus lesion

Changes in the spectrum

The shape of the spectrum depends on the muscle investigated. With spectra that are related to single motor unit patterns the first peak (Ffire) will be present strongly. The amount of power in this peak depends on the constancy of the discharge rate.
Figure 2 applies to a hemiplegic subject a few weeks after the stroke. The shape of the spectrum deviates significantly ('more flat') with the shape of figure 1 in part I. During rehabilitation it seems that the spectrum tends to be more like the 'normal' spectrum.

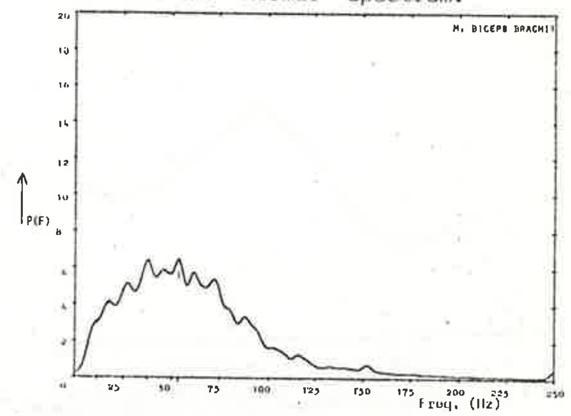

Figure 2. An averaged spectrum of a hemiplegic subject five weeks after the stroke (9 reg.: Fcenter=60, F-10db=120, Prel=15%)

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 force: the value of this parameter usually decreases if an improvement of motoric function is achieved (e.g. from 6% to 2%).
- the parameter EFG: the value of this parameter is often significantly increased. An example of the course of this parameter is given in figure 3.

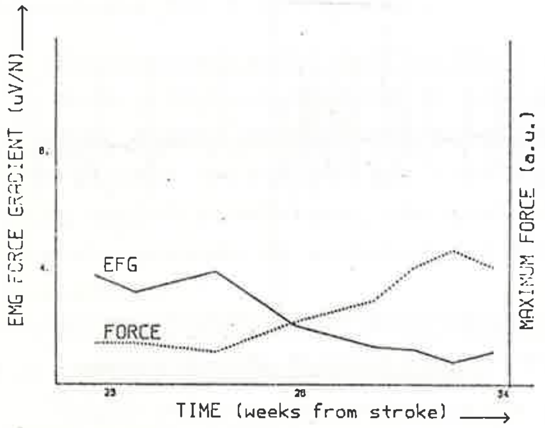

Figure 3. The EMG Force Gradient and the Maximum Force vs. Time with a hemiplegic subject (M.Biceps).

- 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 1! This means that the subject was unable to exert a force without contracting his antagonist to a similar level.

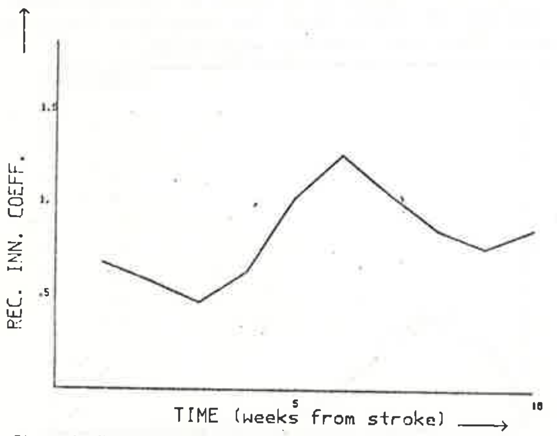

Figure 4. The reciprocal Innervation Coefficient vs. Time with a hemiplegic subject (M.Biceps agonist, M.Triceps antagonist).

- the parameter Ffire: in contrast to healty subjects this parameter is rather irreproducable 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 (Visser and Zilvold, 1978; Prevo, 1979). The EFG is useful to quantify the 'efficiency' of a muscle. Tang (1981) found in half of his group of hemiparetic subjects an increase of the slope between exerted force 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 centres. 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.

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ACETAMINOPHEN OVERDOSE MONITORED BY CEREBRAL EVOKED POTENTIALS

Andre Pugel, M.D., WSU - VA Medical Center, Allen Park, Detroit, Michigan, USA

In the middle 1970's, concern was first expressed in the United States in regard to hepatotoxic reactions from overdose of acetaminophen-containing products. This concern was based on observations made in Great Britain, where consumption of acetaminophen had increased sharply in the middle 1960's because it had fewer adverse effects than aspirin at recommended dosage. This increased use of acetaminophen was accompanied by a corresponding increase in the number of patients using this product for intentional self-poisoning. Reports then appeared in the American literature, suggesting that large numbers of such cases could soon occur in the United States.
Case is presented and monitoring as a guide during treatment and follow-up is demontrated.

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THE QUANTITATIVE ANALYSIS OF NEUROGENIC AFFECTIONS IN ROUTINE CLINICAL ELECTROMYOGRAPHY VIA PHOTO-DETECTOR

Laslo Schwirtlich, Institute of Rehabilitation "Dr Miroslav Zotović", Beograd,

Dejan Popović, Faculty of Electrical Engineering, Beograd.

INTRODUCTION

The quantitative estimation of the maximal volontary 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 activited motor units. The reduction or abscence 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 electormyography. The reduction is usually quantified in the percentage of the full interference pattern, or by description as: full interferential trace, reducted interferential trace, intermediate etc.

The computer analysis has been already introduced in clinical practice. In last 15 years qualitative and quantitative analyses of the electromyography findings were applied mainly in investigations. The digital frequences analysis /1/, pattern recognition /2/, phasic electromyography /3/ or some other 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 volontary contractions, especially in the follow-up of neurogenic affections broght 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 testings. Two reasons were domainant: the possibility of the analysis of the signal and accuracy of the quantitative discrimination.

The eventual application of the photo-detecting quantitative analyser for routine clinical electromyography was the main goal of our work.

METHOD AND EQUIPMENT

From the one point of view the electromyography signal is a scene obtained as the sum of the motor units action potentials.

The simple scene analysis method is convenient for the objective quantitative electromyography analysis /4/. The luminiscence 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 oscyloscope screen. The used instrumentation (Fig. 1.) consists of the concentric needle electrode, preamplifier, 565 dual beam Tektronix oscyloscope, optical adaptation system, selen photo cell, aplifier and LED display. The optical adapter is constructed in such a way that it is compatibile with different oscyloscope screens. The whole screen is in "view" of the photo cell.

The choice of the photo transducer is made after testings comparing the results obtained by computer analysis. The selen 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 the 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 25, 50, 66, 75 and 100 percent of the normal maximal volontary constructions. 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 trimer fittings.

Two effects of nonlinearity must be pointed out. One of these is due to instrumentation and the other one affect of 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 veryfied unilateral lesions of the peripheral nervous system.

The quantitative deficit was tested:

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

by all six patients.

DISCUSION 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 shows that the photo detecting system is simple for application and maintenance, adaptive to the different oscyloscope screens, and objective. Comparing the quantitative results obtained by computer analysis and photo detecting 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 of important for routine clinical work.

The application is possible by unilateral neurogenic diseases (lesions) of central and peripheral nervous system. We also think that this system could by applied with adjustments in some kinesiologyc 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 by photo-detecting device is aproved and described in this paper.

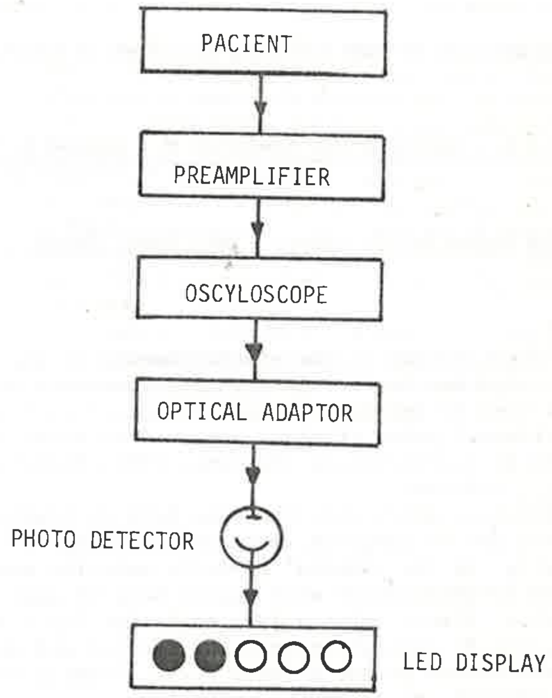

Fig. 1. Model of the photo-detecting quantitative EMG analyses.

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REINNERVATION OF THE TIBIALIS ANTERIOR IN RABBITS IN DIFFERENT REHABILITATING CONDITIONS

Moglia A., Zanlungo M., Lombardi M., Scelsi R.°, Sandrini G., Dalla Toffola E.°°, Alfonsi E. and Arrigo A.

Centro Neurofisiol. clin., °Ist. Anat. Patol., °°Ist. Terapia Fis. e Riabil.- Univers.Pavia, Pavia (Italy)

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., 1979(a)).
The results, which were the same both in treated animals and in controls, were obtained experimentally on the rabbits' tibialis anterior muscle by electrostimulation with square wave 60 msec duration, 5/sec, twice a day, each time for 5 minutes, for 90 days after surgical section and subsequent suture of the sciatic nerve (Pinelli et al., 1979(b); Moglia et al., 1981).
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 kinesitherapeutical 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 perineural neurorraphy were performed according to previously described techniques (Zanlungo 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 isometrical conditions of the muscle.
For the remaining animals very large cages were used, so as 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 (MMCV) 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 MMCV (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 MMCV and M Ampl. obtained in the various groups of animals on the 90th day are shown in Table 1.

Table 1

	MMCV (m/sec)		M Ampl. (mV) p		M Ampl. (mV) d	
	m	SD	m	SD	m	SD
C	56.39	10.06	4.91	0.36	2.74	1.12
E	60.20	8.91	4.00	1.00	3.40	1.21
K	58.39	12.14	5.00	2.00	3.50	1.00

Table 1 supplies the mean values (m) and standard deviations (SD) of neurophysiological results obtained on the 90th day after surgery in control animals (C), treated with muscular electrostimulation (E) and kinesitherapy (K).
The mean value of the fibrillation activity (scale ranging from 0 to 5) that was checked on the 90th day, was 2 in the controls, 1 both in the animals treated with electrostimulation and in those undergoing kinesitherapy. No significant differences were observed among the 3 groups of animals (t 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 regenerated axons with thin myelin sheaths, enclosed within Schwann cell basement membrane.

The muscular specimens were normal on the healthy side, whereas on the side operated on, there was evidence of generalized hypotrophy with atrophy of small groups of adjacent muscle fibres and focal increase of connective endomisial tissues. When observed objectively, no body weight on the limb operated on and presence of eschar (nails, phalanxes and metatarsus) were reported both in controls and in electrostimulated animals, though there was a quite good motility without body weight. In the group treated with kinesitherapy a good recovery of motility was observed in two animals with body weight on the limb operated on and without any eschar; in the 3 animals left, with an incomplete recovery of body weight and with eschar (nails and phalanxes).

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, as 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., 1979(b); Moglia et al., 1981; Arrigo et al., 1981).
Finally, our experimental data seem to confirm what had been reported by Pinelli (1979): vasomotorial factors correlated with the valid muscle contraction seem to recover better in the denervated limb by physical activations which are easier and physiological.

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EFFECT OF HANDIDNESS AND SEX ON CROSS-SECTION AND X-RAY DENSITY OF MUSCLES IN NORMAL HUMAN LIMBS

R. MERLETTI - Politecnico di Torino, Torino, Italy

C. MATHIS, E. RICHETTA, F. REPOSSI - Casa di Cura Cellini, Torino, Italy

Introduction.

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

Materials and methods.

In our study, 7 healthy males (with age ranging from 25y. to 35y. with an average of 28.8 years) and 7 healthy females (with age ranging from 20y. to 42y. with an average of 28.4 years) were selected with no history of neuromuscular diseases or of orthopaedic problems and no activity in asymmetric sports. All subjects had a rigth side dominance.

A Siemens Somatom total body scanner was used to take 7 mm thick scans at 1/3 and at 2/3 of the length of the leg and of the forearm.
Such length was defined as the distance between the head of the fibula and the malleulus lateralis or between the olecranon and the stiloid process of the ulna. X-Ray tube voltage and current were set at 125 KV and 250 mA.
Arm and leg muscles were relaxed.

Area of the muscles and bones as well as area of each muscle lodge (flexors and extensors) were computed with the Siemens E-valuskope System wich also provided the average value and the standard deviation of the density values within the region outlined.
Symmetry indexes for cross-section and density were defined as the ratio of the non dominant to the dominant side values. Fig. 1 shows how the lodges were defined.

Density values were defined according to the Hounsfield scale where 0.0 H.U. is the density of pure water and 1 H.U. is 1/1000 the density difference between water and air.

Results

When muscle lodge density ratioes were plotted against muscle lodge area ratioes for each section, a cluster of points appeared around the 1.1 point. The distribution of such points was not affected by the type of muscle lodge, by the section level or by sex: the points were therefore pooled in a single cluster.
Under the assumption of gaussian distribution of the pooled points, statistical analysis shows that data from 95% of the normal population lie within the ellipses of Fig. 2.
Non-dominant/dominant ratioes of total muscle and bone areas are shown in Table 1. Values from both section levels are pooled together.

Table 1. Average and standard deviation values of the non-dominant/dominant muscle and bone area ratioes (Average ± Stand. Dev.)

	Males	Females	N. of meas.
Upper limb	0.956 ± 0.023	0.975 ± 0.041	14
Lower limb	1.015 ± 0.054	1.012 ± 0.035	14

Density of muscle lodges appears to be affected by section level, lodge 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 lodge is not affected by section level, side dominance, lodge type or sex and has an average value of 11.05 H.U. with a standard deviation of 3.1 H.U.

Discussion.

The ratio of the total muscle and bone area (non-dominant/dominant) is sligthly but significantly lower than one for the upper limb ($p < 0.05$) but not for the lower limb in either males or females (Table 1).

The single lodge density ratoes are lower than one at a marginally significant level ($0.05 < p < 0.1$) while the area ratioes are significantly lower than one for the upper limb but significantly greater than one for the lower limb (Fig. 2).

X-Ray density of the leg muscles is always significantly lower in females than in males and at the distal section level is lower than at the proximal one. Density of plantar flexors of the foot is lower than that of the dorsal flexors. Density values of the arm lodges are mainly affected by sex, the females having lower density than the males ($p < 0.05$).
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 lodge cross-section found in the non-dominant leg is rather perplexing and deserves further investigation and correlation to the individual laterality.
Comparison of our findings with histochemical data is needed to explain the variations observed between the anterior and posterior muscle goups of the leg and between the proximal and distal section levels.

Our data do not match with the observations of Bulcke et al. (1979) who found no significant density differences between section levels and sexes. However, for the age range considered in this study, they also found a significantly higher density in the tibialis anterior with respect to the triceps surae. Our data provide a good reference point for the pathological values already available in the literature (2).

Sabbahi et al. as well as other 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 shapes and to the filtering transfer function of the tissue between the muscle fibers and the electrode.

The latter parameter might well be influenced by the amount of fatty tissue within the muscle or by the thickness of the "panniculus adiposus". Both factors may be easily quantified on C.A.T. scans and some variations of EMG spectral values may be related to them. Further research is under way to estabilish such relationships.

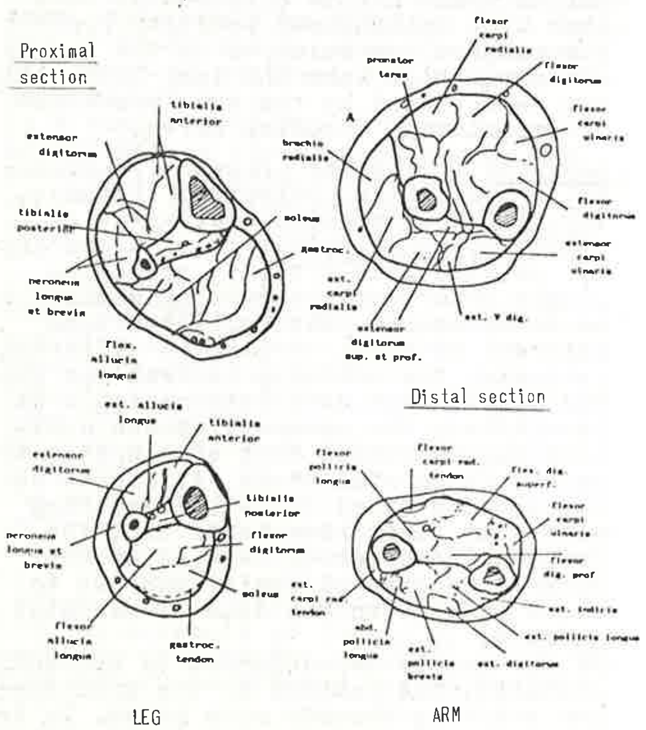

Fig. 1. Definition of the muscle lodges.

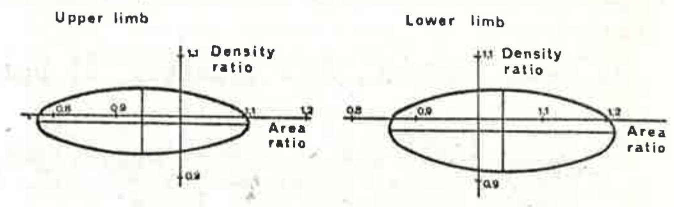

Fig. 2. Cross-section and density ratioes of muscle lodges (non dominant/dominant side) in 14 subjects. The ellipses include 95% of the normal population.

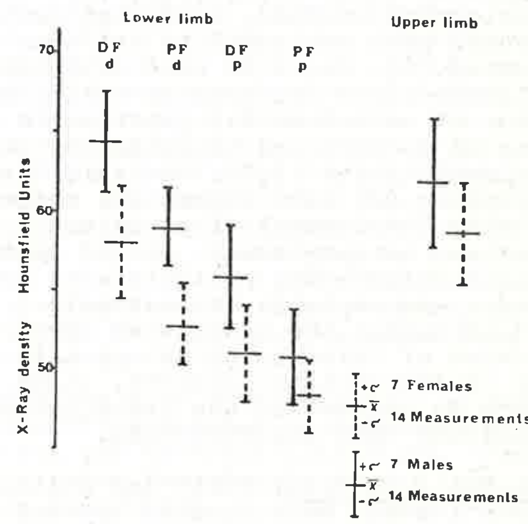

Fig. 3. Density of muscle lodges.
DF: dorsal flexors, P.F.: plantar flexors,
d: distal section, p = proximal section.

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Acknowledgment.

This work was carried out in cooperation with the Rehabilitation Department of the Traumatologic Hospital of Torino. The authors are grateful to Dr. C. Saracco, to Dr. L. Granero and to Dr. M. F. Rolfo for their help and suggestions.

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SPONTANEOUS ACTIVITY IN TETANUS - AN ELECTROPHYSIOLOGICAL STUDY

1) Manik Shahani, 2) F.D.Dastur, 1) D.H.Dastoor

1) E.C.I.Institute of Electrophysiology for Fundamental and Applied Research, Dr.E. Borges Marg, Parel, Bombay 400 012, India 2) Department of Medicine, K.E.M.Hospital, Parel, Bombay 400 012, India

Tetanus toxin is known to have effects at various sites (1,2,3,5,6,7,8). In experimental studies, increased central neuronal pool excitability has been observed (1). There is some evidence to implicate alpha neurones as well (10). Action at neuromuscular junction in terms of failure and blocking has also been demontrated (5,7). Shahani, Dastur & Dastoor (8) have shown that there is definite involvement of peripheral motor and sensory axons. In the context of this information patients with severe tetanus and cephalic tetanus were studied during the stage when there was evidence of trismus and/or spasm in some of the skeletal muscles, in an effort to understand the pathophysiology underlying this manifestation.

MATERIAL & METHODS: Fifty one patients of the tetanus were studied between the third day to about 12 weeks after the onset of trismus. Various skeletal muscles like frontalis, orbicularis oris masseter, biceps, abductor pollicis brevis, abductor digiti minimi, quadriceps tibialis anterior, soleus and extensor digitorum brevis etc. were sampled. Concentric needle electrode was introduced into the muscle and the bioelectrical activity generated was monitored on the oscilloscope and photographed when necessary. The activity was observed for some length of time (few minutes) and changes if any, in the signals or the pattern were noted. In some patients, percussion was applied to the muscle to see if there was any augmentation, while sometimes patients were asked to perform movements in the same direction as the action of the muscle concerned and sometimes in 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. Nerve block with an anaesthetic agent was tried in few patients by an injection of procaine at elbow to block median nerve. The block was verified by proximal electrical stimulation failing to produce "M" 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 masseter muscle (studied in 10 subjects) median nerve was stimulated electrically (1msec at 1 Hz) and the responses were averaged to look 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 mastication when the long loop reflex was demontrated by the same procedure of stimulating a median nerve.

RESULTS: During the presence of chinical spasm 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 oscillations could be identified. The nature of these oscillations resembles that of single motor units. On careful study, it can be observed that these units are occuring almost at fixed time intervals, the frequencies observed have been from 40/sec. in case of facial muscles to about 80/sec. in the case of skeletal muscles.

If the muscle is subjected to mechanical stimulii, the pattern of the spontaneous activity becomes more dense. In demanding voluntary action in the same muscle it is possible to get recruit-

ment of motor units (mixed pattern becoming interference pattern for example) except in those patients of tetanus where there is strong evidence of neuropathy (8). If on the other hand, when the patient is called upon to produce contraction in the opposite direction using an antagonist muscle, the spontaneous activity tends to decrease, but does not get abolished completely; quite often single oscillations continue to appear at fixed time intervals at the same frequency as before.

On electrical stimulation of the nerve supplying the muscle demonstrating spontaneous activity, the activity tended to increase. In some patients, the discrete oscillations appearing at fixed intervals were seen to appear when the stimulation was ceased. On blocking of nerve, the spontaneous activity in terms of discrete oscillations was seen to persist in spite of the failure to get "M" response from abductor pollicis brevis, on supramaximal stimulation of the median nerve proximally to the block.

In the study of masseter muscle giving spontaneous activity, there was no long loop reflex on stimulation of a median nerve in nearly 50 % of patients. However, this could be elicited at a latency of 45-60 msecs. When the subject was asked to contract the masseter muscle voluntarily.

