THIRD INTERNATIONAL CONGRESS OF ELECTROPHYSIOLOGICAL KINESIOLOGY

ANALYSIS OF MOTOR UNITS BY COMPUTER AIDED ELECTROMYOGRAPHY

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INTRODUCTION

In 1948, when Prof. Pinelli joined the Buchthal research team for two years, he made thousands of measurements on motor unit action potentials (MUAPs). These efforts, mainly concerned with MUAP recording methodology and the identification of variables which would best describe the MUAP, were done slowly and carefully by hand. Many EMG researchers have subsequently sharpened our awareness that quantification of the EMG is required for reliable clinical interpretation. And computer methods have evolved to handle the tedium of quantification. However, the facility with which we may now record and quantify the EMG has again forced us to ask what should be measured, how should it be measured, and how may it be interpreted.

It is increasingly clear that collaboration between clinical neurologists, neurophysiologists and computer systems analysts may be a prerequisite for comprehensive understanding and useful clinical application of new trends in EMG. In this spirit the authors, Prof. Pinelli (clinical neurologist and neurophysiologist) and Prof. Leifer (system analyst and neurophysiologist), have collaborated in the production of this report. It is our intention to present an estimate of the form and direction that developments in classical clinical EMG could take in the next 5 years. A survey is not attempted, rather, we wish to cite those experiments and methodologies which in our opinion will contribute most to our understanding of Motor Units and EMG generation. We shall state our opinion in the form of a hypothetical experiment.

GOAL

Our experiments are predicated on the concept that the principle goal of a clinical EMG examination is to obtain a full and unambiguous definition of the structure and functional status of the MU population of a specific muscle on a specific date. We are primarily interested in the MU (fig. 1) because it is the natural unit of activation. There are subunits, the single muscle fibers and neuromuscular junctions, which may be observed to yield information about the MU to which they belong. Following reinnervation, it may be necessary to consider grouped muscle fibers (Buchthal and Rosenfalck 1973; Pinelli 1975). There is also a supraunit, the whole muscle, whose activation pattern may be observed to infer information about the number and form of the MUs of which it is composed.

There are two views of the MU which guide us. The first, figure 2ab, illustrates the perceptions of the anatomist/physiologist. The second view, figure 2cd, represents the MU from the point of view of a systems/signal analyst.

CLINICAL PROBLEM

We suggest that there are two basic principles of clinical EMG which must be accounted for in the development and application of new methodologies.

1) Muscular somatotopic Specificity
2) Motor Unit Activation Specificity

"Muscular Somatotopic Specificity" acknowledges the fact that EMG properties and clinical diagnosis are functions of the specific muscle under investigation, its innervation, function, and utilization. Accordingly, an investigative procedure which can be applied to virtually any muscle of the body will find the widest acceptance. This flexibility is essential to comparative EMG and to full realization of the diagnostic potential of quantitative EMG. We wish to draw a sharp distinction between muscle fiber, MU, and whole muscle measurement methods which are intended to substantiate or make a clinical diagnosis in accord with "Muscular Somatotopic Specificity", and those methods which contribute more general information about biophysical properties of muscle.

"Motor Unit Activation Specificity" acknowledges the fact that the recruitment of single MU's can be modality specific. In disease states, units may be differentially affected (as in the so-called selective MU disorders ascribed to King Engels 1968). According to clinical priorities, we are most concerned with voluntary activation. Tonic and reflex muscle activation studies provide additional information about the motor neuronal substrate upon which voluntary control is superimposed. Supramaximal electrical stimulation and maximal voluntary activation of muscle are important procedures in that they provide a basis for standardization of graded activation. We are skeptical about graded-stimulus-amplitude methods for eliciting single units (McComas et al. 1971). Milner-Brown and Brown (1976) have found that motor unit counts so obtained are in general 'erroneously high'. Graded-time-interval activation methods have proved useful for nerve conduction latency spectrum estimation where no claims are made regarding single unit properties (Hopf 1964, Leifer and Meyer 1976). Averaged single unit surface potentials have been well used in conjunction with supramaximal stimulation of the M-response and histological estimates of the nerve conduction spectrum to predict the number of MU's in the test muscle (Lee et al. 1975). Artificial activation methods provide needed information though technical constraints on their universal application are occasionally severe.

NEUROPHYSIOLOGICAL PROBLEM

Major issues in neurophysiology impinge upon our investigation of MU-EMG.

1) Size Principle
2) Tonic versus Phasic Function
3) Biophysics
   a. motor neuron excitation dynamics
   b. action potential dynamics
   c. neuromuscular junction dynamics
   d. excitation/contraction coupling dynamics.

While it is not generally feasible to study these issues in the context of clinical EMG, we believe there exists a growing need to know quantitatively in what way MU-EMG depends on such variables. As we shall shortly emphasize, predictive modeling of MU properties from subunit and supraunit observations is dependent on quantitative understanding of these physiological variables.

TECHNICAL PROBLEM

We address ourselves here to the need for better signal and system analysis. Few if any basic problems exist now with regard to EMG signal acquisition (electrodes, pre-amplifiers, filters, sampling and digitizing). In this context we wish to focus our attention on the utilization of small computer facilities for the following purposes:

1) Experiment Control
2) Signal Analysis
3) Simulation of Physiological Systems

This classification scheme is useful in defining the requirements for a computer technology solution to our clinical and neurophysiological problems. A single investigation may use the computer in one or all of the above modes. To date, computer methods in EMG have dealt primarily with digital signal acquisition and signal analysis. We want to stress the value of computer controlled experimentation as a means to obtain better and occasionally unique data. With equal emphasis, we note that EMG simulation is an increasingly relevant approach to our theoretical understanding of EMG generating mechanisms and the inference of diagnostic information.

THE EXPERIMENT

A reasonably full undestanding of the MU and EMG generation is obtained when we have quantitative data for each of the signal paths described in figure 2c,d. We propose to use an experimental system as illustrated in figure 4. Today, such a system exists only piecewise or not at all. For illustrative purposes, real and hypothetical results will be given in the following section.

Electrodes

Six electrodes are required for our experiment: 1 for grounding, 1 for reference, 2 for recording, and 2 for stimulating. The first is a large surface electrode mounted proximal to the muscle under examination and used as system ground reference. The second electrode, also on the surface, is small and placed over the distal tendon of the muscle. This electrode is used as the common reference for differential amplification of the two signal source electrodes. Electrode number 3 is an implanted wire macro electrode. The wire itself may be 25μ stainless steel free of insulation over the entire length of its exposure within the muscle under examination. This electrode may be placed via a 27 gage (or smaller) needle using a small hook shape at its distal end to resist withdrawal when the needle is removed (Basmajian 1974). Differentially amplified with electrode 2, this macro electrode will be used for summated EMG, electrically evoked M-responses, and in conjunction with the 4th electrode for averaged MUAP-EMG. The 4th electrode is a small needle electrode for single muscle fiber electromyography (Ekstedt 1964). Differentially amplified with surface electrode 2, this electrode signal is used for measurement of the junctional delay pdf, SFAP, single unit spike train statistics, and in conjunction with the macro electrode as the triggersource for average MUAP identification (Lee et al. 1975). Should this methodology prove successful, electrodes 3 and 4 could be combined into a single device.

Electrodes 5 and 6 are used for electrical stimulation of the motornerve. Both electrodes are surface mounted, one distal (5) and one proximal (6). The interelectrode separation should be as large as possible with 10 cm representing a rough minimum. These electrodes are used for two point paired supra-maximal stimulation of the nerve. The resulting M-responses (via electrode 2) are used for estimation of the axonal conduction velocity probability density function, pdf (Hopf 1964, Leifer et al. 1976a,b) and the whole muscle auto-power-spectrum.

SIGNAL ANALYSIS

Muscle Response M(t)

In our experiment the integrated muscle response M(t) is obtained in two ways via the wire macro electrodes. Firstly, it is obtained by voluntary activation. This may be at graded effort levels, though we are primarily interested in maximum effort activation. In accord with our utilization of linear systems theory, we compute the Fourier auto-power-spectrum of this signal; figure 5a,b. Lacking evidence to the contrary, one initially assumes that maximum voluntary activation is the system response to a maximum amplitude, noise command C(t) which asynchronously activates the motor unit population. Because C(t) cannot be directly observed, M(jw) does not contain phase information about the conductive path from C to M.
For comparative purposes, we also obtain a measure of the maximum M(t) response to a fully synchronized impulse input. This is the response to electrical stimuli applied at electrodes 5 and 6. In this case M(jw) does contain phase and delay information descriptive of the conductive path. The ratio of the voluntary $M^{v}_{max}(jw)$ to the stimulus evoked $M^{s}_{max}(jw)$ provides an estimate of the percent functional muscle available to voluntary drive. In normal conditions this ratio is expected to approach 100%. $M^{s}_{max}(jw)$ will also be used to estimate the number of functional motor units in the test muscle.

Unit Response U(t)

In order to obtain MUAPs over a wide range of activation levels, we chose to use a slight modification of the averaging method as demonstrated by Lee et al. 1975. The single fiber electrode is used as a probe to isolate the activation time for a single motor unit. The SFAP is passed through a digital window discriminator whose pulse output is used to trigger averaging of the macro electrode interference pattern. We do this with 2 synchronous delay buffers. Depending upon the level of the activation, 50-500 averages are sufficient to bring the signal to noise ratio within acceptable limits (preferably 10/1 or better). The implanted macro electrode is chosen over a surface electrode to: 1) avoid signal attenuation and lowpass filter characteristics of the superficial tissues; 2) to have less geometric bias than the more remote asymmetrically placed surface electrode; 3) allow direct comparison of U(jw) and M(jw) from the same electrode; 4) minimize signal contamination from remote muscles, Representative data are presented in figure 6. If one assumes that each of the 20 or so MUAPs obtained by averaging is equally represented in the muscle, then the mean U(jw) may be divided into $M_{max}(jw)$ to obtain an estimate of the number of units active within the muscle. This can be done for the voluntary $M^v_{max}(jw)$ and the stimulus evoked $M^s_{max}(jw)$.

Single Fiber Action Potential F(t)

The single fiber action potential (SFAP) is itself of interest for we can use it to stimulate the MUAP (Boow and Criep 1976). Representative SFAPs, F(t), and their corresponding Fourier auto power spectra are presented in figure 7ab. Based upon biophysical models of action potential generation (Plonsey 1969), the muscle fiber conduction velocity may be estimated from the SFAP duration. The MUAP may be simulated with knowledge of F(jw), the muscle fiber conduction velocity probability density function (pdf), the junction latencies (jitter) pdf, the electrode geometry, and assumption of a geometric model for the MU. This last assumption is closely related to MU density such that one may deduce, quantitatively, MU density by simulation of the MUAP and adjustment of the geometric density to obtain an optimal fit between the observed and simulated MUAPs.