In a number patients with evidence of neuropathy as stated previously, the denervation potentials were also picked up (4,8). In a few patients, pseudomyotonia or high frequency discharge was also picked up from muscles.

DISCUSSIONS: Shahani et al (8) have described three types of spontaneous activity in muscles in severe tetanus. Type I resembling motor unit signals, Type II the denervation potentials and Type III high frequency discharge.

In view of the fact that spontaneous activity (Type I) was not abolished on the antagonistic action by patients, it is possible to assume either an altered mechanism of reciprocal inhibition or that the generator is not central but peripheral. If the activity was originiting as a result of firing of motor neurones in the central nervous system, the time intervals between the succesive signals would not have been more or less fixed. On the other hand, if a potential is generated at an axonal level repeatedly due to some altered state of membrane, it would tend to occur at fixed time intervals. Failure to abolish these signals even after a nerve block strongly suggests that the site of the generator i likely to be in the distal parts of an axon.

The spontaneous activity picked up from the masseter muscle in the state of trismus may be the electrophysiological corelate of the clinical finding. In some patients, there was spontaneous activity but no long loop reflex at the same time on stimulation of median nerve it would suggest that during the study, the trigeminal motor neurones were not in a state of firing (9,11). This corborutes the earlier findings, that signals may be arising peripherally in the axons. It can, therefore, be concluded that trismus is not only due to possible disinhibition of motor neurones, but is also due to regenerating potentials arising in the axons.

The other type of spontaneous activity (Type II) consisted of denervation potentials and positive potentials. This is arising from muscle fibres that have been denervated, due to axonal involvement in tetanus (8).

Rarely spontaneous activity resembled high frequency discharge or pseudomyotonia. This is likely to be arising from muscle fibres, where the membrane stability has been disturbed due to lesion in the parent axon (8).

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TOPICAL ANESTHESIA: KINEMATIC IMPROVEMENTS IN PATIENTS WITH STROKE

M.A. Sabbahi, C.J. De Luca, L.A. Johnson, M.T. Bailin

NeuroMuscular Research Laboratory, Dept. of Orthopaedic Surgery, Harvard Medical School, Children's Hospital Medical Center, Boston, MA; Liberty Mutual Research Center, Hopkinton, MA; and Mass. Rehab. Hospital, Boston, MA, U.S.A.

Introduction

The improvement of motor control in patients with stroke after application of topical anesthesia on the skin has been demonstrated in our laboratory (Sabbahi 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 ± 26) were tested. Patients were afflicted by stroke due to either cerebral vascular accident or head trauma. All had plateaued after participation in physical therapy programs. Four patients were affected with right hemiparesis. Three exhibited mild reduction in skin sensation. Onset of stroke had occurred 7 months to two years prior to this investigation.

Initially, time measurements of 10 rapid repetitive movements (RRM) 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 anesthesia (20% Benzocaine) to the affected upper and lower limbs. For the gait analysis, 3 foot switches (placed under the heel, head of the 5th metatarsal, head of the 1st metatarsal), and 2 electrogoniometers (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 walking speed (WS), temporal asymmetry index (TAI), symmetry factor (SF), and time for single limb support (SLS) (either right or left) during normal walking were calculated as well.

After completion of the preliminary tests, patients were selected randomly to undergo a physical therapy program, and to receive either a topical anesthesia 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 tests were repeated at the end of the two-month period, but without application of topical anesthesia or placebo spray.

Results

Topical anesthesia 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 RRM at the knee joint was measured in 6 patients. Figure 1 shows that in a typical subject, the time for 10 RRM substantially decreased 45 min after application of topical anesthesia to the skin of the lower limb (arrow). Similar reduction was recorded in 5 patients with a range of 6.7%-29% (mean value 17% ± 9%). 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 RRM decreased to 15.4% and 20% 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 RRM at the knee joint. Patients stated that their affected limb was "much looser" after application of topical anesthesia 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 (Dewar 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 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

14.6 ± 3.9) indicating less asymmetry in gait.

The SF (Knuttson, 1972) measures the time ratio between 2 successive steps and assumes a value of 1.0 in normal subjects. Prior to the application of topical anesthesia, the SF for the seven patients tested ranged from 0.59 to 0.93 (average 0.72). Thirty min after anesthesia, the SF increased to a range of 0.63-0.96 (average 0.77), indicating a more symmetrical interaction between the affected and non-affected lower limbs during the gait cycle. Walking speed increased in two patients after the application of topical anesthesia. The SLS of the affected limb increased in 5 patients from 5.8% to 20.5% indicating more functional stability, showed no measurable change in 1 patient, and decreased in one patient (9.2%). The SLS of the non-affected limb decreased in 3 patients. No consistent changes were recorded in the gait parameters mentioned after application of the placebo spray. The short term effects on one patient of both anesthesia and placebo spray on the 5 gait parameters are presented in Figure 2.

Preliminary results of the angular displacement at the ankle and knee joints showed no measurable changes after application of either topical anesthesia or the placebo. However, detailed analysis of a large number of gait cycles remain to be carried out.

The long term effect of topical anesthesia is demonstrated in Figure 1 where the time for 10 RRM progressively decreased throughout the month of treatment with anesthesia. Further investigation is needed to study the long term effect of topical anesthesia in gait parameters.

Discussion

Application of topical anesthesia to the affected upper and lower limbs can improve the active movement patterns in some patients afflicted with stroke. The values of the TAI, SF, WS, and the percentage time for SLS demonstrated a trend to shift toward normal values. Subjective observation of these active movement patterns on videotape showed more striking increases in range, speed, smoothness, and control of movement. Clinical and electrophysiological data presented elsewhere are in agreement with the current results (Sabbahi *et al.*, 1981).

Short and long term effects of treatment varied with each patient, possibly as a result of differences in the pathology. Patients with normal skin sensation showed the most improvement in active movements. Factors such as age and clinical history may have also affected movement performance. Previous electrophysiological studies in our laboratory have demonstrated substantial changes in the interaction of α and γ motoneurons after topical anesthesia was applied to the skin of the lower limb. Correlation between the results of past and present 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.

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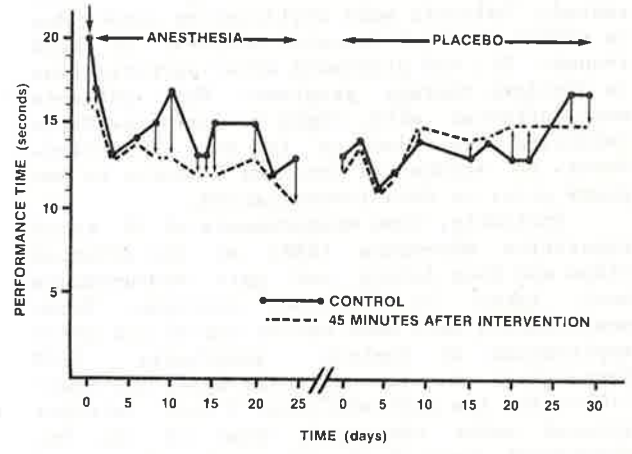

Figure 1

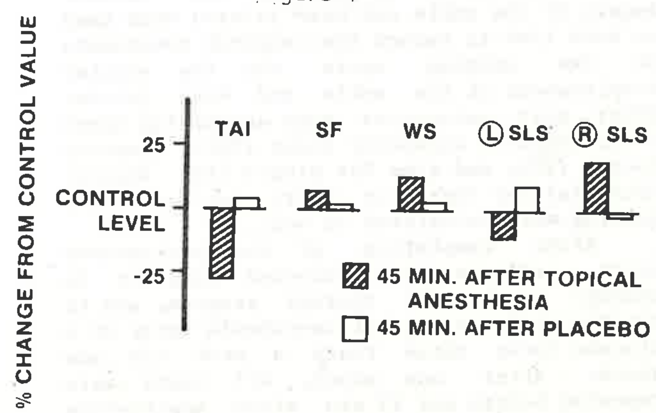

Figure 2

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TREATMENT OF SPASTIC MUSCLES BY MEANS OF PHENOL BLOCKING; EMG ANALYSIS BEFORE AND AFTER TREATMENT

Gerrit Zilvold, Hermie J.Hermens, Kasper L.Boon, Daan Wever
Rehabilitation Centre 't Roessingh, Technical University Twente, Enschede, the Netherlands.

INTRODUCTION

The use of phenol to block peripheral nerves is known since the beginning of this century. Yet it is not applied in most clinics despite the improved tools and advanced knowledge of the effects of phenol (e.g. Halpern, 1977).
In this paper we will discuss the application of surface EMG in order to localise optimal points to inject phenol in a more objective (easily) way. Furthermore the use of surface EMG, in order to quantify the effect of the treatment with phenol is discussed.

METHODS

A. The phenol blocking procedure

The phenol blocking procedure is described in a previous paper (Hermens et al., 1981).
First a bipolar surface EMG electrode is placed on the distal part of the muscle near 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 (EMG) electrode is between the motor endplate zone and the EMG electrode. This however is only possible with muscles with long muscle fibers (e.g. M.Tibialis Ant.). Therefore often the indifferent electrode is placed more proximally and the grounding electrode more distally. The localisation of the motor endplate zone is done on 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 EMG response upon stimulation (10ma, 0,1ms). 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 EMG response and the shape of this EMG response are used to judge the suitability of the point. To obtain the EMG response a scope with a storage unit is used. Some examples of EMG responses are shown in figure 1. It was found that positions that cause clear biphasic responses are the most suited to inject phenol.

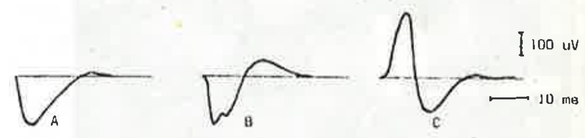

Figure 1. Typical examples of EMG responses. The shape of C coincides with an optimal point.

When such an optimal point is found a little amount of phenol solution (0,2-0,3cc) is injected. The EMG 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. EMG investigation before and after the phenol blocking

In order to evaluate the effect of phenol blocking a separate EMG investigation is made one week before and two weeks after the blocking. For a pilot study we selected eight hemiparetic subjects that were treated at the M.Gastrocnemius. The subject is seated in a device with which it is possible to registrate the isometric force during plantar and dorsal flexion of the ankel joint (figure 2).
Bipolar surface electrodes (Medelec, EL211M) 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 flexors and with the dorsal flexors. During these contractions and during passive dorsal flexion, registrations are made. The exerted force is measured with straingauges and displayed so the patient can maintain the force on a constant level. The following parameters are studied: the standard deviation of the exerted force (SDF), the standard deviation of the EMG signal (SD), the coefficient of reciprocal innervation (RIC) during dorsal flexion and the EMG force gradient (EFG) both at dorsal flexion and at plantar flexion. These parameters are

described in 'the use of surface EMG, part I' (in this proceedings).

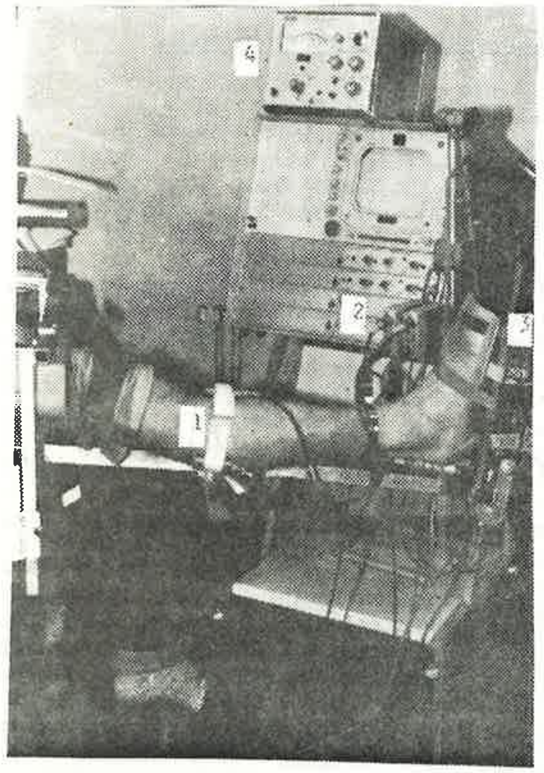

Figure 2. Standard EMG investigation of the M.Gastrocnemius and M.Tibialis: 1: Surface electrodes, 2: EMG amplifiers, 3: Force plate 4: Force transducer and display.

RESULTS

The average amount of phenol used for both parts of the muscles was 4 ± 1cc. Clinically it was observed with all subjects that the tendon reflex lowered (in 5 cases it disappeared) and the ankle clonus (present in 7 cases) disappeared in 6 cases completely. Concerning the changes in the EMG parameters the following remarks can be made:

- the parameter SDF at dorsal flexion did not change.
- the parameter SD of the Gastrocnemius at passive dorsal flexion decreased with all subjects except one. With four subjects this parameter was approximately zero after the treatment. With three subjects this parameter decreased with 50% or more.
- the parameter SD of the Gastrocnemius at maximal plantar flexion decreased with 60% or more with seven subjects. With one subject there was no change of this parameter although maximum force lowered with 50%.
- the EFG of the Tibialis decreased with six subjects; this means that the Tibialis is used in a more efficient way.
- the RIC at dorsal flexion decreased with seven subjects; this means an improved balance between these two muscles.

DISCUSSION

The method of phenol blocking described in this paper is applied during one year at 85 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 neuralgies. Only two cases of neuralgies occurred within this group of 85 patients.

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

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AN ANALYSIS OF THE WRIST EXTENSION IMPAIRMENTS IN HEMIPLEGIC PATIENTS.

Angelo Villani, Paolo Pinelli*, Carlo Pasetti, Giuseppe Di Lorenzo, Dept. of Neurophysiology "Clinica del Lavoro" Foundation Veruno (Italy); * 1st Chair of Neurology, University of Milan.

Specific object of study:
The hand motility has been investigated in 20 acute 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%, 9 patients presented moderate or slight paresis; among these last 9 patients, 5 were improved from previous severe paresis, while 4 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 (pL+) 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 resistence 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 physiokinetic treatment in addition to it, some of them, received electrical stimulation of the E.(20 st./sec).

Results:
I) Different kinds of functional impairment in the E have been ascertained.
1) 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 coinnervation (2b) or flexor reduced elasticity (2c).
2a) A high degree of innervation is reflexly elicited in F (FpL+) by the shortening of E; the FpL+ innervation can act as a mechanical negative force on E contraction and should also exert a secondary effect of antagonistic inhibition. This FpL+ reflex is only elicited in the isotonic contraction (EeL-) with a resulting difference between the innervation of E. in EeL- with respect to the isometric contraction (EeL0)(Fig. 1).
2b) Enhanced coinnervation 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 but with selective innervation of E. were able to develop a higher degree of motion (Tab. I).
2c) Low degree of range motion with respect to the degree of selective innervation of E(EeL-). This EMG/kinetic 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 rather small at the lowest degree of paresis. The distribution of both EeL0 and force in the total population of patients seems to be in agreement

with these findings. (Fig. 2)

B) F.muscles: ratio between increase in voluntary innervation and spasticity.

Voluntary innervation of F.evaluated in isometric (FfL0) contraction showed an increase proportional 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 myoarthropatic disorders, III) antagonistic central coinnervation 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 E.induced a substantial improvement in E. voluntary innervation, in the few patients treated until now.

cases	Selective E innervation a	Selective E innervation b	Coinnervation of E and F c	Coinnervation of E and F d
EeL- μV/sec	162.5	187.5	200	150
FeL+ μV/sec	20	25	45	60
EeL-/FeL+	8.1	7.5	4.4	2.5
FpL+ μV/sec	5	40	15	5
Range (°) of motion	30°	30°	25°	20°

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

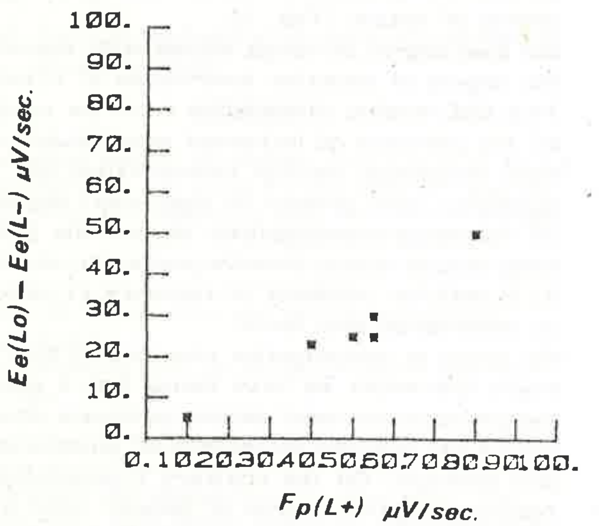

Fig. 1: Difference between isotonic and isometric E. innervation versus F. spasticity in 6 patients.

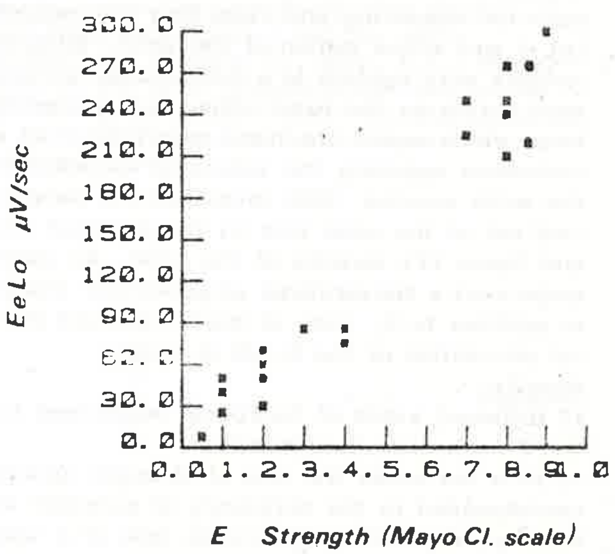

Fig. 2: EMG and strength of wrist extension voluntary contraction in all patients investigated (1st investigation).

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ELECTROMYOGRAPHIC ACTIVITY, MOVEMENT AND FUNCTION OF THE HAND IN SEVERE SPASTIC HEMIPLEGIC PATIENTS.

Thérèse G. Simard, Dépt. d'Anatomie, Université de Montréal and Dept. of Medical Research, Rehabilitation Institute of Montreal, Montreal, Quebec, Canada.

INTRODUCTION

Kinetic alterations and the sensori-motor treatment of spasticity in hemiplegia constitute areas of insufficient knowledge, which require more fundamental and clinical studies. In unilateral brain lesions, integral and educable neurones can regain some coordinate action by different therapeutic approaches. In several stages of the spastic condition, the movements are affected, in part, by an increment of the muscle tone and by a prolonged response to stimuli. The present EMG study was performed to investigate the effect of the degree of upper limbs 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 from cerebrovascular accidents, whose ages were respectively 53, 58, 52, 50 years, were evaluated twice within a three week autonomous program of treatment. Three of these were classified as being severely spastic and one was very severely spastic. Two suffered from an aphasy of expression. Previously, they had all undergone an intensive period of rehabilitation of 12 to 18 month duration. At the time of this study, the post lesional interval was 18, 35, 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 interossei, on line with the head of the first metacarpal bone; 2) of the extensor digitorum, at the center of the superior one third of the back of the forearm; 3) of the flexor digitorum 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 flexed (30°) arm was first immobilized by placing their elbow on a table, the forearm was in pronation and the hand hung from a roll of tissues (Figure 1). This functional resting position was held until the fingers straightened. A second relaxed functional 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 to aid the three first metacarpal segments to slowly relax.

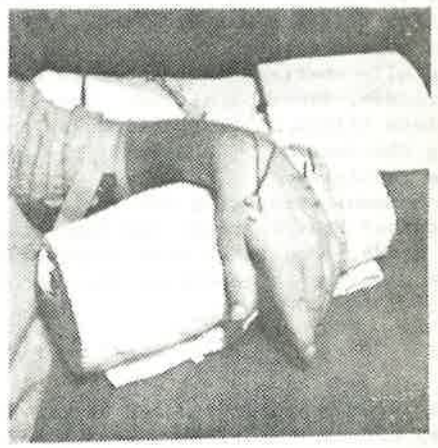

Figure 1 - The first functional resting position in the treatment of a chronic spastic hand.

The forearm and hand were stabilized on the table in midsupination, while the elbow was placed 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 grasp 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 limbs for correct posture, to breathe deeply 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 and a 90° metacarpo-phalangeal flexed position. When no neuromuscular activity was seen on the oscilloscope, the patient was asked to grasp the same cylinder and to release it after a few seconds of effort. During the third trial the EMG was recorded. In the third test, 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 palmar opposition. The wrist was held in 10-15° extension. While the grasp was made, the patients were asked to pay attention to the traces 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 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 ($\bar{X}$s: 3.0-3.4). It is this muscle which contracted tonically to a greater degree than usual in the postoperational period. In contrast the least intense activity was observed in the fourth dorsal interosseus muscle ($\bar{X}$s: 2.9-2.1). While the wrist was at 0° extension and the metacarpo-phalangeal joint flexed at 90°, the flexor digitorum superficialis muscle was as active as the first dorsal interosseus ($\bar{X}$s: 3.1-3.4). The extension of the proximal phalangeal joints in wrist extension slightly decreased the action of the first and fourth dorsal interossei. 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 metacarpo-phalangeal joint either flexed at 90° or extended at 0°, all except one subject performed the grasp simultaneously. In the flexed wrist posture, one patient did not move any fingers during his first evaluation test, but moved all his fingers three weeks later. In this flexed wrist position, the effort given to flex fingers was the longest ($\bar{X}$s: 13 s versus X: 7 s for the two other postures). Three patients shortened their speed at their second evaluation. All four subjects improved the use of their fingers or their range of movement at least in one test at the second evaluation. After the grasp, the opening of the hand was easiest when the wrist was fully flexed. However, the closing of fingers was easiest when the wrist was in extension.

Each patient learned to use their spastic upper extremity, as an assistant hand, in at least one of their daily activities.

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

CONCLUSION

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

Supported by the Medical Research Council of Canada.

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EMG SIGNAL ANALYSIS IN FREQUENCY DOMAIN FOR EVALUATING THE REHABILITATIVE TREATMENT OF HEMIPLEGIC PATIENTS

M.Bracale (1), M.Cesarelli (1).(2), C. Ragosta (1) 1): Cattedra di Eletronica Biomedica, Facoltà di Ingegneria, Università, Via Caludio n.21, 80125 Napoli, Centro di Ingegneria Biomedica e della Riabilitazione (C.I.B.R.), c/o Centro Traumatologico Ortopedico, Viale Colli Aminei n.21, 80131 Napoli
2): Clinica Psichiatrica, Facoltà di Medicina, Università, Piazza Miraglia n.2, 80138 Napoli

This research aims at the evaluation of the motor functional state in patients with damaged central nervous system treated with Functional Electrical Stimulation at the Biomedical Engineering and Rehabilitation Center (C.I.B.R.) in Naples.