Junctional Latencies J(t)

Following the considerable body of work by Ekstedt and Stalberg, a methodology is now established for estimation of the junctional latency pdf. While these authors have been able to further identify physiological substrates for the jitter phenomena, it is the net jitter which is required for our signal analysis. J(t) and its Fourier Transform J(jw) constitute a junctional transfer function which is needed for both MUAP and EMG stimulation. J(jw) has the general character of a low pass filter (figure 8).

Axon Latencies A(t)

Following the methodology initiated by Hopf 1964 and computerized by Leifer et al. 1976, it is possible to estimate the axonal conduction latency pdf using two point paired electrical stimulation of the motor nerve. The paradigm is illustrated in figures 9 and 10.
This technique is based upon two point supramaximal stimulation of the peripheral nerve at progressively longer interstimulus intervals. Due to conduction velocity dispersions effects, the response R1 recorded following distal stimulation at SI will differ from the response R2 following proximal stimulation at S2. If the two stimuli (S1,S2) are paired at successively larger interstimulus intervals (ISIs), the combined effects of dispersive neural transmission and antidromic blockage will lead to an increasing R2 response at increasing ISIs until the R2 response is equivalent to that following a single S2 stimulus. In figure 9b we see the case in which S2

occurs before the antidromic action potential wave front elicited by S1 has had time to pass by the S2 electrode. All action potentials elicited by S2 are occlusively blocked. No response (R2) is recorded. Response R1 is normal. In figure 9c the ISI is longer such that some percentage of the action potentials conducted in response to S1 have passed by the S2 electrode, and the corresponding axons have recovered sufficiently to be excited by S2. The S2 action potential wave front then conducts freely, and dispersively, to contribute a second portion to the combined R12 response. At very short ISIs the R2 portion of R12 is zero. At progressively longer ISIs, R12 grows to approximate the algebraic summation of an R1 + R2 responses. Williams (1973) has demonstrated that the effective transfer function of a dispersively conducting nerve bundle may be estimated by the Fourier Transform of the nerve latency pdf. Leifer and Meyer (1976) have investigated this concept in some detail for peripheral motor nerve bundles. Representative data are presented in figure 11. The dominant feature of the transfer function is that of a low pass filter down -3db at 140 Hz/meter. Peaks and valleys in the function are caused by truncation of the pdf and reflect both the sharpness of the truncation and the width of the truncation. These window effects are present above 400 Hz/meter for normal subjects. First side lobe amplitudes are generally down 30-40 db. A gaussian latency pdf is a good approximation to the experimental data of control subjects ($X^2 = 31.4, \alpha = .01, n = 28$). In the frequency domain, the gaussian transfer function for the nerve bundle appears valid up to about 400 Hz/meter.

Command Input C(t)

The actal input to the motoneuron pool cannot be observed. The output $c_i(t)$ of each motoneuron can, however, be measured using the MUAP or SFAP train to form the inter-potention-interval histogram (the estimated MU activation pdf). An ensemble of such pdfs can be used in conjunction with our system functions to simulate the response to an assumed voluntary command (De Luca 1973, 1976; Agarwahl 1973, Lindstrom 1976). This approach is of particular interest when one wants to infer the intention of a voluntary command from the observable EMG. This is the case where the EMG is used to control a prosthetic device.

SUMMARY

We attempt to illustrate the sense in which our knowledge of Emg generation and the MU in particular is specified by our ability to quantify the signal transmission fuctions of figure 2. The precision and validity of our linear systems analysis determines the extent to which we can predict the basic structural parameters of the MU population which we have set out to obtain. These normally unobservable structural parameters are bound to the functional signal dynamics which we observe electrophysiologically. The computer is an essential tool for quantitative analysis and simulation. We suggest that the complexity of the physiological system we study justifies the complexities of computer methodology for its undestanding.

ACKNOWLEDGEMENT

We wish to acknowledge the contribution of many authors towards our current state of understanding, though we have not attempted a compendium of their important contributions. Examples have been chosen for illustrative purposes. Professor Pinelli is at the University of Pavia and Professor Leifer is at the Biomedical Engineering Institute of the Swiss Federal Institute of Technology (after September 1976 he will be at Stanford Univ., Dept. of Mechanical Engr. Design). Data shown in our figures were obtained in the Neurology Department of the Cantonal Hospital of Zürich.

SOME DEDUCTIONS FOR A CORRECT CLINICAL EMG

We have now to define what essentially is the real meaning of our contribution in the investigation of normal and pathological conditions of the neuromuscular system. The aim is of knowing the nature and the degree of a pathophysiological state; the methodologies we have outlined are reliable for testing the natural course of the disorder and its reactivity to different kinds of treatment.
Therefore we are now in the position to analyse which are the technics that are to be developed in clinical electromyographical investigation. We share the opinion of many other clinical neurologists and myopathologists (K. Engel), that the actual efficiency of most laboratories of clinical EMG is very poor. Their methodological approach is not reliable, the results are inadequate, their evaluation is too subjective; likewise the final conclusions, expressed in the answers that from the EMG laboratory come to the clinical department, are condensed in a few statements, inspired by last century nosological hypotheses, that today have any longer definite sense. We are concerning, for instance, with wordings like: «electromyographical findings of myogenic type» (versus neurogenic one).
In order to avoid such a confusing and restrictive «intuitive clinical interpretation», we are now justified to suggest the following methodological standardisation in clinical EMG routine, concerning the field of voluntary activity and motor nerve stimulation.