The study consists of electromyographic signal analysis both in the domain of frequency and of time by defining parameters fitted to quantify the muscular state.

Patients are made execute a series of isometric contractions with increasing efforts, which allow evaluations of the variations of selected parameters at the different levels of the muscolar effort thus individuating trends. The results are opposite to those observed in a large number of healthy individuals, therefore, a comparison is made during the therapeutic treatment to examinate whether or not the curves tend to move towards normality and, therefore, the level of rehabilitation reached by the patients. The test is carried out in two different phases: firstly, the signals are recorded on a frequency modulation magnetic tape at the hospital; then they are elaborated with a 5451B HP Fourier System at the "Laboratorio di Elaborazione Dati e Segnali Biomedici" of the "Cattedra di Elettronica Biomedica" of Naples University.

The computations are carried for determining the average frequency and the percentages among selected bandwidths (5÷ 25 Hz; 26÷60 Hz; 61÷350 Hz) power spectrum of the surface EMG signal during voluntary contractions.

An other computation is for considering the dissimetry coefficient of the waveshape of the EMG signal power spectrum.

ABNORMAL FORCE-EMG RELATIONS IN PARETIC LIMBS OF HEMIPARETIC HUMAN SUBJECTS.

W.Z. Rymer and A. Tang, Rehabilitation Institute of Chicago and Northwestern University Med. Center

Chicago, IL 60611 USA

Experimental Objective

The weakness of hemiparetic-spastic human subjects is usually attributed to interruption by the pathologic lesion of descending excitatory projections from forebrain and brainstem. However, recent reports[1] describe significant disturbance in motor unit discharge rate in supraspinal lesions. It appears possible that disorganization of motor unit output, arising at the segmental 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 relation between surface emg and limb force in hemiparetic human subjects.

Methods

The relation 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 medial triceps brachii muscles, in precisely measured locations.

EMG was amplified, full wave rectified and band-pass filtered at 10-200 hz, sampled at 500 Hz by A-D converters, and stored in 4 second epochs on computer diskettes. Tension measurements were also recorded and stored.

Subjects were required to generate a particular level of isometric force, by matching 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 9 with stroke, and 5 with carniocerebral 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 ranges 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 muscle. 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., n=17), and from 0.3 to 2.8 in normals (BB 1.19 ± 0.38 S.D., BR 0.98 ± 0.58 S.D., n=11). Subjects displaying abnormal slope ratios were not found to have significant contraction 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 abnormalities of rate and recruitment might also contribute to the sense of weakness and of increased effort that is often reported by paretic subjects.

1) Rosenfalk, A. and Andreassen, S. Impaired regulation of force and firing pattern of single motor units in patients with spasticity. J. Neurol. Neurosurg. Psychiatry, 1980, 43: 907-916/

THE BIOFEEDBACK APPROACH IN THE REHABILITATION OF HEMIPARETIC PATIENTS

P.Albertini (1), M.Bracale (1), R. Castaldo (1), C.Ragosta (1),G.Rutoli (1),

L.Belfiore (2), S. Riccio (2), C.Serra (2)

This work is concerned with the research activity carried out at the Centro di Ingegneria Biomedica e della Riabilitazione (C.I.B.R.) within the limits of the existing convention between the "Cattedra di Elettronica Biomedica" of Naples University and the "Centro Traumatologico Ortopedico" of Naples.

The therapy of biofeedback comes down directly from the concepts of "negative reaction" and "stability" which are tipically of the physical system controls. The application of these concepts, with the relative implications, to a biosystem, becoms particularly interesting if one keeps in mind its possible "learning function" which allows to memorize for longer or shorter periods of time the favourable effects of the "biofeedback" intended as a negative reaction when applied to the physiological system under study.

It has been attempted to bring into the rehabilitative therapeutic approach this fundamental concept.

Infact, in the experience of several years of activity carried out at the C.I.B.R. in the tratment of centrally-damaged (hemiparetic) patients, it has been tried, through treatment methods and relative "ad hoc" instruments, to realize such shemes of "physiological reaction".

The fundamental concept is that of elicitating 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 is reported and the realized devices are described from a technical and functional point of view in a clinical enviroment.

1) Cattedra di Elettronica Biomedica, Facoltà di Ingegneria, Università, Via Claudio n. 21, 80125 Napoli, Italia
Centro di Ingegneria Biomedica e della Riabilitazione (CIBR), c/o Centro Traumatologico Ortopedico, Viale Colli Aminei n.21, 80131 Napoli

2) Centro di Ingegneria Biomedica e della Riabilitazione (CIBR), c/o Centro Traumatologico Ortopedico, Viale Colli Aminei n.21, 80131 Napoli

INFLUENCE OF ELECTRICAL STIMULATION ON ANTAGONIST MUSCLE SPASTICITY

Tadej Bajd, and Bruce Bowman*, Faculty of Electrical Engineering, Edvard Kardelj University, Ljubljana, Yugoslavia, and *Rancho Los Amigos Rehabilitation Engineering Center, Downey, California, USA

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 restrengthen 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.

METHODS

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

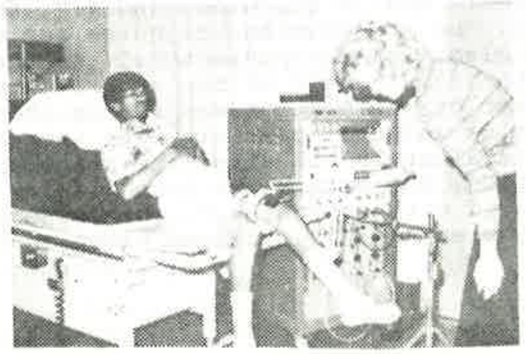


Fig.1

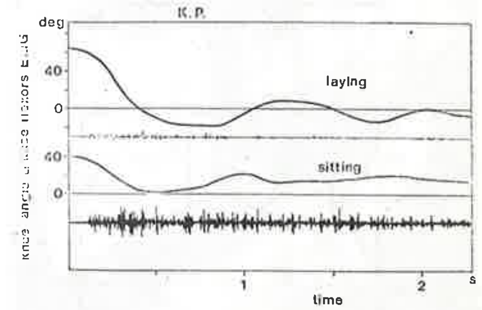
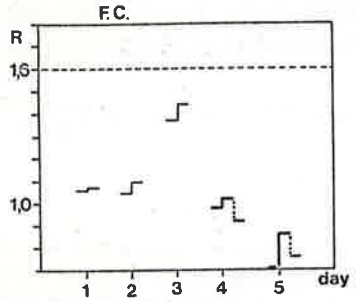


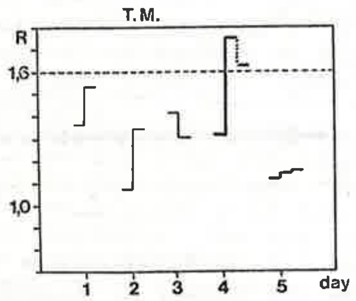
Fig.2

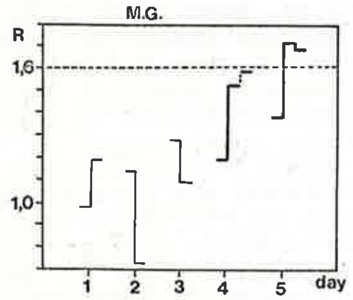
to the resting angle. By measuring the normal population it was found that the value 1.6 of the relaxation index means no spasticity. Zero relaxation index belongs to extreme spasticity. The values between 0 and 1 (the lower leg does not reach the resting position during the first swing) correspond to 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.

The influence of electrical stimulation of the knee extensors on spasticity of the knee flexors was tested on five consecutive days. On the first three days, two measurements were performed with 30 min of rest between measurements. Three measurements were performed on the fourth and fifth days, 30 minutes apart. Cyclic electrical stimulation was applied to the knee extensor muscles for 30 minutes between the first two measurements. The cycle times of stimulation were 6 seconds of stimulation followed by 12 seconds of rest. Stimulation







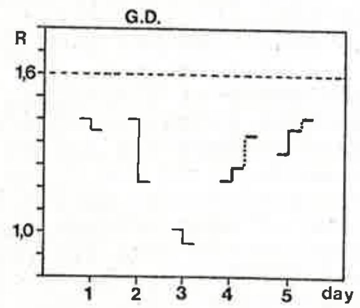


Fig.3

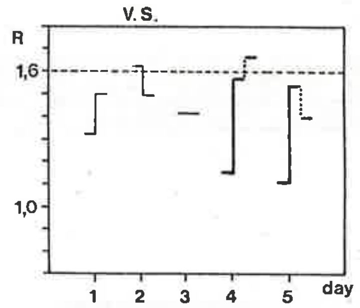


Fig.3

frequency was 33 Hz with a pulse duration of 0.3 ms.

RESULTS

Five quadriplegic patients participated in the program described (Table I). The reasons for their accidents were in most cases motor vehicle accidents and injuries from diving and surfing. They were measured from 3 to 9 months after the accident.

Table I.

No.	Initials	Sex	Age	Lesion
1	F.C.	M	17	C6 inc.
2	T.M.	M	23	C5 inc.
3	M.G.	F	15	C5 comp.
4	V.S.	F	35	C5 comp.
5	G.D.	F	21	C4 comp.

The behavior of the hamstrings spasticity testing was rather irregular during the first three days (Fig.3). With the patient G.D. spasticity was increased in all three days when no stimulation was applied, while with thepatient F.C. it was slightly decreased. In all three other patients the spasticity was whether increased or decreased. Taking into account these measurements it can be stated that electrical stimulation on fourth and fifth day relieved the spasticity in the patients M.G.,V.S., G.D., and T.M.(fourth day). The stimulation was quite successful also with the patient F.C. on the fifth day. In half of the experiments the effect of stimulation was rather short lasting and the spasticity was increased again after 30 min after the stimulation session. The only known irregularity occured with patient V.S. on the second day, when valium had been given to her.

A NONLINEAR ANKLE JOINT MATHEMATICAL MODEL

A.Stefanovska, S.Reberšek, N.Gros*, L.Vodovnik
Faculty of Electrical Engineering, Edvard Kardelj University of Ljubljana, Yugoslavia
*Rehabilitation Institute, Ljubljana, Yugoslavia

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-lenght. Trnkoczy et all (2), Reberšek et all (3) developed a linear sixth and three order models of the ankle-joint system respectivelly. 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 dynamical behaviour.

METHODOLOGY

The electrohydraulic servosystem (Fig. 1) being used in the study of spasticity 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, lightly differed from a sinusoidal waveform. There were two possible reasons for that:

- the subject under consideration was not enough relaxed; this implies the additive neural 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

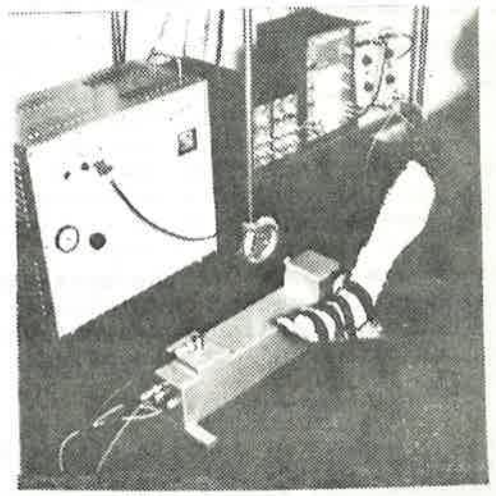


Fig. 1. The experimental set-up

the subject under a deep hypnosis. The EMG measured in such conditions was practically negligable (5 μV) 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. 2e) 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 differencial equation having the coefficients dependent on position and velocity of the ankle.

The order of system was determinated 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" impuls was limited downword by the servosystem dynamics being equal to 240 ms. The amplitude of the delta impuls 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

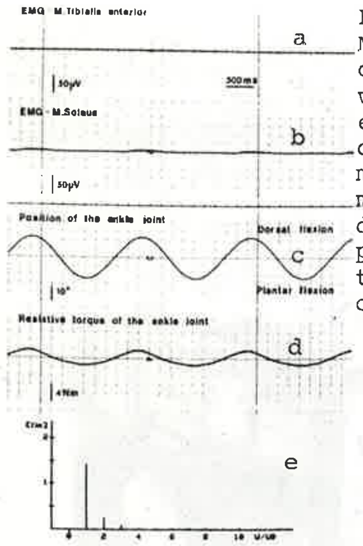


Fig.2 a) EMG of M.Tibialis anterior was measured with Ag-AgCl disc electrodes; b) EMG of M.soleus was measured with stimulation electrodes (changes in position during the measurement cause the artefact)

function was similar to the transfer function of the second order linear system. Thus, our nonlinear system cyn be described by second order nonlinear equation:

$$J\ddot{\varphi} + B\dot{\varphi} + k\varphi = M \tag{1}$$

where J - moment of inertia of the foot; B-coeficient of the viscous damping of the ankle-joint; K-coeficient of elasticity of the ankle joint.

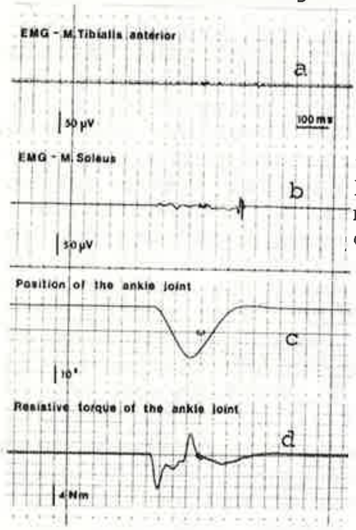


Fig. 3.d) Waveform of the ankle-joint resistive torque (contains an additive component because of the moment of inercia of the foot-plate)

When a sinusoidal displacement of the foot-plate is applied, i.e.

$$\varphi = \varphi_0 \cdot \sin \omega t \tag{2}$$

the second and the third component from the Fourier specter were taken into account while other were neglected. The equality of the Fourier transforms of the left side and a right side of eq. 1. can be approximately guaranted if we take:

$$B(t) = B_0 + B_1 \sin \omega t + B_2 \cos \omega t + B_3 \sin 2\omega t + B_4 \cos 2\omega t$$
$$k(t) = k_0 + k_1 \sin \omega t + k_2 \cos \omega t + k_3 \sin 2\omega t + k_4 \cos 2\omega t \tag{3}$$

Values of parameters in (3) can not be measured. We obtained them by means of the parameter identification based on the reference model. The parameters' values were calculated for two subjects. The results of this identification shows practically equal values for parameters K_0 and B_0 widely used in models of other authors. Beside this the frequency dependence of K_i and B_i parameters mentioned by Agarwal (4) 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 coeficients 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 spasticity measurements and so on.

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MOTONEURON POOL INHIBITION REVEALED IN NON-AVERAGED BUT STATISTICALLY SAMPLED FOURIER TRANSFORMS OF EMG SIGNALS

Robert P. Lehr, Jr.[1], Albert L. Allen[2] and Randall E. Schumacker[2]

[1]Department of Anatomy and Department of Communications Disorders and Sciences, Southern Illinois University, Carbondale, Illinois 62901, U.S.A.
[2]Academic Computing, Southern Illinois University, Carbondale, Illinois 62901, U.S.A.

INTRODUCTION

Sherrington (1910) spoke of reciprocal inhibition of antagonistic muscles as one of the physiological mechanisms which allowed smooth controlled movement. In this sense motor control is as much a function of inhibition of unwanted activity as it is the willing of a desired muscle movement. Indeed this has been demonstrated at the level of the single motor unit (Smith, Basmajian and Vanderstoep 1974). At this level the training of the firing of a single motor unit through biofeedback techniques show that neighboring units progressively stop firing as motor units are isolated. This inhibitory activity on motoneuron pools has been suggested as the mechanism which is responsible for the rotation of separate motor units in a motoneuron pool which prolongs the duration of a skilled task (Lehr and Kasser 1978 and Lehr and Hoshiko 1979). The present investigation was performed to see if significant information could be extracted from the EMG in skilled movements which would reveal inhibition at various levels of maximum contraction.

EXPERIMENT

Thirteen young adults participated in the study. Paired surface electrodes were placed over the first dorsal interosseous muscle of the right hand then over the same muscle of the left hand. The hand was stabilized and the index finger linked by a cable loop to a force gauge whose output was displayed on a two channel oscilloscope. The force of a 10 second maximal voluntary contraction (MVC) was obtained simultaneously with the recording of the EMG. Following a rest period of five minutes, counter balanced acquisition of percentage MVC's of 80%, 60% and 40% were obtained. Each period of data acquisition was followed by a five minute rest period. The trial was repeated on the same hand following a 15 minute rest period. The protocol was repeated on the opposite hand, again care was taken to alternate the selection of the hand used to begin the experiment.

The EMG signal was amplified by a Teca JM Electromyograph (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 by task was submitted to BMD-03T 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 population of frequency spectrums patterns were analysed by Q-factor analysis for similarity of pattern. Q-factor is a statistical procedure documented for broad pattern recognition (Bopp and Biggs 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 parameter. The Q-factor patterns and mean frequency patterns were then compared.

RESULTS

Q-factoring for four factor retention is presented in Table 1. A perusal of the mean frequency patterns shows a similarity of Q-factor to mean frequency as follows: Factor 1 ∿60% MVC; Factor 2 ∿ 40% MVC; Factor 3 ∿ MVC; and Factor 4 ∿ 80% MVC. The 3D graphic of the MVC (Fig.1) shows multiple peaks spread broadly across the frequency spectrum. The number of peaks are reduced in the 80% MVC samples and are more distinct (Fig.2). As the force decreases to 60% 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 Smith, Basmajian and Vanderstoep (1974) who showed a suppression of neighboring motor units. The distribution of the peaks across the frequency spectrum is 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 Kasser 1978 and Lehr and Hoshiko 1979) account for the distinctiveness of the peaks across the various levels of contraction. The relationship of increased distinctiveness with the increased force required 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 activity is reflected in the agonist motoneuron pool rather than exclusively in the antagonist motoneuron pools of muscles providing for postural fixation. Using the methodology of this investigation (Q-factoring, mean-frequency patterns and 3D graphics) on a broad sample of muscles and movements we may approach a more comprehensive understanding of movement.

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FIG 1
MAX

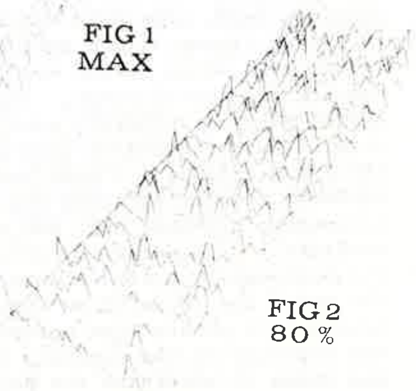

FIG 2
80 %

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EFFECTS OF PASSIVE STRETCH OF THE AGONIST ON THE APPEARANCE OF PREMOTION SILENT PERIOD

K. Yabe, K. Mita, H. Aoki, R. Tsukahara and K. Mimatsu, Institute for Developmental Research,

Aichi Prefectural Colony, Kasugai, Aichi 480-03, Japan

INTRODUCTION

It has been shown that there is an EMG silent period just before a rapid voluntary movement (premotion silent period). The silence seemed to be related to anagility of voluntary contraction, since this phenomenon could no 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 liftoff, so that all extensors are stretched (Walmsley et al, 1978). The gravity might be adequated to accomplish the passive brief stretch of the extensors. The present study was aimed to investigate the effects of passive stretch of the agonist on the appearance of premotion silent period with respect to the combinations of direction of movement and gravity.

METHODS

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 experimenter'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 2-5 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 RECMT residing on a NEAC-3200 digital computer (32 k core memory).

RESULTS AND DISCUSSION

Fig. 2 shows a characteristic pattern of EMG activity with right knee extension. The tonic discharges 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 premotion silent period. Two types of premotion silence were found: one of them was the complete disapperance 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 premotion silence in the vastus medialis was 60 $\pm$ 23 % (mean $\pm$ SD) in the supine, 70 $\pm$ 26 % in the sitting, and 62 $\pm$ 22 % in the prone. No significant difference was found in the rate of occurrence of premotion silence in the three positions. The mean latency of silence for the five subjects was 115 $\pm$ 15 msec in the supine, 120 $\pm$ 17 msec in the sitting and 105 $\pm$ 11 msec in the prone. The mean duration of silence was 42 $\pm$ 17 msec in the supine, 46 $\pm$ 15 msec in the sitting, and 44 $\pm$ 14 msec in the prone. There were no significant differences in latency and duration of silence, and onset of phasic discharge with premotion silence in the three positions.

It was found that the premotion silence occurred not only in the agonist but rarely occurred in the antagonist simultaneously. The most consistent results were obtained in the supine (70 $\pm$ 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 suggested that the apperance of premotion silent period was not due to the passive stretch of the agonist before a rapid voluntary movement.

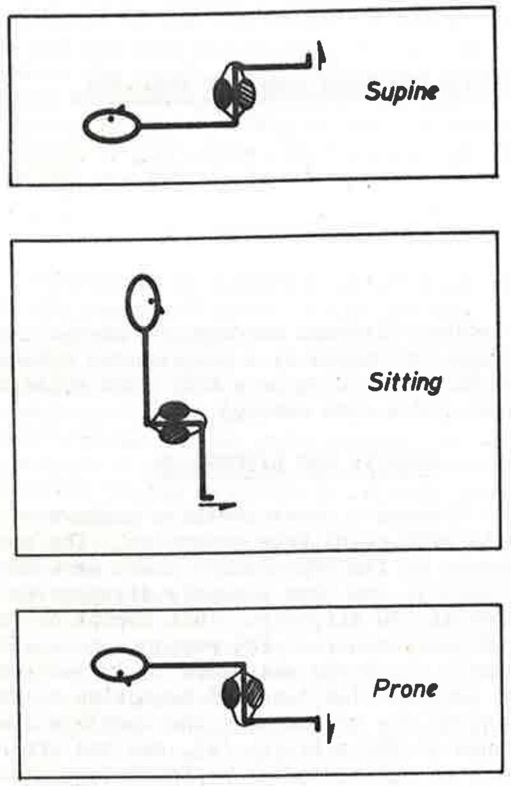

Fig. 1. Schematic representation of experimental arrangement.