1) - First at all, according to the criteria of somatotopic specificity, the clinical neurologist has the task of indicating the muscles where the investigation must be done.

2) - A) The electromyographical investigation will be carried out: averaged M.U.a.ps. will be recorded by the macro-electrode and triggered by a.ps. recorded by the selective electrode; the mean value of the various M.U.a.p. parameters for at least ten of such M.U.a.ps. will be calculated (see p. 20).

B) EMG of maximal voluntary contraction will be recorded and its auto-power spectrum will be obtained.

C) The maximal M response will be elicited with stimulation of at least two points of the nerve; the corresponding auto-power spectrum will be obtained according to the methodology previously presented.

D) The paired stimulation will be applied and velocity response density functions will be calculated.

E) From the EMG recorded by means of Ekstedt-Stålberg's electrode during voluntary activity the probability density function for junctional latency will be calculated.

3) - The following informations will be transmitted to the clinical department.

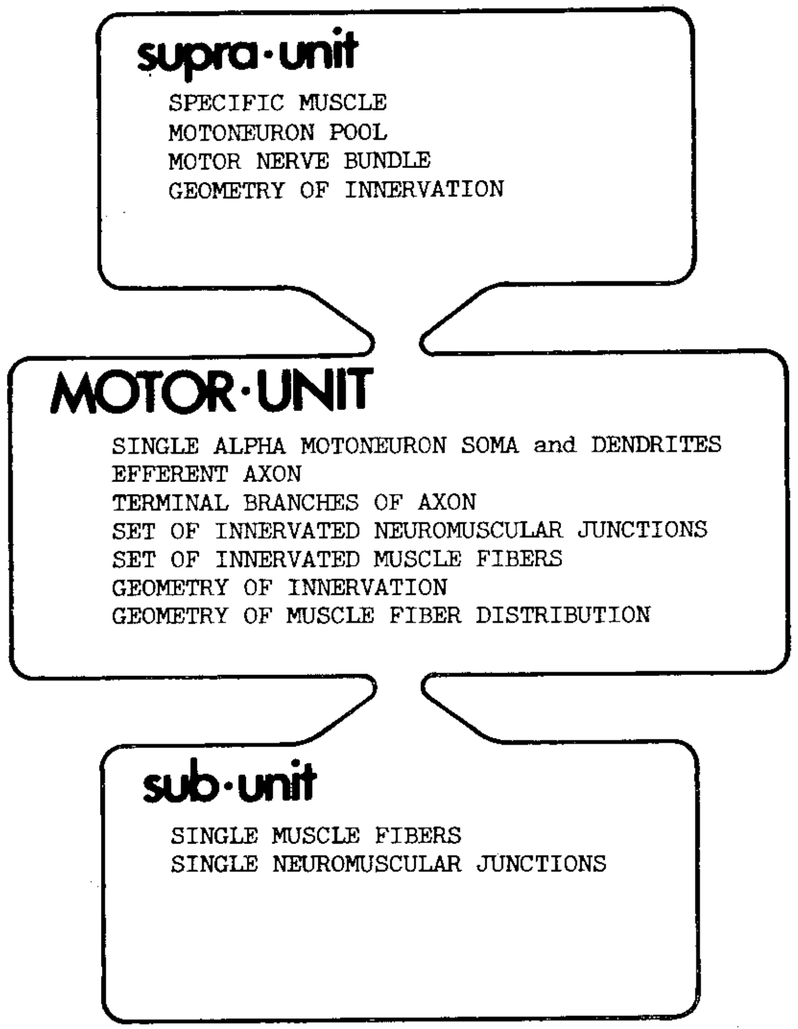
A) The muscle fibers density within the Mu.U. (muscular unit) is in the normal range, or below it or over it.

B) The number, firing frequency and a.p. area of M.U.s elicited by maximal voluntary contraction is in the normal range or below it (the calculated percentage will be reported).