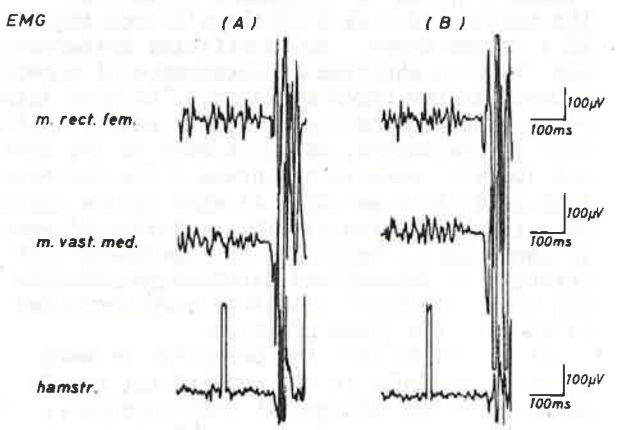

Fig. 2. EMG records of complete (A) and incomplete (B) premotion silent period.

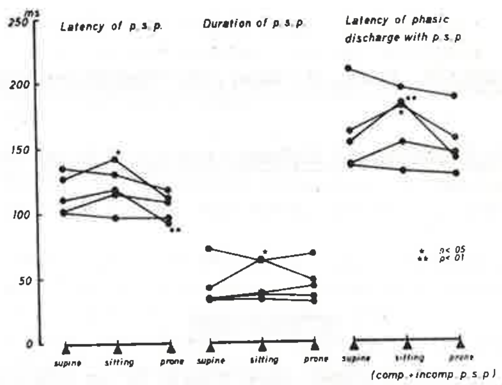

Fig. 3. Latency, duration of premotion silence and onset of phasic discharge with premotion silence in the three positions.

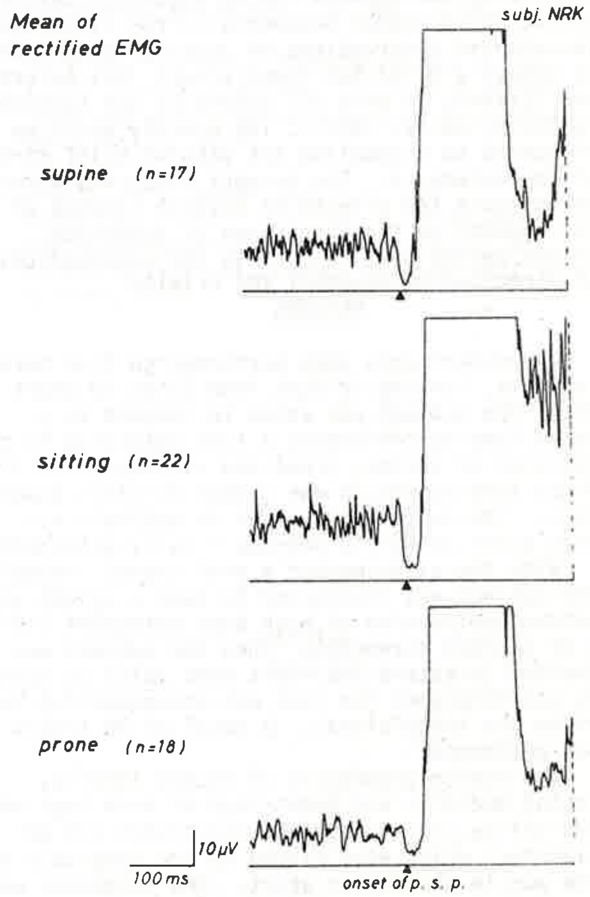

Fig. 4. Mean activity (ensemble average) of rectified EMG responses in the three positions.

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THE STRETCH REFLEX OF PARASPINAL MUSCLES OF NORMAL MAN AND HEMIPLEGICS

Ken Akashi,M.D., Masaki Ise,M.D. and Fukumi Hiragami,R.P.T. Dept.of Rehabilitation Medicine,

Kawasaki Medical School, Kurashiki, Japan

INTRODUCTION

The stretch reflex of paraspinal muscle (SRPM) was reported by Trontelji of Yugoslavia regarding to scoliosis and Carlson of Sweden who examined SRPM of cats, both in 1978. Lundh of Sweden reported SRPM of the patients with back pain, in 1979. Reading these articles, we were interested in SRPM and we thought it could be useful to investigate the effects of the postures to these 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-3, L-1, 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, EKG was taken from V-4 position.

The SRPM were elicited by the electromagnetic hammer. Tapping was given interspace 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 give. (Fig. 1)

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

SRPM was recorded in two ways. One was on the monitoring by EEG machine. One channel of EKG, 8-channels of SRPM and tapping, totally ten channels of records were taken simultaneously. Another was on the computor system. Eight channels of SRPM and tapping were fed to the averager simultaneously. Ten reflexes were summated.(Fig. 2)

SRPM were taken in three essential postures, namely prone, sitting on the stool and standing in ease. Face was hold forward at first then turned to the right and finaly to the left. So, there were nine records in each subjects.

The amplitudes of the waves were measured. A graph was drawn by plotting the amplitude of waves tranversely. So we can see them at once. (Fig. 3)

SRPM were examined in 20 normal person, 20 right hemiplegics and 20 left hemiplegics.

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 generaly higher in left hemiplegics.
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.

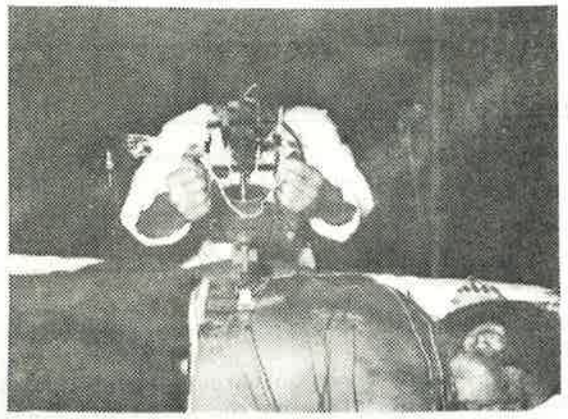

Fig. 1

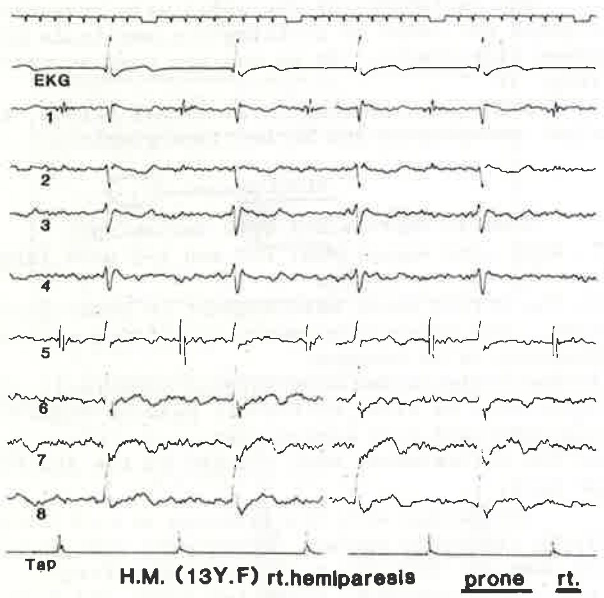

Fig. 2

DISCUSSIONS

There were several interesting findings looking at level, laterality and affected side of hemiplegics. However, the facts that SRPM were higher in normal than in hemiplegics, are very attractive to me. Making everything very simple, there are three factors for motor control. They are, motor cortex impulses, inhibiting factors and excitating factors. Apparently, motor cortex impulses are reduced in hemiplegics. Stretch reflex of affected limbs of hemiplegics is exaggerated and these are explained by disturbed inhibiting mechanisms. In the paraspinal muscle, stretch reflex is reduced in hemiplegics comparing to normal person. There might be some differences in the inhibiting mechanism between muscles of limbs and paraspinal muscles. Or initially excitating mechanism could be less than that of muscles in extremities. Or the motor cortex impulses might be working as a facilitating factors. The SRPM was effected by position of the necks and postures. So SRPM can be influenced more by lower level of central nervous system than motor cortex.

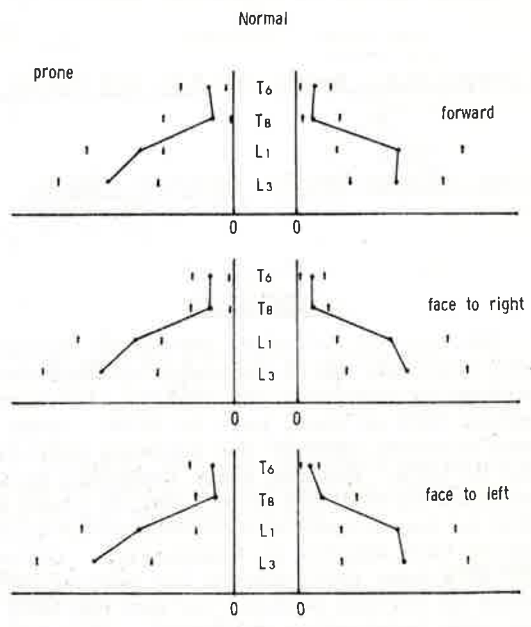

Fig. 3

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BLINK REFLEX INFLUENCED BY TONIC VIBRATION REFLEX (TVR)
---- SOME OBSERVATIONS UPON THE PATIENTS WITH NEUROLOGICAL DEFICIT ----

Masahiko IOKU, Department of Neurosurgery, Kinki University, School of Medicine
Osaka, Japan

Introduction

An unilateral electrical stimulation to the supraorbital nerve elicits the early response R_1 of m. orbicularis oculi on the ipsilateral side to the shock and the later response R_2 on both sides. In case of brain-stem lesion R_1 is usually abolished or reduced. In case of lesion situated in the thalamus R_2 is said to be reduced or abolished.

When vibration (vibratory stimuli) is given to the upper extremity some influences might be observed in the amplitude and/or duration of the late response R_2 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

Electrical stimulation of 0.1 msec. duration square wave is given to the supraorbital nerve, and the blink reflex of R_1 and R_2 to the shock is recorded with two pairs of surface electrodes placed at the infraorbital regions (right and left). Amplitude and duration of the late response R_2 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 R_1 on about 10msec. latency could be recognized ipsilaterally to the shock, and late response R_2 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 ipsilateral to the electrically stimulated side.

The response R_2 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 supraorbital nerve contralateral to the thalamus lesion elicited the R_1 response on the stimulated side but no R_1 on the other side, and elicited R_2 response on the other side (contralateral to the stimulated side, ipsilateral side to the lesion).

Vibration given to the upper extremity contralateral to the thalamus lesion made neither augmentation nor disperive change in amplitude or duration of R_2 of contralateral side to the lesion. Whereas on the ipsilateral side to the lesion some dispersive changes in amplitude and duration of R_2 response could be recognized.

Electrical stimulation to the ipsilateral side to the lesion elicited R_1 and R_2 responses on the ipsilateral side, but no response on the other side (contralateral side to the lesion).

Vibration given to the upper extremity ipsilateral to the lesion made dispersive changes in amplitude and duration of R_2 response on the ipsilateral side but no influence on the other side.

(b) Tumor situated in the thalamus.
5 cased were examined in this study.

Electrical stimulation to the contralateral side to the lesion elicited only R_1 response on the stimulated side but no R_1 response on the other side. Whereas the R_2 response on the ipsilateral side could be recognized.

Vibration to the upper extremity contralateral to the thalmic lesion made neither augmentation nor dispersive change in amplitude or duration of R_2

response on the stimulated side. Whereas the R_2 response on the ipsilateral side showed some dispersive change.

Electrical stimulation to the ipsilateral side to the thalamus lesion elicited the R_1 and R_2 responses on the ipsilateral side. No response could be seen on the other side.

Vibration to the upper extremity ipsilateral to the lesion made some dispersion in the R_2 response on the ipsilateral side, but no change on the other side.

3) Ponsglioma

Electrical stimulation to the ipsilateral side to the lesion did not elicit neither R_1 nor R_2 on the ipsilateral side. R_2 response was made on the contralateral side.

Vibration to the upper extremity ipsilateral side to the lesion gave no influence on the blink reflex (no response).

Electrical stimulation to the contralateral side to the lesion elicited the R_1 and R_2 on the contralateral side to the lesion.

Vibration given to the upper extremity contralateral to the lesion made some dispersive change in both amplitude and duration of R_2 response on the contralateral side to the lesion.

Discussion

In this study, the blink reflex influenced by upper extremity vibration could be observed.

In the normal situation, vibration given to the upper extremity made some dispersive changes in both amplitude and durtion of R_2 response.

Whereas in case of thalamic lesion there was no R_2 response on the contralateral side to the lesion, in which no influence of vibration could be recognized.

The early response R_1 which appeared ipsilaterally on about 10msec. of latency is considered as muscle potentials evoked through a single neuron arc which connected with both the ipsilateral sensory nucleus and the facial nerve. The late response R_2 appears bilaterally after the early response R_1 on about 30msec. of latency and its reflex arc is through the trigeminal spinal tract nucleus and sensory nucleus, the contralateral sensory ascending tract of the trigeminal nerve, the contralateral thalamic nucleus and the bilateral facial nuclei.

Vibration given to the peripheral receptors contralateral side produce impulses upward ascend into the thalmic VL nucleus, then some influence might be given to these reflex arc for the late response, so dispersive changes could be observed in normal case in the late response R_2.

This examination could be used for analysis of disease states in the central nervous system.

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AN IMPROVED NON-INVASIVE METHOD FOR MEASUREMENT OF NEUROMUSCULAR LATENCIES

Harald Reucher, Jiri Silny, Günter Rau, Helmholtz-Institute for Biomedical Engineering,

Goethestr. 27/29, 5100 Aachen, West Germany

INTRODUCTION

Non-invasive techniques are of growing importance for neuromuscular function analysis. However, non-invasive examinations still require great care and take some time; this renders a broader clinical application more difficult. Clinical tests of non-invasive techniques showed the difficulty to obtain reproducible results. Electrophysiological muscle response to nerve stimulation varied strongly dependent on pick-up electrode position. Above all, the problem of electrode positioning for latency measurement caused us to analyse the contribution of propagation of excitation and volume conduction to the shape of recorded signals. Results from experiments and theoretical considerations will be summarized and their consequences for latency measurement will be discussed. An improved device for non-invasive examinations based on these investigations will be presented.

INVESTIGATIONS AND THEORY

The shape of intra-muscular as well as surface-recorded signals is strongly influenced by volume conduction /2,4,5/. When considering the rise of surface potentials the assumption of an unlimited medium must be replaced by the introduction of boundary layers (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 fastest motor-units at t1 causes a steep 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 propagating with muscle conduction velocity (MCV) 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 preceeding 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).

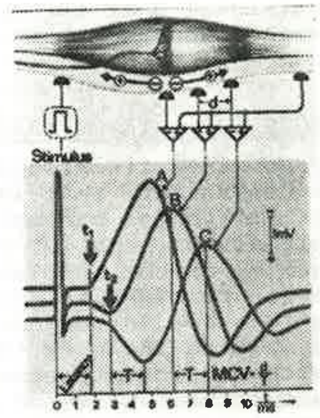

Fig. 1: Shape of electrophysiological muscular stimulus response

A marked increase of IPP-steepness is recorded (signal C at about t2) when depolarization has come nearer to the electrode. Therefore, in the case of long muscles (e.g. flexor carpi ulnaris muscle) the beginning of IPP seems to preceed the negative deflection by a constant time-interval. Similarly structured signals have been observed by other authors /1, 6/. When conduction time of peripheral nerve is determined by measurement of muscular stimulus response latencies (e.g. for measurement of motor nerve conduction velocity NCV) latency of the fastest nerve fibres is mostly derived from the earliest deflection of muscle response /3/. In the case of monopolar measurement in "belly-tendon-position" the different electrode is rarely positioned in direct proximity of the endplate region. Then latency is often measured to the beginning of IPP, assuming the first visible signal deflection to be caused by the fastest motor-units. This is correct in the area near the endplates, since early depolarization causes early volume-conducted currents. In practice, however, the slow initial increase of IPP is often distorted by interference of potentials of adjacent muscles. In the case of a marked IPP 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 the IPP. Before depolarization and preceeding IPP 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 IPP's shape. Therefore, too, latency determination at the begin of negative deflection after preceeding IPP (signal B at t2) only measures the mean conduction time of many fibres.

CONSEQUENCES FOR LATENCY MEASUREMENT

The above considerations supported by experiments demonstrate that latency of the fastest motor units can only be derived from signals without IPP recorded in direct proximity of the endplate region. A device for fast, exact and reproducible non-invasive latency determination should be able to identify the endplate region with sufficient accuracy (maximum deviation about 6 mm for observation at small muscles). Greater deviations cause a decrease in steepness of the negative deflection (imagine transition from signal A to B). The microcomputer-based system for non-invasive measurement of neuromuscular parameters realized in our laboratory (Fig. 2) utilizes multielectrodes consisting of 5 - 9 longitudinally arranged electrodes. Electrode diameters are 4 mm, distance from center to center is 5 mm. When the multielectrode 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 reproducibility are the same as are achievable with careful manual latency determination. Manual correction is possible.
Uncomplicated electrode positioning and time-saving interactive computerized measurement open a broader field of clinical application to non-invasive analysis of neuromuscular function.

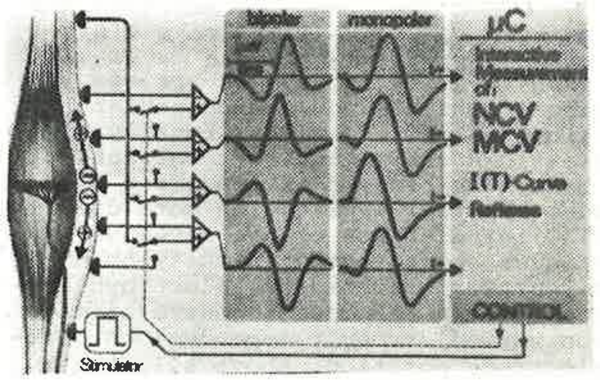

Fig. 2: Improved system for non-invasive measurement of neuromuscular parameters

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A NON INVASIVE MEASUREMENT OF TENDON STRAIN IN VIVO

Ingemar Petersén — Department of Clinical Neurophysiology, University of Göteborg, Sahlgren Hospital, Göteborg, Sweden
Christian Högfors — Department of Theoretical and Applied Mechanics, Chalmers University of Technology, Göteborg, Sweden
Roland Kadefors — Department of Applied Electronics, Chalmers University of Technology and Projekt Lindholmen Center, Göteborg, Sweden

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 improved methods to assess working postures from an ergonomics point of view. Electromyographic signals, particularly through the use of signal spectrum analysis, have proved useful in studies of muscular work (for a review, see (1)). The information obtainable through electromyographic methods concerns muscle load and physiological consequencies of the load as well.

However, clinical experience seems to indicate that complications due to temporal or longterm overload of muscles, in working life or in sports, focus on the tendons rather than on the muscle itself. Electromyographical methods give only indirekt information about the physiological consequencies for the tendons in connection with loads on the musculoskeletal system or in relation to e.g., aging phenomena. There is a lack of methods for studies of the effect of mechanical strain on tendons directly. The present preliminary report concerns the possible use of acoustic waves to detect, in quantitative terms, the level of strain applied to tendinuous tissue.

THEORETICAL BACKGROUND

The most obvious way for studying the physical properties of tendons noninvasively would be by investigating how the state of the tissue affects wave propagation, Electromagnetic as well as acoustic waves would be as relevant a priori candidates for methodological development.

From a mechanical point of view the optimal information to be obtained would be the 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 "window" in the microwave - infrared 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 particular 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 elastic waves and pressure waves. Due to inhomogenities, non-isotropy and possible non-linear properties of the tendon itself as well as boundary conditions, involving effects of shape and influencies from the embedding tissue, these "modes" will interfere and scatter in a complicated manner. However, if it is tentatively assumed that the mixing is a second order effect, at least in the absence of ruptures, an experimental study ought to start with the identification of the modes and the influence of stress on their speeds of propagation. These factors could be measured without a too sophisticated apparatus.

Having identified the modes the continued investigation will start with analysing the response to mono- and bi-frequent wavetrains and various pulse shapes, disclosing any presence of strong nonlinearities. Correlation of the mode signals will give informations regarding the second-order effects. On the basis of such results it should be possible to make an elastomechanical model of the tendon, and relate measurements to mechanical properties.

There are of course several technical problems encountered in pursuing an investigation of this kind. Since the signals are to be injected and recorded noninvasively through surrounding tissue, the most easily detectible mode should be the transverse one. A preliminary study regarding the qualitative effect of total stress on the propagation speed of this mode is reported below.

METHODS AND RESULTS

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 tissues 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 they are then fed into a phase-shift 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 measurement. 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 flextions applied at the force-plate. 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 effects are first-order and easily detectable. In the example given in the figure, the phase shift is of the order of $+1.0^{o}/N$. Similar recordings carried out on the tendon of the biceps brachii muscle have given average phase shifts of $+1.9^{o}/N$, 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 force. The results are unequivocal in this respect. Observations made during these preliminary experiments indicate, however, that there is a considerable non-linear behaviour 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 the longitudinal and the transverse modes. Such methodological development is in progress.

ACKNOWLEDGEMENT

This work was supported by the Swedish Medical Research Council, grant no. B82-04X-05926-02.

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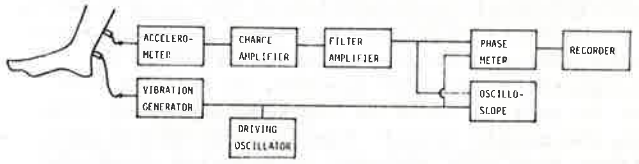

Figure 1.

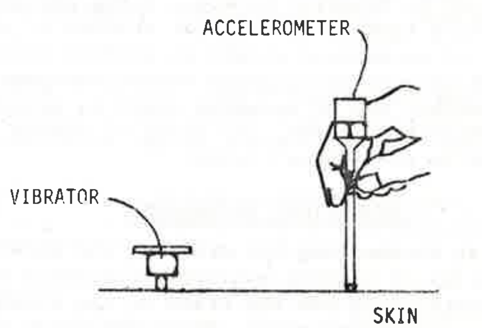

Figure 2.

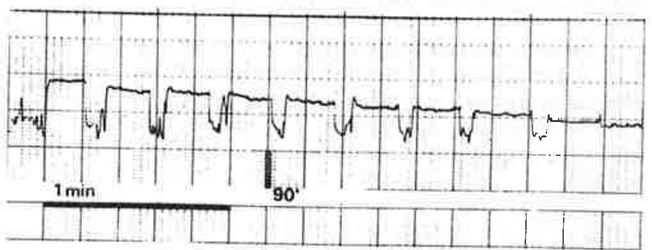

Figure 3.

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COMMON DRIVE CONCEPT OF MOTOR UNIT FIRINGS

Carlo J. De Luca

NeuroMuscular Research Laboratory, Dept. of Orthopaedic Surgery, Harvard Medical School, Children's Hospital Medical Center, Boston, MA and Liberty Mutual Research Center, Hopkinton, MA, U.S.A.

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 (MUAPTs) (LeFever and De Luca, 1982; LeFever et al., 1982). It consists of a multi-channel (via one electrode) ME signal recording procedure, a data compression algorithm, and a hybrid visual-computer decomposition scheme. The algorithms have been implemented on a PDP 11/34 computer. Of the four major segments of the technique, the decomposition scheme is by far the most involved. The decomposition algorithm uses a sophisticated 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 a decomposition. A skilled operator may decompose an ME signal containing 6 MUAPs with an accuracy of 99.8%. Up to 8 MUAPTs 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 (FDI) muscles of the right upper limbs were studied. The force output of each muscle was sensed 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 80% of maximal force (MVC) level at three 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 MUAPTs were examined in the two muscles. The inter-pulse interval between adjacent MUAPs belonging to the same train were measured and 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 force were extracted using computer-implemented cross-correlation routines after the data had been dc 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 1a provides a sample observation obtained from a constant-force contraction of the FDI muscle. Figure 1b 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. An 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 communality 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 2. 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 infraspinal) to the motoneuron pool in the anterior horn has a simultaneous and common effect on all the motoneurons which are excited at any given time. The recruitment properties of a motoneuron are determined by the Size Principle, but once a motoneuron 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 motoneurons increase in like fashion. A correspondingly parallel behavior is noted for increased inhibitory inputs (or decreased excitatory inputs).

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Acknowledgement

Drs. R.S. LeFever and A.P. Xenakis, and Mr. M.P. McCue performed most of the experiments and data analysis. This project was supported in part by NIAMDD grant AM 19665 and the Liberty Mutual Insurance Company.

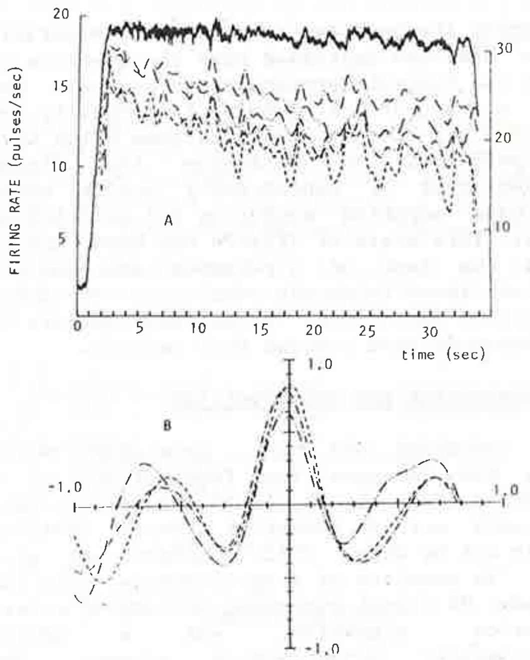

Figure 1

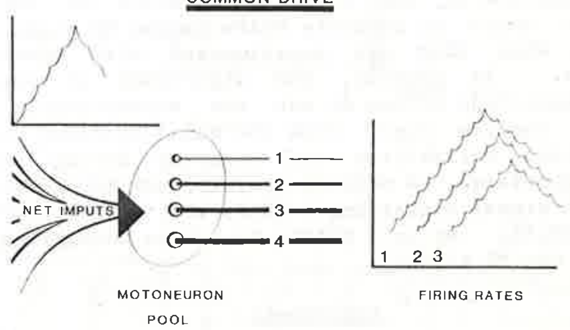

Figure 2

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VISUAL INPUT AND ALPHA (α) NEURONAL POOL IN MAN.

Manik Shahani. B.C.I. Institute of Electrophysiology for Fundamental and Applied Research, Dr. E. Borges Marg, Parel, Bombay 400 012, India.

There is a large body of literature in relation to the excitatory influence of Gr.IA input on the alpha (α) neuronal pool (2,6,7). Eldred and Hagbarth et al (1) have also clearly demonstrated the influence of cutaneous input on the specific neurones. In the case of supra spinal influence, vestibular influences being excitatory selectively for extensors has been studied by various workers (3,15). Shahani (11) demonstrated the excitatory influence of visual input on alpha (α) neuronal pool in man in the patients with corticospinal lesions (11) and later in normal subjects (12).

In order to study the influence of visual input on alpha (α) neuronal pool in man, a simple experiment was set up. The subjects were made to sit comfortably on a chair, with the right arm resting on a plinth. The subjects were made to hold a light weight object (a plastic container of 35 mm film) between a thumb and the index finger. A concentric needle electrode was inserted into the muscle abductor pollicis brevis. As the object held was very light, it was possible to record a single motor unit firing at a particular frequency from the muscle abductor pollicis brevis. The rate of firing was counted per second for the consecutive 10 seconds. The mean rate of firing per second was then calculated. The experiment was repeated in the same fashion, but the subjects were blind folded for two minutes. The mean rate of firing of motor units fell in all the subjects after blindfolding i.e. when the visual input was cut. However, the subjects continued to hold the plastic container. It was also noticed that for each subject there was a certain minimum rate of firing around 4 per second, below which the rate never decreased in any second.

In some other subjects, median nerve was stimulated supramaximally at the wrist, while 'M' response was obtained from muscle abductor pollicis brevis. It was noticed that there was also a late response i.e. 'F' wave. On averaging the 'F' response of 10 sweeps, the latency, the peak to peak amplitude, and the duration of the response were measured. The failure rate of 'F' response was also studied i.e. how many times out of 10 stimulii, 'F' failed to appear. The experiment was repeated in the same subjects when visual input was cut by blindfolding for two minutes. It was found that the amplitude of averaged 'F' response reduced and the duration of the 'F' response increased; there was an evidence of desynchronization of the 'F' wave on elimination of visual input.

In the third experiment, surface electrodes were placed on soleus muscle for recording, while posterior tibial nerve was stimulated (.5 msecs duration at .5 Hz) with a strength of stimulus being much higher than the threshold to give a good 'H' response. Silent period following 'H' response was measured. This study was done while the patient was standing on one foot: in this situation there is adequate ongoing activity in muscle soleus. On repeating the experiment when the subject was blindfolded it was noticed that the silent period was reduced or failed to appear.

DISCUSSION

Firing rate of motor units in man has been studied by many workers (4,5,8). The reduction in the firing rate of the motor neurones on cutting off of the visual input suggests that the visual input has got facilitatory influence on alpha (α) neuronal pool. This might be exerted through descending pathways like cortico-reticulospinal and/or tecto-spinal pathways (11,12). Similar influence, though not so pronounced has been observed on the firing rate of motor units when other inputs like Gr.IA, and/or cutaneous are reduced or eliminated (12). However, at all times for each individual for a particular programme of motor activity (rigid 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 tends to augment the excitatory response of motor neurones so as to provide a greater safety margin for any possible failure or a need for change of programme. This establishes a certain redundancy in the central nervous system which subserves a definite biological function (14).

The changes seen in 'F' 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 (α) neuronal pool. The desynchronization of 'F' wave indicates that some failure of Renshaw inhibition, thus more motor units are invaded later are able to get depolarized (10). This shows that visual input has a general excitatory influence on neurones in the anterior horn, including Renshaw cells.

The reduction in silent period after 'H' response picked up from soleus muscle in standing suggests many possible mechanism. Either the subjects become apprehensive on blindfolding and are now driving the motor neuronal pool more intensely, resulting in the reduction of silent period (9,13). Or more likely in view of the above evidence of failure of Renshaw inhibition observed when studying 'F' waves, it is the same mechanism operating in this experimental situation as well.

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QUANTITATIVE ANALYSIS OF JAW MUSCLE EMG

H.J. de Jongh, R. Dantuma, A. den Arend, Department of Anatomy and Embryology,

and J.R.Luyten, Computer Centre University of Groningen, The Netherlands.

An interactive program package was developed, written in Fortran-4, for use on a DEC PDP11/34 computer with AD11-K a/d converter. The program allows simultaneous digitization and storage of eight EMG-signals and a synchronization 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 EMG 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 particulary suitable for analysis of cyclical phenomena such as chewing, but it may be adapted for analysis of relative muscle forces during other movements also. An advantage is the possibility of simultaneous analysis in multi-channel EMG at fairly high sampling frequency.

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STUDIES ON PHYSIOLOGICAL TREMOR

Halil BENGI, Ph.D.
Electrical Engineering Department, Middle East Technical University, Ankara, TURKEY

ABSTRACT

In this work physiological tremor in human subjects was detected by means of a very small piezo-electric 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 its 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 phsical, physiological and psychological effects. In the so-called "normal" or "physiological" tremor pathological states are excluded.

INSTRUMENTATION

In this work physiological tremor movements in human subjects were detected by means of a very small and light piezo-electric accelerometer. Its out-put was then amplified and filtered by using precision electronic techniques and apparatus which were specially designed for this work in this 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.

METHODOLOGY

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 120 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 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 rotation; mental calculation; "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;

The influence of the retinal illumination on finger tremor is negligibly small.

CONCLUSION

Results of measurements indicate that physiological tremor should not be attrib uted 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.

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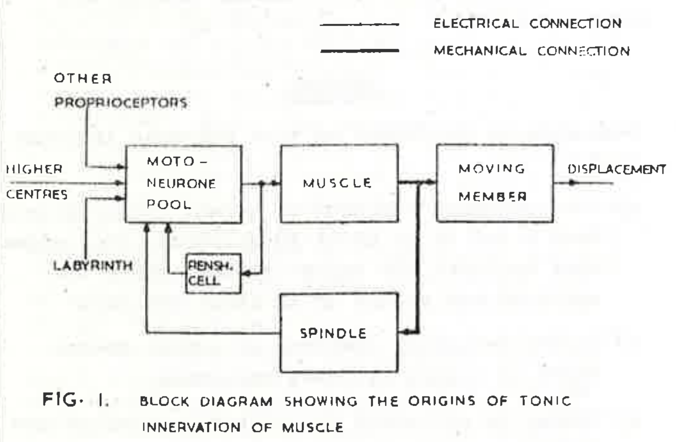

FIG. 1. BLOCK DIAGRAM SHOWING THE ORIGINS OF TONIC INNERVATION OF MUSCLE

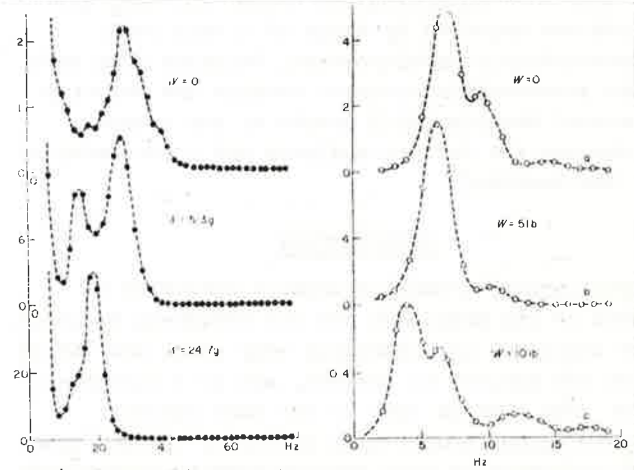

Figure 1 a. VSD graphs of finger tremor when weights have been fixed to a finger.

Figure 1 b. VSD graphs of head tremor with weights fixed on the head.

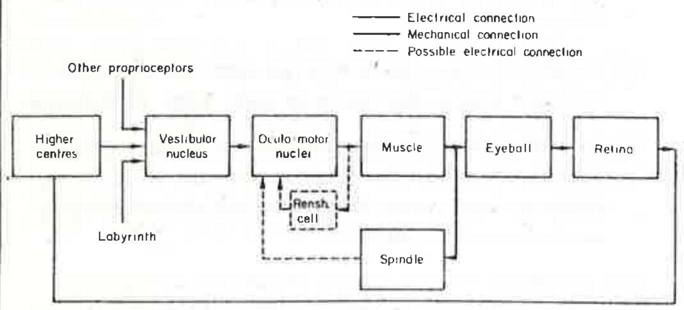

Figure 2. Diagrammatic representation of possible sources of ocular tremor.

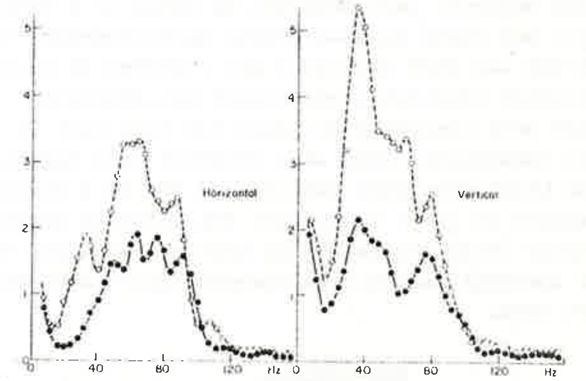

Figure 2 a. VSD graphs of ocular tremor when fixating and not fixating. o-o-o fixating; ●-●-● not fixating

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SARCOMERE LENGTH AND EMG-ACTIVITY OF THE RABBIT MASTICATORY MUSCLES

W.A.WEIJS, Department of Anatomy and Embryology, 69, Oostersingel.9713 EZ Groningen.

The Netherlands.

Sarcomere length of the masseter, temporal, medial and lateral pterygoid and digastric muscles was determined in anaesthetized rabbits, with their mandibles fixed in seven different positions, representative for the normal jaw excursions during feeding. From EMG's 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 sarcomere shortening in both jaw openers and closers. The jaw closers started to fire at sarcomere lengths of 2.6 - 3.1 um and, during closure and occlusion, shorten till 2.1 - 2.3. um. The jaw openers started firing at the same length, but did not shorten (at maximum jaw opening) beyond 2.6 - 2.7 um. 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 lengths. During initial (fast) 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.

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MYOELECTRIC FUNCTION ANALYSIS (MFA): A MULTICHANNEL APPROACH TO QUANTIFY MUSCLE FUNCTION.

R. SOERJANTO, State University Leiden, Academic Hospital, Rehabilitation Dept., Leiden, Netherlands

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 5 (normal, healthy muscle) to 0 (no perceivable contraction in a totally paralysed muscle) [1].
In case of neurological disorders some electrical activity can remain present in the paralysed muscles. It is of utmost importance that these activities can be detected in a very early stage of the healing process as early muscle re-education, e.g. by myofeedback techniques benefit the patient by:

a. reactivation of the muscle control system;
b. reducing muscle atrophy;
c. motivating him for the always laborious training sessions.

As there is apparently a need for assessing paralysed muscles (graded 0 to 1), we developed a multichannel EMG recording and analysing system with surface electrodes. This equipment should meet the following conditions:

1. the recording and analysing procedure 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 paramedical personnel without the assistance of a technician;
3. the results of the measurements should be objective, reproduceable 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 neurologist's conception of the (needle) 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 paralysed 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 it. 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 microvoltseconds.
By recording 12 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 paralysed;
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 myofeedback techniques [2].
By repeating the MFA recordings after each three months, a possible 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 80 to 1200 Hertz. This amplifier is designed and built by the Institute of Medical Physiscs TNO, Utrecht, Netherlands [3]. The electrodes are also manufactured there and consist of flexible PVC carriers 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 (Scotch nr. 2181 or Hellige nr. 217.123.01).
The EMG amplifiers each also contain 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 measured EMG signal in microvoltseconds and is also printed out by a small printer.
The 12 EMG signals can be recorded on paper by a Honeywell 1858 ultraviolet Visicorder. Simultaneously, the signals can be stored on magnetic tape by a 14-channel Honeywell 101 instrumentation taperecorder.
In this setting the processing can be performed on-line and off-line as well.
The EMG signals are monitored on a Knott 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 of 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. Prenger and B. de Wit in: Progress Report 7 (1980), Inst. Med. Phys. TNO, 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 12 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[4] disorders, but also in pure kinesiological studies with specially adapted electronics to eliminate disturbances like motion artifacts and noise. Although only superficial muscles can be recorded, in practice it appears that the method meets most of the requirements.

Summary.

An easily applicable method of multichannel EMG-recording, called the Myoelectric Function Analysis (MFA) is presented, by which electrical activity of muscles can be measured by quantifying the EMG signals in microvoltseconds. The method is especially suitable to be applied in the assessment of paralysed muscles, which have no perceivable contraction, particularly combined with biofeedback muscle re-education.

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MASSETERIC SILENT PERIOD IN PATIENTS SUFFERING FROM CEPHALALGIC SYNDROME WITH OR WITHOUT TMJ PAIN-DYSFUNCTION

G.Sandrini, G. Nappi, V. Collesano°, E. Alfonsi, A. Moglia, A. Arrigo.

Centro Cefalee, Centro di Neurofisiologia Clinica e Clinica Odontoiatrica° dell'Università di Pavia,Pavia (Italy)

Beginning with Costen's article in 1934, functional 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, aching pain centered in the ear or preauricolar area and radiating variably to the temple, the side of the neck, and the angle of the jaw. The pain is usually continuous, and unremitting, unilateral, often exacerbated by mandibular 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 is said to be a common cause of facial pain (Pincus 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 (Schwartz, 1968). Tenderness involving masticatory, facial, posterior cervical, trapezius and sternocleidomastoid muscles can be seen in TMJ pain-dysfunction as in tensional or tension-vascular (mixed) headache (Reick and Hale , 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., 1976).
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 too.

METHODS

The study was performed in 40 patients (26 were females and 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 successively examined in the Dental Clinic 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 (1962). Tensional and mixed headache were considered in the same group. TMJ pain-dysfunction syndrome was diagnosed according to criteria proposed by Reik and Hale (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 solenoid-driven 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 I. This table also indicates the criteria for the classification of the groups, as well as the number of cases for every single group.
The mean age was homogeneous 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 groups A and C ($p < 0.001$), A and Normal subjects ($p < 0.001$), B and C ($p < 0.01$), B and Normal subjects ($p < 0.001$).

TABLE I

	HEADACHE GROUPS			NORMAL SUBJECTS
	A	B	C	
Duration of masseteric silent period (msec) (mean ± SD)	53.6 ±19.1 (16)	50.1 ±17.2 (12)	29.7 ± 8.5 (12)	23.6± 5.3 (10)

STATISTICAL EVALUATION (Student t test)

A - B	NS
A - C	$p < 0.001$
B - C	$p < 0.01$
A - Normal subjects	$p < 0.001$
B - Normal subjects	$p < 0.001$
C - Normal subjects	NS

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.

( ) : number of cases
NS : Not Significant

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., 1976; Collesano et al., 1979). The mechanism of this alteration is not well known. The muscle spasm of the masticatory muscle has been hypothesized as a source of active inhibition of the masseteric motoneuron pool (Bessette et al., 1971). In tensional and mixed headache terderness can be seen also in the muscle of mastication without evidence of a dental malocclusion. However, no pathological increase of the duration of the masseteric silent period was observed in these patients during our investigation. This fact could be explained by a different degree of muscle spasm in group C in comparison with A and B.
The presence of phase plane error in TMJ pain-dysfunction syndrome has also to be stressed according to Bailey et al. (1976). Finally, the nociceptive stimulus is known to produce a longer silent period as compared with the non-nociceptive stimulus (Bratzlavsky, 1973) and it is possible that TMJ pain exacerbated by clench usually occurring in TMJ pain-dysfunction syndrome can justify our data.

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ZDRAV VESTN 51 /1982/ - 5th CONGR I.S.E.K. LJUBLJANA 82

POSSIBILITY OF DETECTING THE RESTING-INNERVATION BY EMG DURING THE FACILITATION ACC. TO VOJTA (ESP. MMC).

Hartmut Bauer and Vaclav Vojta, Dept. of Physiotherapy, Kinderzentrum München, Munich, Germany

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 electromygraphy. 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 associatively) 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 instable. During Vojta's reflex-facilitation the following changes often occured:

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 these 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 neurophysiological 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.

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EMB-CORRESPONDANCE OF NEURONAL ACTIVITY TO THE SEGMENTAL AND COMPLEX FACILITATION ACC. TO VOJTA IN PLEXUS-PARESIS.

Harmut Bauer and Vaclav Vojta, Dept. of Physiotherapy, Kinderzentrum München, Munich, Germany

INTRODUCTION

Primary facilitation in the traumatic paresis of plexus brachii during the neonatal phase is absolutely necessary. Vojta's segmental and complete facilitation (reflex-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 neuronal activation under the facilitation and the difference to possible voluntary maximal innervation by EMG.

METHODS

see: H. Bauer, Possibility of detecting .... (5th Congress of ISEK)

RESULTS

Since 1978 we have recorded EMG in 17 children with perinatal traumatic plexus paresis brachii, mostly in a long-term study (9). Not all of these children were trated 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 phasic and even of the isometric motivity 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 recreation. Under Vojta's segmental and complex pattern of facilitation we have always recorded a good influence to the discharge of frequency of the motor units as well as a recruitment. The effects were even stronger than the effects in the EMG-patterns of MMC.

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 Vojta'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 neurophysiological prooves we are of the opinion that the reflex-locomotion is the therapy of first choice in peripheric neurological damages.

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EMG OUTPUT IN PASSIVE AND VOLUNTARY MOVEMENTS IN CEREBRAL PALSIED CHILDREN; MUSCULAR HYPERTONUS AND RANGE OF MOTION

K. Postema, A.J.H. Prevo, S.L. Visser, T.W. Vogelaar, Department of Rehabilitation Free University, Amsterdam, The Netherlands.

In clinical practice, spasticity denotes a characteristic syndrome of abnormal patterns of muscular activity, leading to hypertonus and deficit of active range of motion. McLellan 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 palsied children are rare.
The aim of our study is to relate the EMG output of passive and voluntary movements with the existance of muscular hypertonus and with the range of motion of voluntary movements.