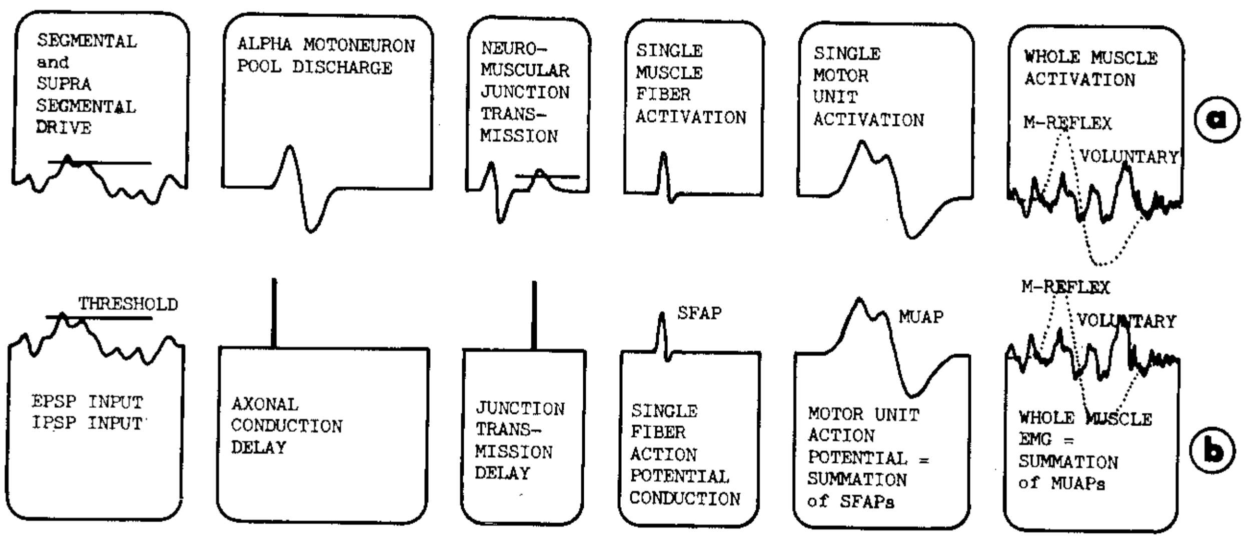
C) The number, nerve conduction velocity dispersion and a.p. area of M.U.s elicited by artificial electrical motor nerve excitation is in the normal range or below it (the calculated percentage will be reported).

D) Maximal and minimal motor conduction velocities will be given (see p. 23).

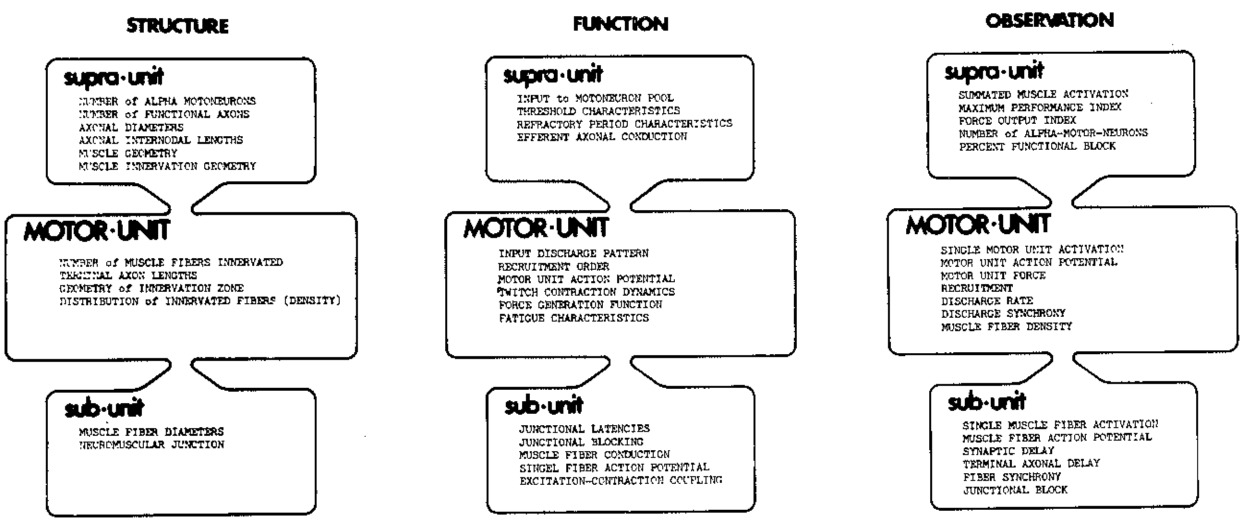
E) Variability of junctional latency will be reported.
These quantified findings will be combined with i/t curves findings, with the results of myographic investigation performed according to Taylor's and Hainaut's methodologies, with sensory neurography findings and eventually with muscle biopsy findings. On the basis of these data and all clinical and genetical data, the right diagnostical and prognostical classification may be achieved or useful contribution will be so brought to the knowledge of the yet unknown field of many neuromuscular disorders.



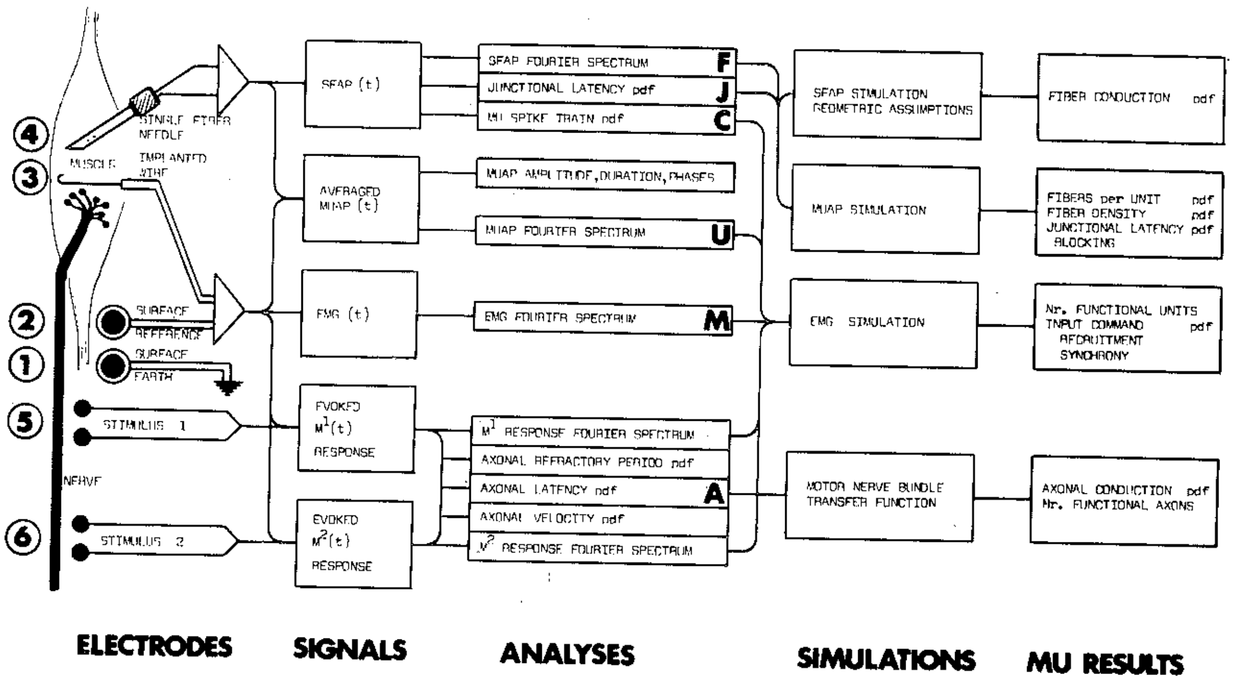
1.


SEGMENTAL and SUPRA SEGMENTAL DRIVE
ALPHA MOTONEURON POOL DISCHARGE
NEURO-MUSCULAR JUNCTION TRANS-MISSION
SINGLE MUSCLE FIBER ACTIVATION
SINGLE MOTOR UNIT ACTIVATION
WHOLE MUSCLE ACTIVATION
M-REFLEX
VOLUNTARY
a
THRESHOLD
EPSP INPUT
IPSP INPUT
AXONAL CONDUCTION DELAY
JUNCTION TRANS-MISSION DELAY
SFAP
SINGLE FIBER ACTION POTENTIAL CONDUCTION
MUAP
MOTOR UNIT ACTION POTENTIAL = SUMMATION of SFAPs
M-REFLEX
VOLUNTARY
WHOLE MUSCLE EMG = SUMMATION of MUAPs
b

2.