Up to now 21 cerebral palsied children were involved in the study. 10 girls and 11 boys in the age from 6 to 17 years. All children were premature or suffered asphyxia at birth. 9 of them suffered paraparesis, 10 quadriparesis and 2 children suffered a hemiparesis. 11 of the patients were wheelchair bound; 1 could walk with the aid of crutches; 9 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 seated in a specially constructed chair. The seat, back and arms of this chair were adjustable, so the tall as well as the small children sat in the same position. The thighs and hips were fixed with straps. The lower leg hung free. Muscle activity was recorded with surface electrodes. The electrodes were placed on the middle of the quadriceps and hamstring muscles. An electro-goniometer was tied to the knee to record the angular displacement. In a gentle way 15 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, 15 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 was 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 hamstringmuscles were tested. The scores of hypertonus of both musclegroups were added together.

In the recordings of the passive movements of the normal subjects there was no electrical activity in either the quadriceps or hamstringmuscles. In spite of lengthening the muscles no response activity occured.
In the recordings of the patients four different categories were found.
Category I: In 23% there was no response activity seen in both muscles during the passive movement.
Category II: In two patients there was only activity in the quadricepsmuscles during the passive movements. The mean activity was 16.5 uV. (S.D. 0.5 uV)
Category III: In 23% there was only response activity on lengthening in the hamstringmuscles. The mean activity was 41 uV. (S.D. 15 uV)
Category IV: In this category, 43% of the patients, the response activity in both the quadriceps and hamstring muscles on lengthening during passive extension flexion movements, was the most prominent. The mean activity in the quadricepsmuscles was 31 uV. (S.D. 20 uV) The mean activity in the hamstrings was 45 uV.(S.D. 27 uV)

It is well known that the hamstringmuscles are often more spastic, while the quadriceps muscle if often more paretic.
Yet in the 2 patients of category II, there was only response activity during passive

lengthening in the quadricepsmuscles. The absence of response activity in the hamstrings might possibly be attributed to the translocation of these muscles according to the procedure of Eggers.

These four categories were well related to the existance and severity of the muscular hypertonus. The muscular hypertonus was scored with the help of a four-point scale. The scores for the quadriceps and hamstringmuscles were added together.
Category I: The patients without any response activity during passive movements showed only a weak resistance or a normal tonus. The mean score for the hypertonus was 0.6. (S.D. 0.5)
Category II: The 2 patients with response activity in the quadricepsmuscle, showed both a score for hypertonus of 3.
Category III: In the patients where only response activity in the hamstrings occured, there was a mean score for the hypertonus of 2.8. (S.D. 1.1)
Category IV: In this category where the response activity was the most prominent the mean score for hypertonus was also the highest: 4.8.(S.D. 0.9)
So indeed, a positive correlation seems to exist between the response activity during passive movements and the hypertonus.

With the aid of an electrogoniometer the active range of motion was recorded. The mean displacement of 15 voluntary movements was computed. Here was also a correlation with the four categories. The mean angular displacement in the normal subjects is 72° (S.D. 8°). The angular displacement in patients is expressed in 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 59% of the normal. (S.D. 2%)
Category III: Activity in the hamstringmuscles seems to impair the voluntary movement more than activity in the quadricepsmuscle. The mean angular displacement in category III is 49% of the normal.(S.D. 15%)
Category IV: The deficit of the voluntary movement is the biggest in the children in this category. The mean angular displacement is only 33%.(S.D. 9%)

So the active range of motion in the patients is clearly impaired compared with the normal subjects and a 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 21 patients is very different. For calculations the mean value of the integrated peak amplitude of the 15 movements was used.
In the normal subjects this mean amplitude of the agonistic quadricepsmuscle amounts 390 uV.(S.D. 22 uV) The mean amplitude of the antagonistic activity of the hamstringmuscles is 33 uV. (S.D. 18 uV) Patients show low amplitudes of the quadricepsmuscles compared with the normal subjects. It is only 63 uV.(S.D. 31 uV) The amplitude of the antagonistic hamstringmuscles, in contrast, is higher: 49 uV.(S.D. 41 uV)
It is quite clear that the amount of activity in the agonistic quadricepsmuscle of the patients is very low. These pathological values are due to the existance of paresis in the quadriceps muscle.
The high values of the antagonistic hamstringmuscles are due to the spasticity.
It is evident that the activity in quadriceps and hamstringmuscles during voluntary movements are pathological.
However, the values of the activity of the quadriceps and hamstringmuscles during voluntary movements do not show a correlation with the four categories as the hypertonus 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 conclusions:

A positive correlation exists between the response activity on lengthening in passive movements in the quadriceps and/or hamstring muscles and the hypertonus 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/or hamstringmuscles and the deficit of active range of motion of the knee.

The EMG activity during voluntary movements shows pathological values. There is a decrease of the activity in the agonistic quadricepsmuscles and an increase of the antagonistic hamstringmuscles.
Up to now these values during voluntary movements do not show a clear correlation with the four categories as the hypertonus and active range of motion do.

More research will be done to confirm these conclusions.

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APPLICATION OF FES TO THE UPPER LIMB

J. Allin, J. Toorn and G.F. Inbar, Dept. of Electrical Engineering and the Julius Silver Institute of

Biomedical Engineering, Technion - Israel Institute of Technology, Haifa, Israel.

Introduction

The purpose of the project is to provide hemi- and quadraplegics 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 changing 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 can then 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 (Mann, 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 a 4th order system most of whose elements are highly nonlinear (Inbar & 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. To at least partially negate this dependence, gain parameters for biceps and triceps are measured by the computer, prior to each series of experiments, at three elbow angles. A parabolic fit is made to this data, and used by 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' Minc-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 turntable, to a horizontal plane, with the elbow free to rotate about a vertical axis. The turntable 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 guage 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. Stein and others have modelled changes in muscle damping (as well as stiffness) with increasing active state. Various workers (Polit et al, 1979) have suggested that the

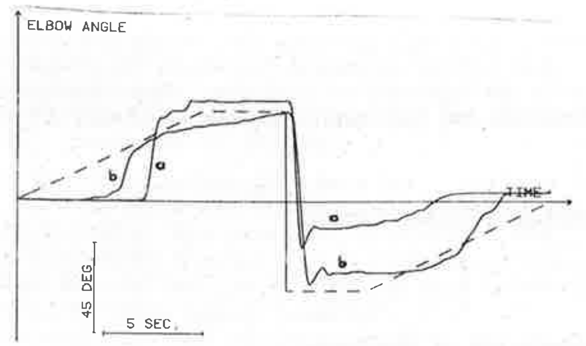

Fig. 1 - Open loop response. (a) No gain compensation (note weak triceps response compared to biceps, and biceps 'dead space'). (b) Triceps and biceps gains calculated separately, and compensation made for their variation with angle.

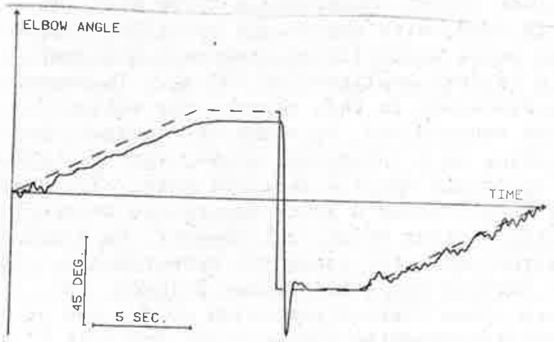

Fig. 3 - Control by cancellation of the plant's dominant complex poles.

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 systems; 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 adjusted to achieve the best response. Fig. 1 is the open loop response, illustrating the effectiveness of measuring biceps and triceps gains separately. It also shows that reasonable results can be achieved 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 1.2, 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 for a controller can be seen, though at this stage it is not appropriate to say which of the two examined is superior.

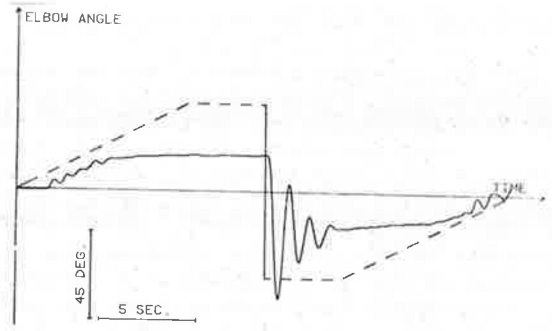

Fig. 2 - Closed loop response with a proportional controller.

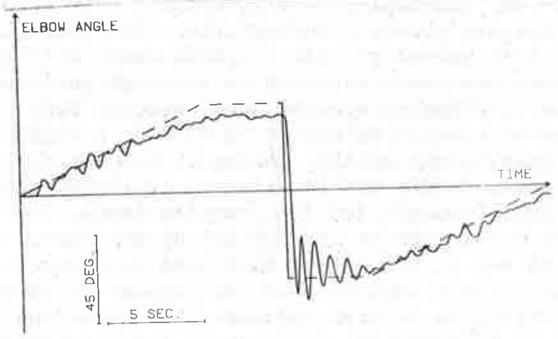

Fig. 4 - Control by variable damping, using coactivation.

In addition to work detailed above, control of wrist rotation is being examined, together with techniques of parameter identification.

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THE EFFECTS OF FUNCTIONAL ELECTRICAL STIMULATION (FES) TRAINING ON THE ACTIVATION PATTERNS OF ANTAGONIST MUSCLES

Pierre Lagasse, Claude Rouillard, Martin Normand and Louis Tremblay, Laboratoire des Sciences de l'Activite Physique, Universite Laval, Ste-Foy, Quebec, Canada G1K 7P4

INTRODUCTION

The use of functional electrical stimulation (FES) to restore motor patterns in patients suffering from 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 retraining or 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 point of the pectoralis major, posterior deltoid, biceps brachii and triceps brachii were utilized to monitor the electromyographical activity of these muscles. Potentiometers 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 (Hewlett Packard, 8811A) and recorded on photosensitive paper by a multichannel fiber optic cathode ray tube recorder (Honeywell, visicorder 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 two 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 brachii and triceps brachii 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 emg 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. 58019) square wave stimulators delivering pulses of 1 ms duration, 15 volts above rheobase and at a frequency of 100 Hz and a Stoelting (model no. SA600) four** channel universal timing module were used to stimulate the muscles.

DATA ANALYSIS

For each trial, several measurements were derived from the records. Total movement time (TMT) was the time interval between the onset of movement and 75° of forearm flexion, proximal movement time (PMT) was the time elapsed between onset of movement 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 during which the posterior deltoid (PDAT) and the triceps brachii (TBAT) were active during the execution of the movement. The time elapsed between the onset of antagonist muscle activity and the instant when either the arm or the forearm reached 45° of adduction or 75° of flexion were the posterior deltoid-target latency (PDTL) and the triceps brachii-target latency (TBTL).

RESULTS

The results pertaining to the effects of 800 practice trials of the movement on the activation patterns of antagonist muscles have already been presented by Normand et al. (3), who reported significant reductions in TMT, PMT and DMT, as well as in PDTL and TBTL attributable to the training regimen. The activity time of the two antagonist muscles, PDAT and TBAT, did not change between testing sessions one and two.

The analysis of variance model for repeated measures on the same subject (5) was used to assess the effects of the FES training regimen on arm adduction and forearm flexion speed and on the activation patterns of the antagonist muscles. The results of these analyses demonstrated that TMT and PMT decreased significantly ($p < .01$) after FES training, whereas DMT remained unchanged. The statistical analysis also demonstrated that the activity time of the two antagonist muscles, PDAT and TBAT, remained unchanged after FES training, whereas antagonist muscle target latencies, PDTL and TBTL, decreased significantly ($p < .05$) between testing sessions two and three.

DISCUSSION

The results of the present investigation clearly demonstrate, as is the case for a traditional training regimen involving actual physical practice (3), that a FES training program based on the temporal sequence of activation of the contralateral muscles facilitates the learning process of a novel task, as shown by the reduced arm adduction-forearm flexion movement times. Furthermore, and as observed for the regular training program applied to the right upper limb, modifications occurring in left arm adduction forearm flexion speed of movement were accompanied by significant changes in the activation patterns of the antagonist muscles, as shown by reduced posterior deltoid and triceps brachii-target latencies following FES training. Thus, FES with no actual physical practice produced changes in the central nervous system that are similar to those observed following traditional training. This motor skill acquisition was achieved strictly through a sensory pathway.

These results strongly support the model of Vodovnik (4) concerning efferent and afferent FES. In this model, Vodovnik (4) proposes that FES produces two effects: a direct and immediate functional motor response resulting from stimulation of the efferent nerves and an indirect and lasting functional motor response which changes the excitability of the spinal motorneuron pool resulting from afferent fiber stimulation. Even though Vodovnik's model concerns patients suffering from hemiparesis resulting from brain damage, it may very well be, in the case of the present investigation, that the hypothetical changes in the excitability of the motorneuron pool attributable to afferent fiber stimulation were directly responsible for the acquisition of the optimal motor program required to execute the fast speed arm adduction-forearm flexion movement.

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Supported by grant no. A-9432 from the Natural Sciences and Engineering Research Council of Canada.

EMG AND KINESIOLOGIC CONTROL STUDY ON ELECTRICAL THERAPY OF THE HEMIPLEGIC HYPOTONIC SHOULDER.

Vittorio Alfieri and Alberto Vitale, Dept. of Physical Therapy and Rehabilitation, Nuovo Ospedale di Monza, MONZA, Italy.

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 21% 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 of pectoralis major; lately we are using this therapy also in quite hypotonic hemiplegic shoulders, extending the stimulation to supra- and infraspinatus muscles. In order to investigate on the real value of electrotherapy of hypotonic hemiplegic shoulder, a comparative experiment has been made recently in our department.

Material and Method.- Twelve hemiplegic patients with drop painful shoulder (8 males and 4 females; age, 48 to 75; time from stroke, 1 to 8 months) have been randomized in two groups of 6; the 1st group received electrostimulation of deltoid, supra- and infraspinatus muscles and passive movements; the 2nd group received short waves, ultrasounds, massage and kinesitherapy. Parameters of electrostimulation are: for the 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, 100-300 msec width and 0.5 Hz. EMG of middle deltoid and pectoralis major and sometimes of supraspinatus, X-rays and photographs of the shoulder and kinesiologic assessment have been made before and after treatment. We made use of raw EMG and IEMG of deltoid muscle; moreover we detected the raw EMG 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 90°/sec, and held in position for some seconds; sometimes we detected EMG from both deltoids for a comparison between affected and sound sides. The number of treatment sessions has not been less than 15 in both groups; in particular, the 1st group received from 15 to 50 electrostimulations and kinesitherapy (average, 25.5); the 2nd group had from 10 to 15 sessions of ultrasounds, short waves and massage and from 20 to 45 sessions of kinesitherapy (average, 29.6).

Results and comments.- Diagnostic remarks. Seven of the twelve patients (about 60%) had more or less remarkable hypotrophy 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 fibrillation in an early stage and oscillations of high amplitude (about 3mV) in the stage of reinnervation; these cases of peripheral denervation were confirmed by strength/duration curves, demonstrating mainly an increase of rheobase and a strong reduction of accomodation ratio. This could be ascribed to neuroapraxia and probably to an ascending suffering 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 5 of the 2nd, but increasing of trophism was more evident in the 1st group. Voluntary activation of deltoid muscle was at least doubled up in every patient of the 1st group, while in the second 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.

Pain: in the first group 1 patient had complete relief of pain, 1 no relief and the others partial improvement; in the 2nd group decreasing of pain was less sensible. Stiffness: about the same results registered for pain.

In conclusion, though in some patient treated with electrostimulation relief of pain is dramatic, the results concerning pain and stiffness are not so good as expected.
The best results of electrostimulation, compared with other treatments, are obtained in reducing spasticity of pectoralis major and in increasing trophism and voluntary activity of the muscular suspension system of the upper limb.

Fig. 1 - Mr. A.P.; right hemiplegia; stroke, 8th Nov., 1981; registration 10th Dec 1981; middle deltoid at rest: fibrillation.

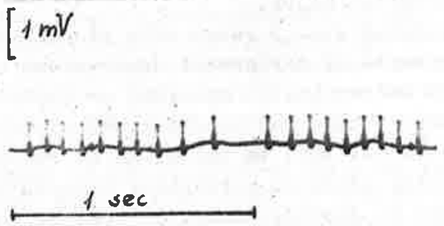

Fig. 2 - Same registration of fig.1; maximal voluntary activation.

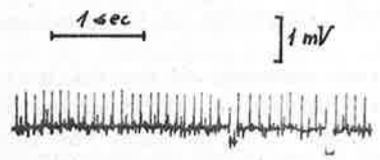

Fig. 3 - After 21 sessions of electrotherapy.

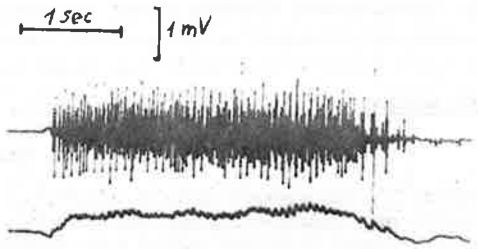

Fig. 4 - Same patient; voluntary activation of deltoid muscle after 50 sessions of electrostimulation.

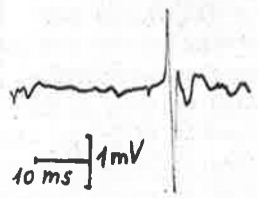

Fig. 5 - Registration as above; at minimal effort, single MUAP of high amplitude, as expression of reinnervation by collateral branching.

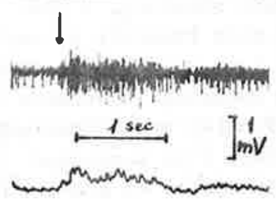

Fig. 6 - Same patient; registration at the same date of fig. 3; reflex activity of pectoralis major; arrow: stretching; lower trace: IEMG (mean voltage 100 msec).

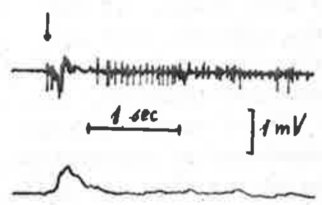

Fig. 7 - Reflex activity of pectoralis major after 40 sessions of electrotherapy.

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8-CHANNEL ELECTROSTIMULATION SYSTEM FOR CORRECTION OF WALKING IN PATIENTS WITH DISORDERS OF SUPPORTIVE-LOCOMOTOR SYSTEM

N.I.KONDRASHIN, A.S.VITTENZON, V.G.ZAREZANKOV, B.A.ARANSON

Multichannel functional electrical stimulation of skeletal muscles in the process of walking in an effection 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 biomechanic structure of gait and to form a more correct gait

In the CRIP an 8-channel system has been designed, which contains an electrical stimulator, flexible multilay electrodes, synchronization 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 8 pairs of electrodes with time division of channel impulses and switch division of electrodes. Synchronization 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 comisting 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, visial or digital indication is stipulated, as well as outputs for recorders. For convenience of use we have provided the generations of inner rythm of the stimulation program, a large range of smoothly regulated parameters of stimulting signals.
System tests on patients have shown its high effectiveness, resistance to disturbances, convenience for use with correctiona aims in patients of various etyology.

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MEASUREMENTS OF GAIT CORRECTION WITH HEMIPLEGIC PATIENTS BY MEANS OF ELECTRICAL STIMULATORS

Aćimović-Janežič R.[+], N. Gros[+], J. Krajnik, M. Malezič, M. Stopar, U. Stanič, M. Kljajič

REHABILITATION INSTITUTE LJUBLJANA[+], INSTITUTE JOSEF STEFAN UNIVERSITY OF LJUBLJANA

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 (Kljajič 1979, Krajnik 1981) and electrogoniometers (Bajd 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 circumference 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 implantible 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.F. 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 (Aćimović 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. gluteus maximus, m. quadriceps, knee flexors, m. quadriceps surae, n. peroneus and m. triceps bracchii.

In the second phase we stimulated m. quadriceps and n. peroneus. In the third phase stimulation was performed via n. 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 disclocated. 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 1). There sets in a more adequate symmetry with respect to the healthy extremity (Fig 1). Electroganiometers 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 anomalies, namely, from 18 to 21,5. The stride time decreases from 2,23 to 1,79 a certain difference 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. Evident 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 n. 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 parametrs 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 continously applied the one-channel surface ES (Fig 4). Therefore, we decided to apply an implantible system making possible ES via n. peroneus. The measurements after the implantation showed an increased velocity of gait with decreased stride time from 2,22 to 1,80 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 weaks), 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 the other groups of patients. They mostly corroborate the clinical observations and complete them from the quantitative point of view. By this method of measuring we successfuly evidenced quantitative biomechanical data, particularly with respect to the changes in the transfer of weight, loading of the affected lower extremity, thereat proving also changes in the velocity and the symmetry of gait. The computer out-prints made possible a not time-consuming and up-dated 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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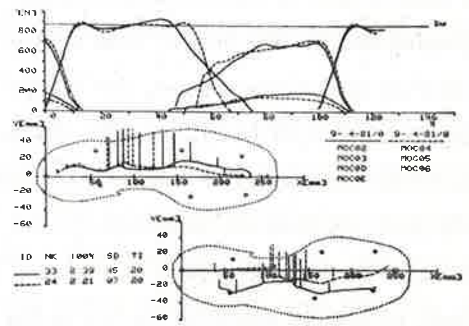

Fig. 1. The difference between No ES(——) and six-channel ES (----) (M.F.,-H.D.)

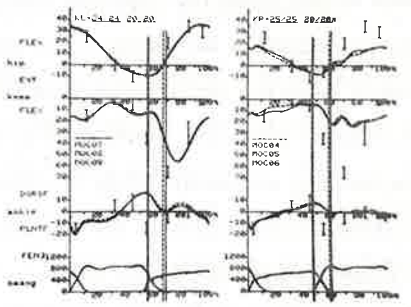

Fig. 2. Goniograms with two channel (——) and 6-channel ES (----) (M.F.,-H.D.)

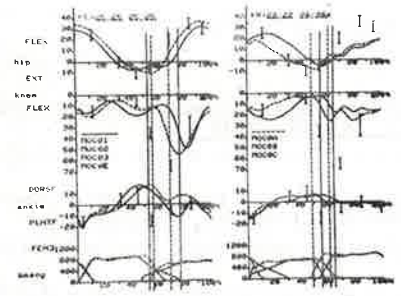

Fig. 3. Goniograms with No (——) and one-channel ES (----) (M.F.,- H.D.)