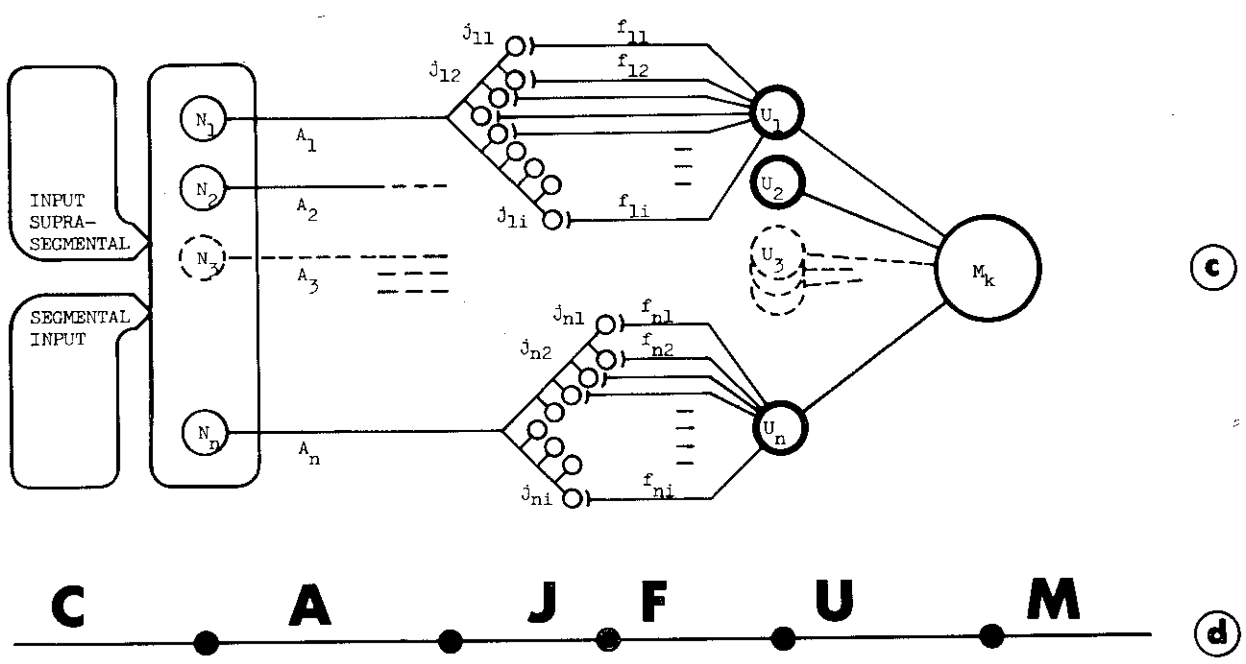

STRUCTURE
supra·unit
NUMBER of ALPHA MOTONEURONS
NUMBER of FUNCTIONAL AXONS
AXONAL DIAMETERS
AXONAL INTERNODAL LENGTHS
MUSCLE GEOMETRY
MUSCLE INNERVATION GEOMETRY
MOTOR·UNIT
NUMBER of MUSCLE FIBERS INNERVATED
TERMINAL AXON LENGTHS
GEOMETRY of INNERVATION ZONE
DISTRIBUTION of INNERVATED FIBERS (DENSITY)
sub·unit
MUSCLE FIBER DIAMETERS
NEUROMUSCULAR JUNCTION
FUNCTION
supra·unit
INPUT to MOTONEURON POOL
THRESHOLD CHARACTERISTICS
REFRACTORY PERIOD CHARACTERISTICS
EFFERENT AXONAL CONDUCTION
MOTOR·UNIT
INPUT DISCHARGE PATTERN
RECRUITMENT ORDER
MOTOR UNIT ACTION POTENTIAL
TWITCH CONTRACTION DYNAMICS
FORCE GENERATION FUNCTION
FATIGUE CHARACTERISTICS
sub·unit
JUNCTIONAL LATENCIES
JUNCTIONAL BLOCKING
MUSCLE FIBER CONDUCTION
SINGEL FIBER ACTION POTENTIAL
EXCITATION-CONTRACTION COUPLING
OBSERVATION
supra·unit
SUMMATED MUSCLE ACTIVATION
MAXIMUM PERFORMANCE INDEX
FORCE OUTPUT INDEX
NUMBER of ALPHA-MOTOR-NEURONS
PERCENT FUNCTIONAL BLOCK
MOTOR·UNIT
SINGLE MOTOR UNIT ACTIVATION
MOTOR UNIT ACTION POTENTIAL
MOTOR UNIT FORCE
RECRUITMENT
DISCHARGE RATE
DISCHARGE SYNCHRONY
MUSCLE FIBER DENSITY
sub·unit
SINGLE MUSCLE FIBER ACTIVATION
MUSCLE FIBER ACTION POTENTIAL
SYNAPTIC DELAY
TERMINAL AXONAL DELAY
FIBER SYNCHRONY
JUNCTIONAL BLOCK

3.



SINGLE FIBER NEEDLE
MUSCLE
IMPLANTED WIRE
SURFACE REFERENCE
SURFACE EARTH
STIMULUS 1
NERVE
STIMULUS 2
SFAP (t)
AVERAGED MUAP (t)
EMG (t)
EVOKED $M^1(t)$ RESPONSE
EVOKED $M^2(t)$ RESPONSE
SFAP FOURIER SPECTRUM
JUNCTIONAL LATENCY pdf
MU SPIKE TRAIN pdf
MUAP AMPLITUDE, DURATION, PHASES
MUAP FOURIER SPECTRUM
EMG FOURIER SPECTRUM
M^1 RESPONSE FOURIER SPECTRUM
AXONAL REFRACTORY PERIOD pdf
AXONAL LATENCY pdf
AXONAL VELOCITY pdf
M^2 RESPONSE FOURIER SPECTRUM
F
J
C
U
M
A
SFAP SIMULATION GEOMETRIC ASSUMPTIONS
MUAP SIMULATION
EMG SIMULATION
MOTOR NERVE BUNDLE TRANSFER FUNCTION
FIBER CONDUCTION pdf
FIBERS per UNIT pdf
FIBER DENSITY pdf
JUNCTIONAL LATENCY pdf
BLOCKING
Nr. FUNCTIONAL UNITS
INPUT COMMAND pdf
RECRUITMENT
SYNCHRONY
AXONAL CONDUCTION pdf
Nr. FUNCTIONAL AXONS
ELECTRODES
SIGNALS
ANALYSES
SIMULATIONS
MU RESULTS