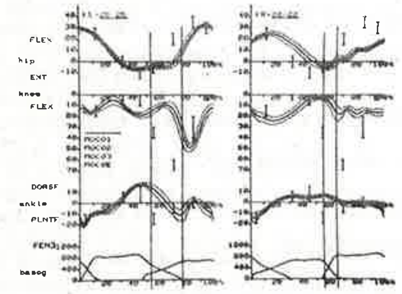

Fig. 4. Goniograms with No ES (——) and standard deviation (----) (M.F.-H.D.)

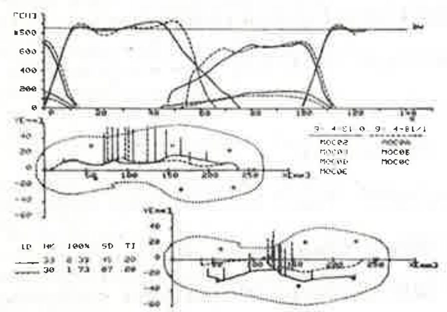

Fig. 5. The difference between No ES (——) and impl. one-channel system (----) (M.F.-H.D.)

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FEASIBILITY OF FUNCTIONAL ELECTRICAL STIMULATION IN LOWER MOTONEURON LESION

Martin Štefančič,[1] Lojze Vodovnik,[2] Vojko Valenčič,[2] Tina Jelnikar,[1] Vanja Simič[1]

[1]REHABILITATION INSTITUTE and [2]FACULTY OF ELECTRICAL ENGINEERING Ljubljana, Yugoslavia

According to a belief as prevailing in the world, the use of functional electrical stimulation (FES) is indicated primarily in case upper motoneuron 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 peroneal muscle is mostly used to prevent spastic equinovarus in 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 sucessful 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 motoneuron lesions.

Contrary to this we wished to apply FES also on certain patients with lower motoneuron lesions. In the course of gait, patients with lesions of motor roots, caused by diseases of the lumbosacral spine, above all herniae of intervertebral disci, manifest foot-drop (similarly as hemi - paretic patients). We chose 5 such patients with a one-side foot-drop, namely, 3 women an 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 peroneal nerve had been preserved as much as to enable us to apply such parameters of electrical stimulation as normally used with patients with upper motoneuron 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 FEPA-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

musculature cannot be prevented, though it might be slowed down to a certain extent.

Patients with a denervated musculature belonging to the peroneal 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 paralyisis due to an impaired lower portion of the spinal cord as a result of L I 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 three of them the musculature of the peroneal 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 20 ms, frequency 25 Hz, large disk surface electrodes were positioned above the musculature on the anterolateral side of the shank. By an appropriately selected duration of a pulse train and a 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 favourable 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 obsolete partial lesions of the nerves of peripheral type, there remains an unsolved problem of the therapeutic value of denervated musculature. 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.

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MOVEMENT CONTROL OF AN ATHLETE IN THE PROCESS OF RUNNING AIMED AT REHABILITATION OF MOTOR-SUPPORT SYSTEM

RATOV I.P., Pavlov L.V., Kriazev V.D., Bashkirov V.F., Selivanova T.G., Popov G.I.,

Abrosimov V.V., Laboratory of biomechanics of All-Union Scientific Research Institute, of Physical Education, MOSCOW USSR

Hypothesis

Injured lower extremities' rehabilitation can be accomplished in the process of running by means of purposefull condition modification of athletes' and environment interraction and muscle activity control electric stimulation.

Methods

12 middle and long distance runners underwent a 12-week treadmill running programm (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 analoge 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 a special optic electronic device (error ± 2 msec). Shock acceleration on athlete's waist was measured by "Bruel & Kj r" accelerometer. Cinematic characteristics were registered by "Actionmaster-500" camera. Film analysis was performed by an automatic analyser "Nac sportias". 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. Differencies 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 insignificant. Vertical occillation of general center of gravity during injured extremity support phase decreased.

Discussion

Rehabilitation intensification after knee and ankle joints' injuries were determined by creation of comfortable conditions of running, which provided shock load decrease and economic muscle work. Electric stimulation of m. gastrocnemius fixed the extremity position at the moment of its contact with support, which lessened joint's flexion, vertical occillations and provided for better absorbtion of shock influencies. Movement control in running improved intermuscular coordination and decreased the differencies of strong capacity of injured and sound legs.

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ELECTROMYOGRAPHIC INVESTIGATION OF THE DIAPHRAGM AND INTERCOSTAL MUSCLES IN TETRAPLEGICS

J.R. Silver, National Spinal Injuries Centre, Stoke Mandeville Hospital, Aylesbury, England.

R.P. Lehr, The Department of Anatomy, School of Medicine, Southern Illinois University, Carbondale, Illinois, U.S.A.

INTRODUCTION

The current investigation was undertaken to correlate the EMGs of the diaphragm and intercostals in order to demonstrate this local activity in the intercostal muscles, and to show that it was not transmitted from the diaphragm.

MATERIALS AND METHODS

Five male tetraplegics were studied. Only patients with complete lesions of the cervical cord as a result of trauma were investigated. Completeness of the lesion was determined by neurological examination, which revealed complete loss of sensation and motor function below the level of trauma. The patients performed a series of activities which included quiet and deep breathing, coughing and straining. Each patient completed several series for each recording site.

An intra-oesophageal electrode was utilised in three subjects for the diaphragm electromyographic recordings. The placement of the intra-oesophageal electrode was accomplished through the nasal orifice into the pharynx and subsequent swallowing by the patient. The recording apparatus was monitored as the patient alternatively swallowed and took a deep breath until the maximum electromyographic activity was obtained, at which point the lead to the electrode was held in place by securing it to the face with tape. Surface electrodes were placed on the left side of the supine patient from the third to the eighth intercostal spaces sequentially at distances of 9 and 18 cms. from the midsternal line. In three subjects fine wire electrodes were placed in the intercostal muscles directly beneath the surface electrodes to correlate the activity isolated within the surface electrodes. It has been demonstrated that the fine wire electrode has a range limited to no more than 1.5 mm. similar to the restricted area of detection of the bipolar needle electrode. The combination of intra-oesophageal and bipolar fine wire electrodes allows the differentiation between diaphragmatic and intercostal EMG. The amplifier-recording system consisted of four high-gain pre-amplifiers (Medelac A9) and a jet spray minograph (Siemans Medical). The preamplifiers had bandpass filters of 100-1000Hz. and the gain was adjusted to 250 V/cm. The galvanometers of the recorder had a resonance frequency of approximately 500 Hz. No attempt was made to quantitate the EMG activity but instead close attention was given to the presence or absence of activity and the onset of activity in each of the EMG channels. The EMG activity was related to the simultaneously recorded spirometry tracing to correlate with the phases of respiration.

RESULTS

The aim of the investigation was to demonstrate that the activity detected with the surface electrodes over the intercostal muscles arose locally from these muscles. Electromyographic evidence of the activity of the intercostal muscle during a deep breath is shown in Figure 1a. There was synchronous onset of activity during inspiration in both the fine wire and surface electrodes of the eighth intercostal muscle. This onset contrasts with the asynchronous initiation of activity in the diaphragm. The continuation of the activity beyond the expected inspiration into the expiratory period should be noted in the intercostal muscles. Additional continuous simultaneous after-discharge spikes may be observed in both channels beyond the major activity but were not present in the diaphragm recording. This was evidence of local activity of the intercostals and not of the diaphragm radiating through and being recorded by the intercostal electrodes. While the continuous spike activity seen in the intercostal muscles was not present in every recording it was seen so frequently in both the acute and chronic patients to be a prominent feature.

Figure 1b shows two breaths in patient E. There is well marked diaphragmatic activity at the initiation of each breath. The intercostal activity is more marked in the third than in the fourth space and is present when the diaphragm is silent.

DISCUSSION

The results confirm previous findings that the activity detected in the intercostal muscles is arising within these muscles since it was detected with wire, and wires only detect activity 1.5 mm. from their 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 removed 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 paralysed muscles, and we believe it to be reflex, the afferent limb 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.

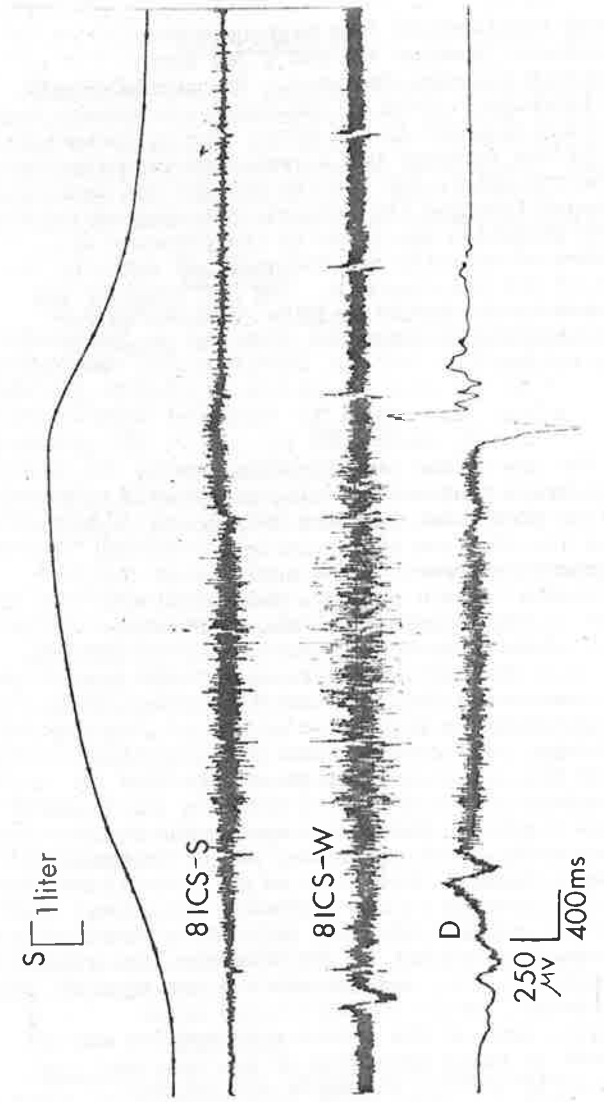

Figure 1a. Channel 1 - S, spirometer trace; Channel 2 - 8 ICS-S, eighth intercostal space surface electrode; Channel 3 - 8ICS-W, eighth intercostal space, wire electrode; Channel 4 - D, diaphragm.

The full text appears in J. Neurol, Neurosurg, and Psychiat. Vol 44, pps. 837-841, 1981.

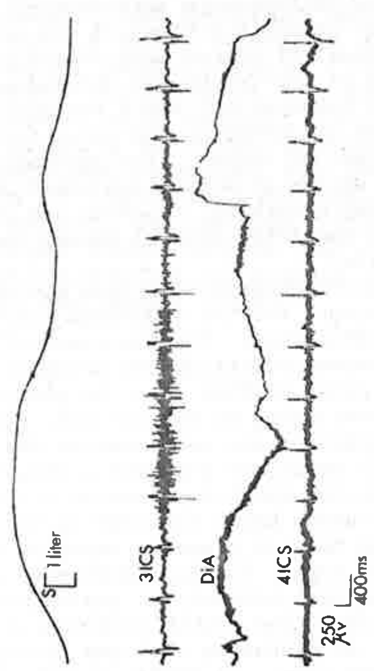

Figure 1b. Channel 1 - S, spirometer trace; Channel 2 - 3ICS, third intercostal space, wire electrode, Channel 3 - Dia, diaphragm, intra-oesophageal electrode; Channel 4 - 4ICS, fourth intercostal space, wire electrode. Activity detected with wire in the third and fourth spaces is well removed from the diaphragm.

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DYSPNOEA DURING GENERALISED SPASMS IN TETRAPLEGIC PATIENTS

J.R. Silver, National Spinal Injuries Centre, Stoke Mandeville Hospital, Aylesbury, England.

R.P. Lehr, The Department of Anatomy, School of Medicine, Southern Illinois University, Carbondale, Illinois, U.S.A.

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 C4 to T6 as the last intact segmental level have 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 spasms. 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 some 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 electrodes for the intercostal muscles. An intra-oesophageal electromyographic electrode was used to record the diaphragmatic EMG. The combination of intra-oesophageal and bipolar fine wire electrodes allows the differentiation of the diaphragmatic and intercostal EMG. Simultaneous spirometry was carried out, Silver and Lehr (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 their chest. The dyspnoea was a continuation of generalised body spasms which were initiated by sudden movements suc[illegible]s having their position changed, when they rec[illegible]ed passive movements to the arms or legs, or as part of another reflex, for example plantar 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 chest 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 spasticity develops 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 adapt; thus once a spasm has occurred subsequent spasms are more difficult to elicit 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 down-going 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 the spasm of the diaphragm ceases. The normal activity of the diaphragm ceases and an abnormal EMG interference pattern of increased amplitude is apparent. Concurrent EMG activity is noted in the seventh intercostal space EMG trace. It should be noted that onset of activity in the three EMG channels is not synchronised.

Detailed calculations of the duration of the diaphragmatic EMGs for five spasms in one patient provide a mean of 8.7 seconds with a range of 3.5 to 12.1 seconds.

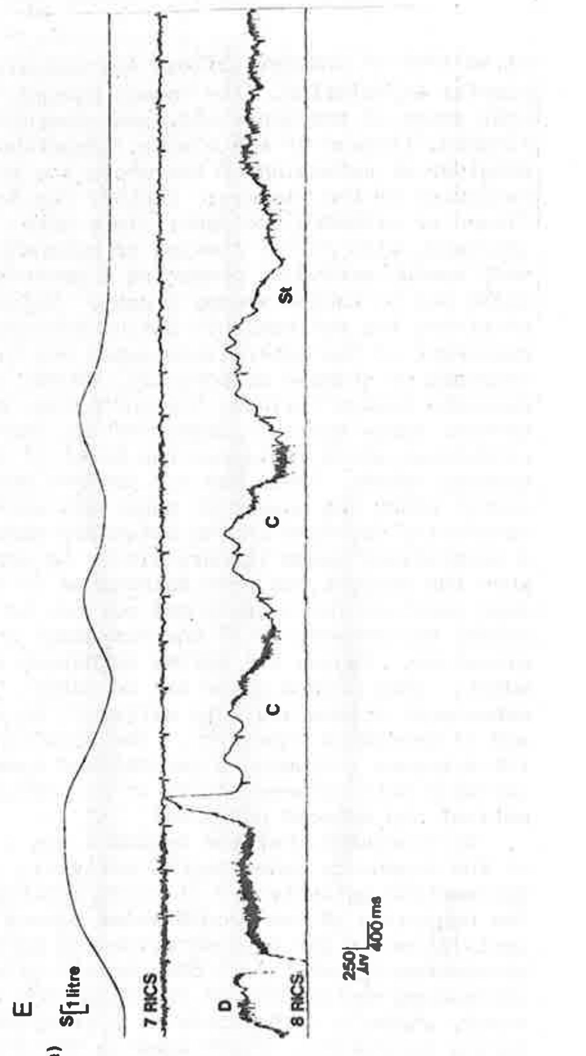

Figure a. Channel 1 - S, spirometer trace; Channel 2 - 7RICS, seventh right intercostal space, wire electrode; Channel 3 - D, diaphragm; Channel 4 - 8RICS, eighth right intercostal space. C - cough, St. - strain, Sp. - spasm. The trace is continuous.

CONCLUSIONS

We have not seen this form of dyspnoea 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.

The full text appears in J. Neurol. Neurosurg, and Psychiat. Vol 44, pps. 842-845, 1981.

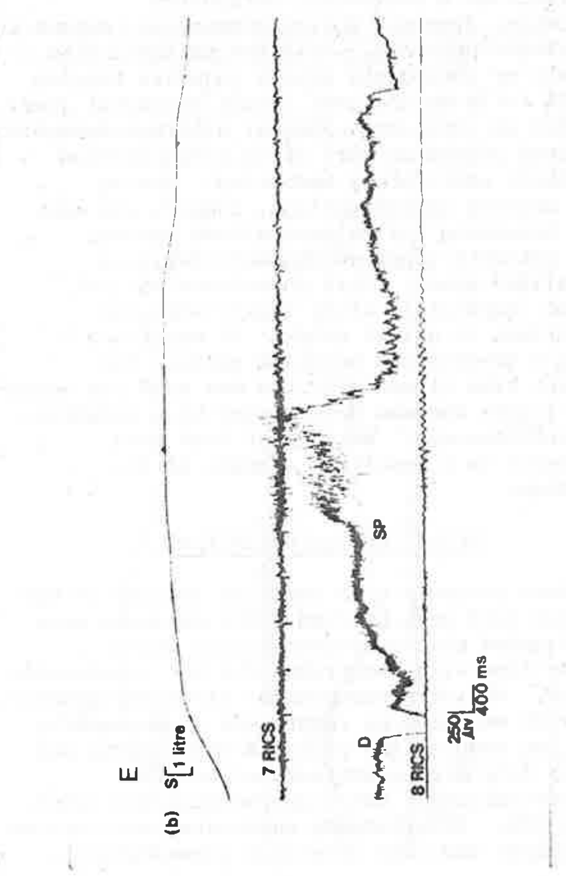

Figure b. Note the spasm of the diaphragm in trace b.

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ORGANOPHOSPHOROUS PESTICIDES AND NEUROMUSCULAR SYNAPSE TESTING

A. Jušić and S. Milić

Center for neuromuscular diseases, Zagreb and Department for neurology and psychiatry, Osijek, Yugoslavia

The purpose of this study was to evaluate electromyographic neuromuscular synapse testing and neurological examination for early detection of organophosphorous pesticides intoxication.

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, Jušić 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, Jušić et al. (1977).

METHOD

Neuromuscular synapse testing was performed according to technical modifications of the method described by Jušić et al. (1977). This development permited high constancy of recorded muscle signals inspite of voluntary contraction movements.

RESULTS

No significant difference between exposed groups and the control group was found.

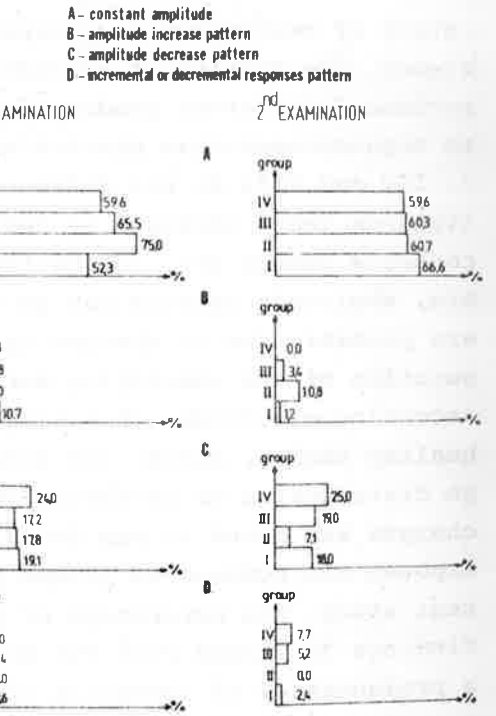

Fig. 4. Comparative analysis of four types of changes in evoked serial muscle potentials (A, B, C, D) 5 sec after volitional activation in Groups I-IV.

In two cases of suicidal poisoning during the standardized testing, which was repeated three times, there were no 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 organophosphorous 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.

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THE CORRELATION BETWEEN RESULTS OF SENSORY NERVE CONDUCTION AND TERMOGRAPHIC TESTS BY WORKERS EXPOSED TO LOCAL VIBRATION

Jadwiga Koczocik-Przedpelska, Szczesny Gorski and Elzbieta Powierza

Dept. of Pathophysiology of Locomotor Organs, Institute of Ortopaedic Surgery and Rehabilitation, Medical Academy, Poznan, Poland.

INTRODUCTION

We chose the electroneurographic and the termographic methods for early diagnosis of vibration syndrom. 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 temporary.

We have performed following examinations: 1/ sensory nerve conduction test of n. radialis, n. medianus and n. ulnaris in the region: finger-wrist, by using Disa electromyograph, 2/ termographic investigation of dorsalis and palmaris surface of both hands before and after cooling test.
The following parameters were analysed: 1/ the subjective and the objective response threshold in mA, 2/ the maximal amplitude of potential received from sensory nerve /in uV/, 3/ the conduction velocity in m/sec. Above mentionned parrmeters were measured for each finger separately. In such a way we have obtained 40 results for each person.

Termographic measurements we took twice: before and after cooling test. For the statystical 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 uV and the conduction velocity from 15 m/sec to 59 m/sec.

We present the typical examples of electroneurograms. The first /Fig. 1/ belong to worker by whom all conductability parameters were within normal limits.

The Fig. 2 presents electroneurograms of worker, by whom 28 of total 40 parameters surpassed normal limits + S.D. By this worker we did not state any evoked potential when stimulating IV and V finger, 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 termograms shows normal temperature of finger after cooling test.

The electroneurograms with low amplitude find their analogue in the termograms with characteristic lowering of temperature after cooling test.

Presented 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 electroneurogram and termogram evaluated in points number

The pathological changes in sensory nerve conduction are more marked as these in termograms. The concomitant use of electroneurography and termography helps in early detection of vibration syndrom.

[illegible] Results [illegible]

Fig.1

From the left to the right the electroneurograms obtained seperately from the first to the fifth finger.

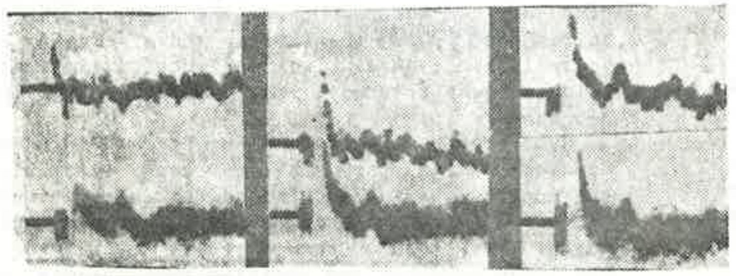

Fig.2

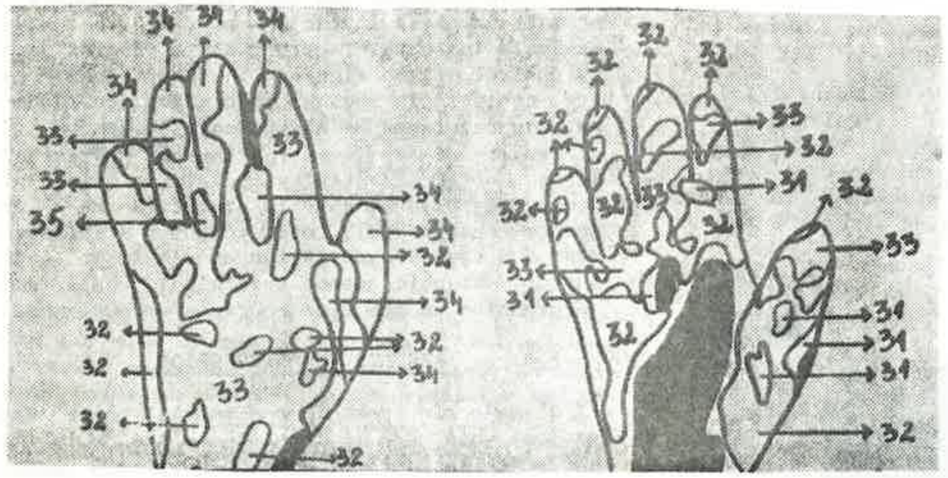

Fig.3

On the left- the temperature after adaptation to the environment.
On the right- temperature after cooling test.

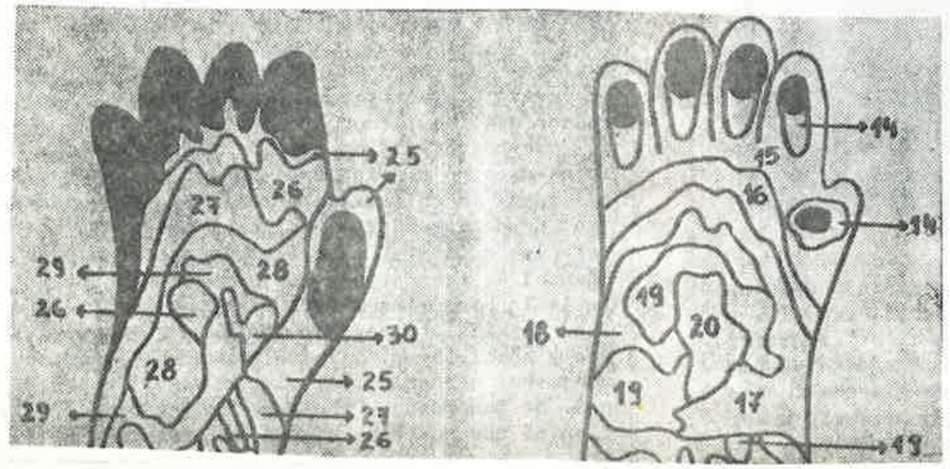

Fig.4

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STUDIES OF TRAINING TECHNIQUES AND MUSCULAR LOAD IN A SIMULATED WELDING TASK

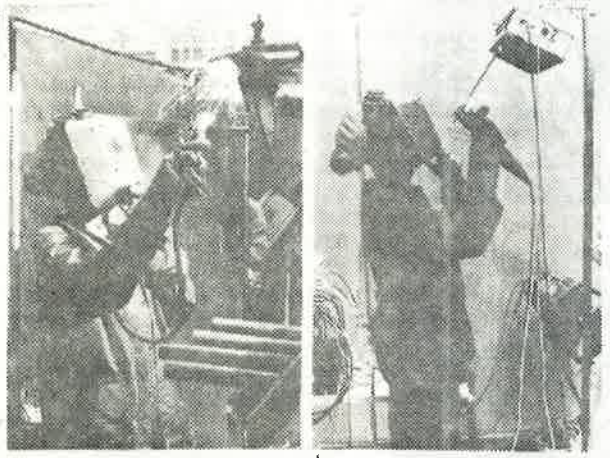
Roland Kadefors, Anders Arvidsson and Ingemar Petersén
Projekt Lindholmen Center and Laboratory of Clinical Neurophysiology, Sahlgren Hospital, Göteborg, Sweden

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 standardized welding tasks in 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 with 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 skill, we have used a simulator developed at the MRC 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.



a b

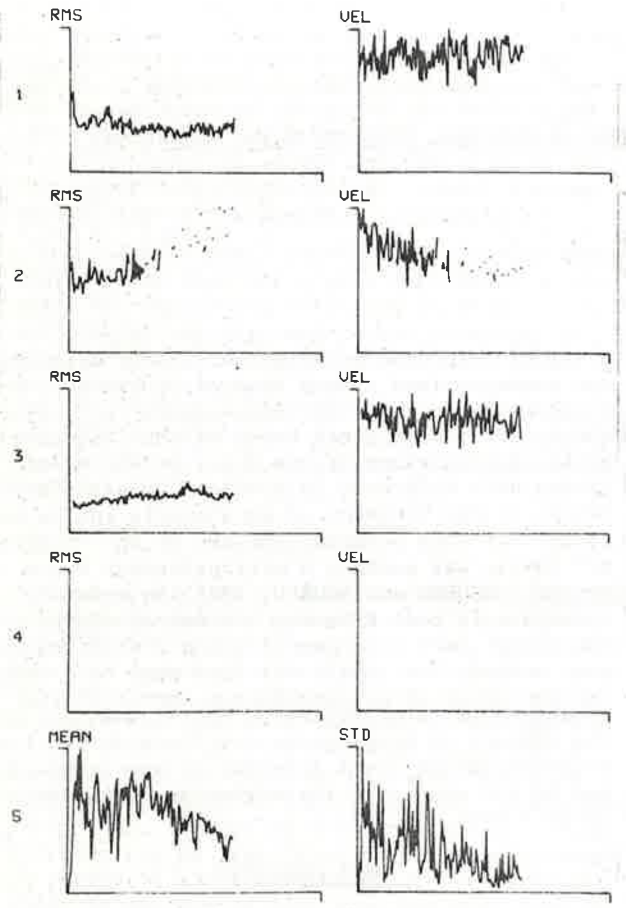
The simulator consists of an artificial welding joint with a light-sensitive device travelling underneath the joint. At the tip of a simulated welding rod there is a lightsource affixed. When the electrode tip is properly positioned at the joint, the light-sensitive device travels on at a realistic welding 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.

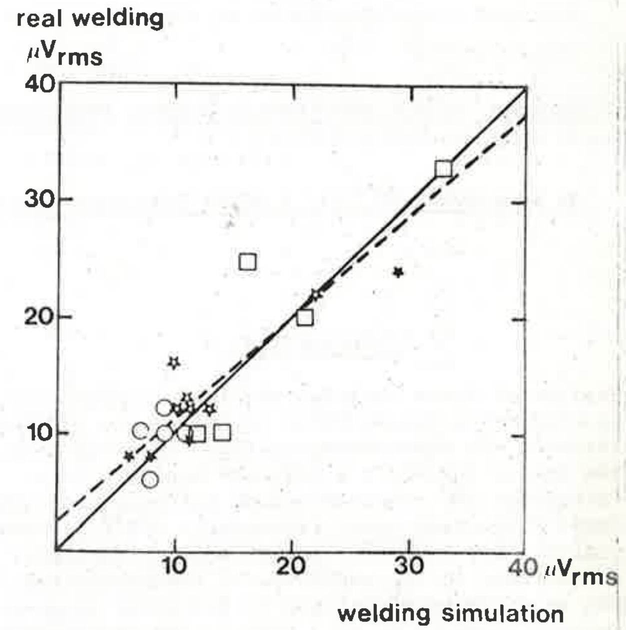
Myoelectric signals are collected in four muscles during the simulated task. Initially, the upper trapezius and the medial deltoid muscle, and forearm flexor and extensors are monitored. The signals are directly fed into a PDP 11/44 computer and analyzed subsequently. Of primary interest in the EMG part 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 EMG activity level in microvolts r.m.s., fatigue measure in relative mean frequency change (3), and endpoint stability in mean and standard deviation of simulator sensor output. Graphic 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 (left), as well as the corresponding fatigue plots (right). Omitted data points indicate data not accepted. It is 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 persons (five experienced welders) welded 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 resonably 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 components, is higher in overhead welding than in downhand welding.

The present project is concerned with studying the possibilities to assess level of skill using the simulator, and to investigate if the motor patterns of unexperienced and experienced welders differ using multiple discriminant analysis or related techniques. If this is the case, we will have a powerful method to evaluate the acquisition of skill in welding school and to contribute to the learning process from the point of view of reducing the muscular load in welding related to lack of experience. It is believed that such measures be of importance in order to provide for improved working techniques also in the long run, and therefore decrease the risk for workload related diseases in the musculo-skeletal system.

ACKNOWLEDGEMENT

This investigation is supported by the Swedish Work Environment Fund.

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MUSCULAR FATIGUABILITY OF AN HETEROGENEOUS MUSCULAR GROUP

Jacques Duchêne and Francis Goubel, Département de Génie Biologique, Université de Technologie de Compiègne, BP 233, F 60206 Compiègne Cedex

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 - MPF - determination). For isometric submaximal contractions, an increase in the amplitude of the integrated EMG as a result of fatigue is described by several authors (LIPPOLD et al., 1960 ; LIND and PETROFSKY 1979) and associated with failure in excitation-contraction coupling. Likewise, in the course of fatigue, several experiments have shown a decreasing in the MPF value (LINDSTROM et al., 1977 ; KOMI and TESCH, 1979) ascribed to reduction in muscle action potential velocity. In the present investigation, an attempt was made to evaluate fatiguability at the level of an heterogeneous muscular group, the triceps surae.

METHODS

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 120° 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 piezoresistive 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 % MVC to fatigue. During both the experiments the signal from the dynamometer was used to provide a visual reference to help the subject maintain the predetermined torque (isometric isotonic contractions). In the third experiment, the subject had to maintain a given value of EMG ("isoenergetic" 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 microprocessor based system (DUCHENE and GOUBEL, 1982) in order to calculate in real time the electrical energy of the signal over a 1s period using a shifting average method. The result was then sent to a vertical amplifier of an oscilloscope after digital to analog conversion. Feedback control was made by the subject in 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 32 spectra obtained from consecutive time windows of 0.5s duration. This resulting spectrum was defined by 256 points in module and phasis in the 4 Hz-516 Hz bandwidth. The module values were transmited to a microcomputer by means of a GPIB interface system ; then, the spectrum was divided into 20 equal bands. The computer calculated the total energy and the MPF of the mean spectrum and the relative energy of each bandwidth. The results were visualized in a graphical or a numerical form on a video display unit and printed on an alphanumeric/graphic printer.

RESULTS

. During brief isometric contractions the MPF was only found to be related 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 - 28 Hz) was found (see fig. 1).

. For isometric isotonic contractions, after some arrangements in the EMGs at the beginning of the test, two kinds of results were found : a continual increase in energy occured at the level of each muscle or "alternations" (decrease in EMG of a muscle associated with increase at the level of an agonist). In the first case, an increase in the relative energy of the first bandwidth was always found while in some tests, the classical decrease in MPF was not obvious. In the second case, alternations induced a large variability in the spectral parameters.

. For "isoenergetic" 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, ample variations and even "silences" in the EMGs of the uncontrol led muscles were found (see fig. 2). No significant variation in spectral parameters preceded these variations.

CONCLUSION

From these results it appears that, at the level of an heterogeneous muscular group, modifications in EMG spectral parameters 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 MPF and the interpretation of this spectral variation in terms of fatigue could be erroneus (see also HAGBERG and ERICSON, 1982). In some experimental conditions, the use of MPF 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 most part of energy is found around the MPF : thus, a slight variation in the median bandwidths can lead to notable fluctuations in MPF. 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.

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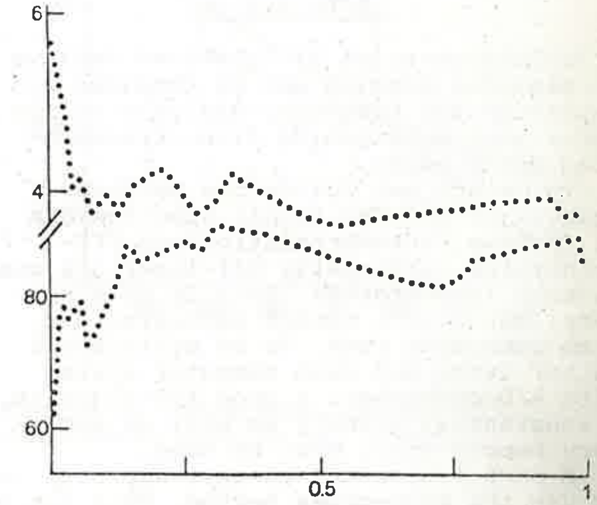

Fig. 1 : Changes in relative energy of the first bandwidth (upper trace, in %) and MPF (lower trace, in Hz) as a function of the total energy during brief isometric contractions.

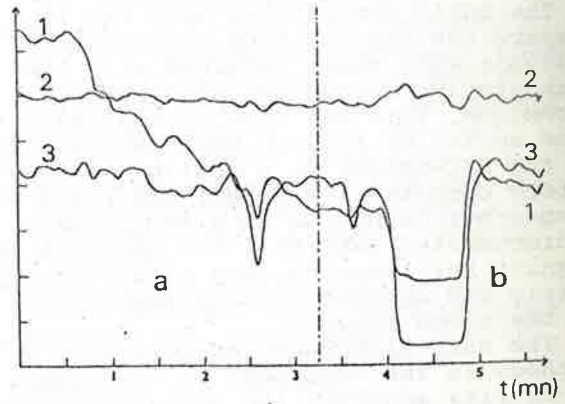

Fig.2 : EMGs and torque behavior during an isoenergetic experiment with the electrical energy of the gastrocnemius medialis taken as a reference 1:External torque,2:gastrocnemius medialis,3:soleus. - Note in a, the decrease of the torque and in b the appearance of a silent period in the soleus EMG

EMG FATIGUE ASSESSMENT BY FFT- AND ZERO-CROSS METHOD (A COMPARISON)

Bengt Eric Ericson

National Board of Occupational Safety and Health, Work Physiology Division, Box 6104, S-900 06 Umeå, Sweden.

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, autokorrelation 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-algoritm, 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 (HP 21 MX) and parts by zero-cross technique on a small table microcomputer (ABC-80). The analysed EMG signals were sampled from different isometric 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 (Hanning). Then a FFT-algoritm calculated a set of frequency component parameters from which the signal spectrum was obtained.

Each acheaved spectrum can be successfully characterized by a single parameter, the Mean Spectral Frequency (MSF). The MSF is the quotient between the first and the zero:th spectral moment i.e. is 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 acheaved "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 and 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) versus time were performed on both MSF and zero-cross data versus time to the function

$$f = A \exp(-Bt)$$

where A and B are regression coefficients. This function represents a decaying exponential process where $T = 1/B$ is the time constant.

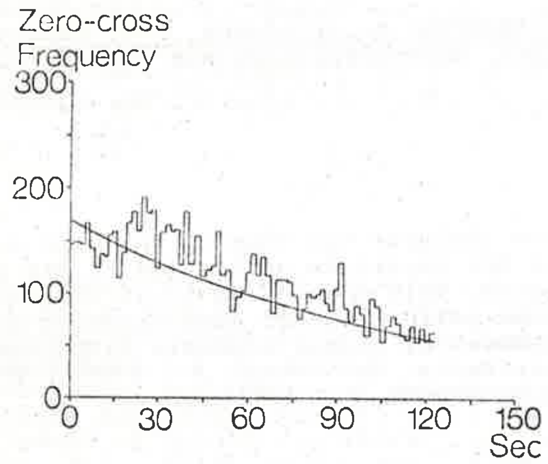

Figure 1. On-line plot from sustained isometric contraction (m. brachioradialis, 40 % MVC).

The smooth line in fig 1 is drawn by the computer as a check of the performed regression analysis.

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 analysed EMG signal.

Inevitably there is a much larger spreading in each measured zero-cross frequency point then 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 horisontal bar represents the average of the last 10 points. Major advantages 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.

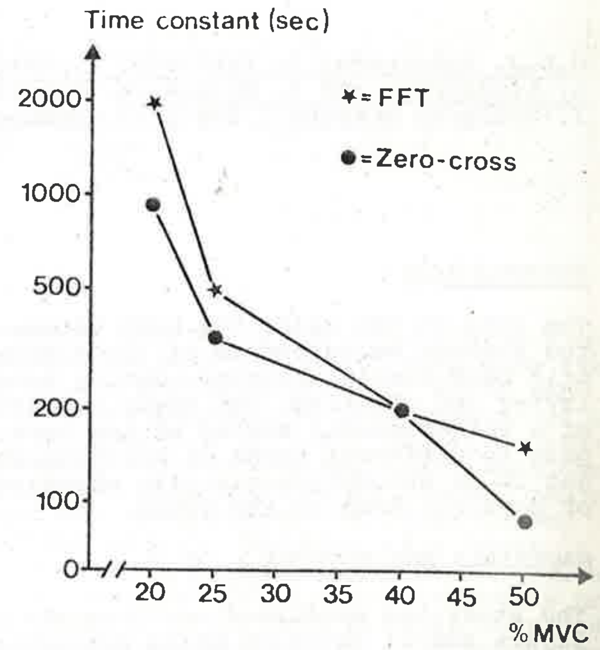

Figure 2. Comparison between FFT- and zero-cross results (m. biceps).

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MYOELECTRIC BACK MUSCLE ACTIVITY DURING BUILDING WORK. A COMPARISON OF BRICKLAYERS AND PAINTERS.

G.B.J. Andersson, I. Petersén, R. Örtengren, C. Ålmström, A. Arvidsson, P-O. Funnemark, L. Lindström, and R. Malmqvist. Departments of Clin. Neurophysiology and Orthop. Surg. I, Sahlgren Hospital, S-413 45 Göteborg, Sweden.

INTRODUCTION

The load on the spine has been estimated through measurements of the myoelectric back muscle activity, during brick laying and painting. The study is part of a 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 13 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 T8. The signals were amplified in body carried amplifiers and fed via a multi-wire 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 proceedures were entirely computerized. The purpose of the signal processing was to estimate the amplitude of the myoelectric signals (r.m.s.) and also changes in the power spectrum of the myoelectric signals. Mean values and standard deviations of the signal amplitudes were calculated for the different classes of work tasks. Also, the variations of the muscle loadings 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, Kadefors, Petersén (1977). The power spectrum analysis was done at the same time as the amplitude evaluation 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. & Örtengren, R. 1981).

Shifts 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 musclemetabolism. Therefore, there were no indications of local muscle fatigue.

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APPLICATION OF THE ORTHOSIS FOR CORRECTION OF FRONTAL BODY FLEXIONS IN PATIENTS WITH INFANTILE CEREBRAL PALSY

Semyonova K.A., Zhuravlyov A.M., Zarezankov V.G., Perkhurova I.S.

USSR. The all-Union of the Order of Red Labour Banner Research Institute for Common and Forensic Psychiatry named after V.P. Serbsky

Functional electrical stimulation of muscle is applied more and more wide for clinical treatment of infantile cerebral palsy.

Successful correction of abnormal leg motions has been achieved, for instance, by means of functional electrical muscle stimulation which improves to a great extent functional walking.

However motional disorders at the infantile celebral palsy are characterized not only by discoordination of leg motions but the pathological deviations of the body as well in the sagittal and frontal planes while walking.

Normalization of sagittal flexions of a body is possible with the help of treatment including surgical. These flexions depend significantly on a fixed forward pelvis bending caused by the spastic contracture of the hip joint flexors:i.e. musculus idiopsoas, musculus rectus femories and musculus tensor fasciae latae femoris. In this biomechanical state abnormal deviations of a bod.. in the sagittal plane probably play the role of a compensatory mechanism that promotes locomotion. By means of surgical intervention aimed at releasing the pelvis from the fixed bending position it is possible to achieve a decrease of these pathological flexions of a body. The most effective one an operation of bringing down the musculus rectus femoris has been proved. This method was suggested by one of the authors in 1974. By this time more than 15o patients have bben operated by the method.

It is much more difficult to normalize frontal flexions ofa body. Firstly, they are conditioned by disorder of a force muscle balance in the field of a hip joint. These muscles provide movements in in the frontal plane, they are adductors and abductors of a thigh. A positive Trendelenburg system is often marked in connection with above-stated disorder of muscle balance. Secondly, frontal flexions of a body are associated with the back muscles weakness. The above mentioned muscles limit displacement of a total center of gravity in the frontal plane.These muscles are as follows:sacrospinal muscle,musculus gluteus medius and musculus gluteus maximus. During normal walking flexions of a trunk in the frontal plane do not exceed ± 5°. At the same time these characteristics are increased in the most part of disabled patients approaching to ± 3o° in severe cases.These movements are caused by dystonia and force disbalance of trunk muscles and pelvis girdle.Voluntary correction of these flexions by the patient himself is impossible becouse of existting motional pathology.

In connection with above-stated an electrical orthosis for correction of pathological frontal movements of a body while walking has been worked out. The orthosis in assembly comprises a portable 2-channel electrostimulator,a sensor of frontal flexions of a body, belts for fixation and wires. The electrostimulator weighs 25o g, has the dimensions 33x75x158 mm and generates series of electrical stimulating impulses the amplitude of which is controlled disscretely from 25 to 7o V with maximal medium and instant powers respectively till o.12 and 3o VA. The period of longevity of impulses, delay and duration of stimulation is controlled smoothly within the range of 1o-25 ms, 3o- 3oo us, o-o.5 s and o.2 -2.o s respectively.Stimulation is indicated by a sound.

Electrodes of a woven fabric with a thickness up to 4 mm and working surface dimensions from 2ox2o to 4ox25o mm are placed at the clinically conditioned combinations on the following muscles: quadriceps, musculus gluteus medius and musculus latissimus dorsi. The sensor of frontal flexions comprises two controlled lockers, reacting to bendings with respect to the horizontal. The sensor is mounted on the back of a patient. At the bendings exceeding the established level the sensor generates triggering of an appropriate channel of the electrostimulator and excitation of muscles.

There were lo boys and 12 girls at the age from lo 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 inside, making jerky movements in the frontal plane. In all the patients a positive Trendelenburg symptom was marked. The course of treatment consisted of 2o 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 in expressed in the distinct visual decrease of the trunk flexions in the frontal plane during walking, Trendelenburg symptom disappearance and gait pattern improvement. A decrease of the body flexions on an average to 5o per cent has been registered, with normalization in 5 cases. Increase of stability, improvement of biomechanical parametrs of walking relief of walking and awareness of a motional comfort by the patient have been retained after the course of treatment gradually decreasing during a half of a year.