4 a, b



INPUT SUPRA-SEGMENTAL
SEGMENTAL INPUT
N_1
N_2
N_3
N_n
A_1
A_2
A_3
A_n
j_{11}
j_{12}
j_{1i}
f_{11}
f_{12}
f_{1i}
j_{n1}
j_{n2}
j_{ni}
f_{n1}
f_{n2}
f_{ni}
U_1
U_2
U_3
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M_k
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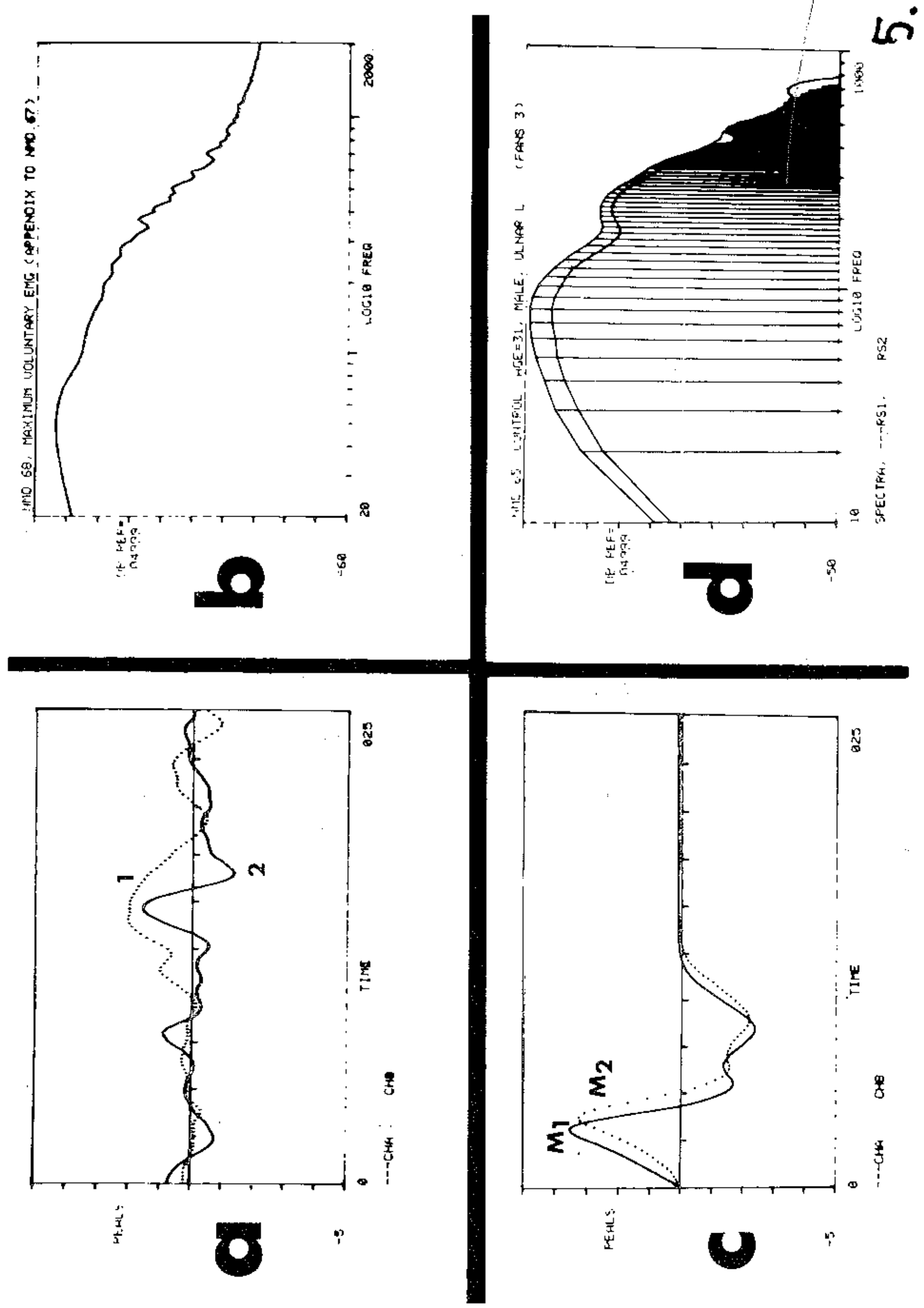
C A J F U M
d

4 c, d



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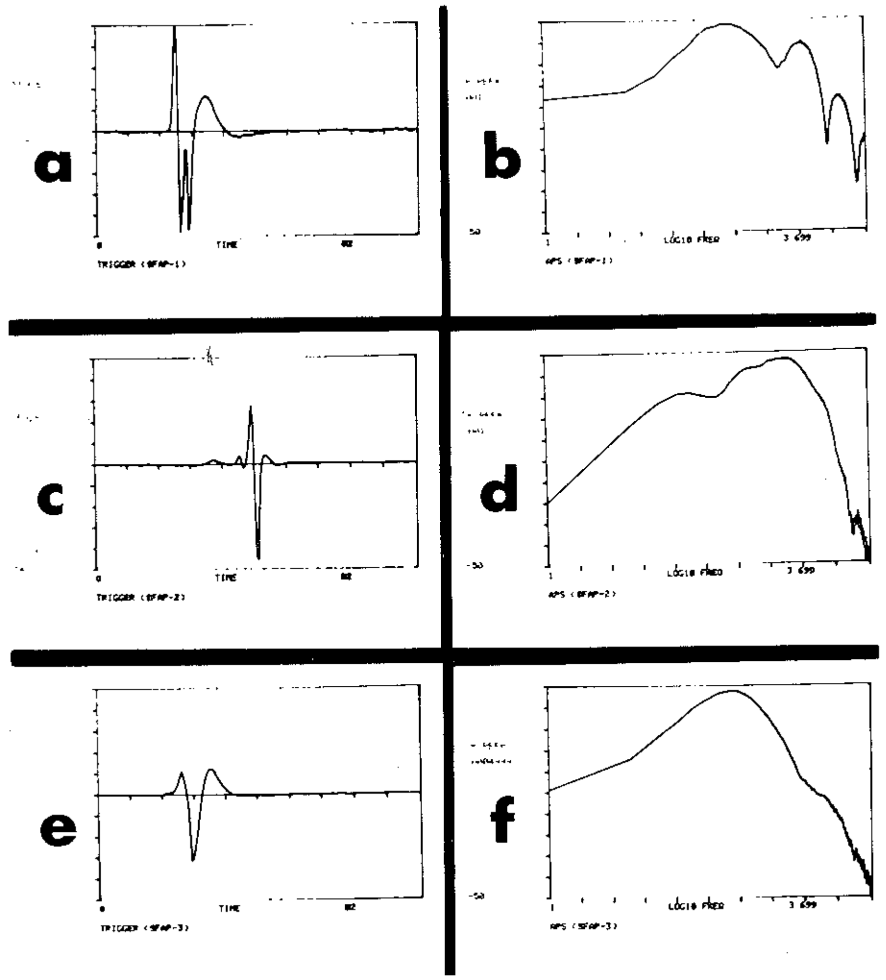

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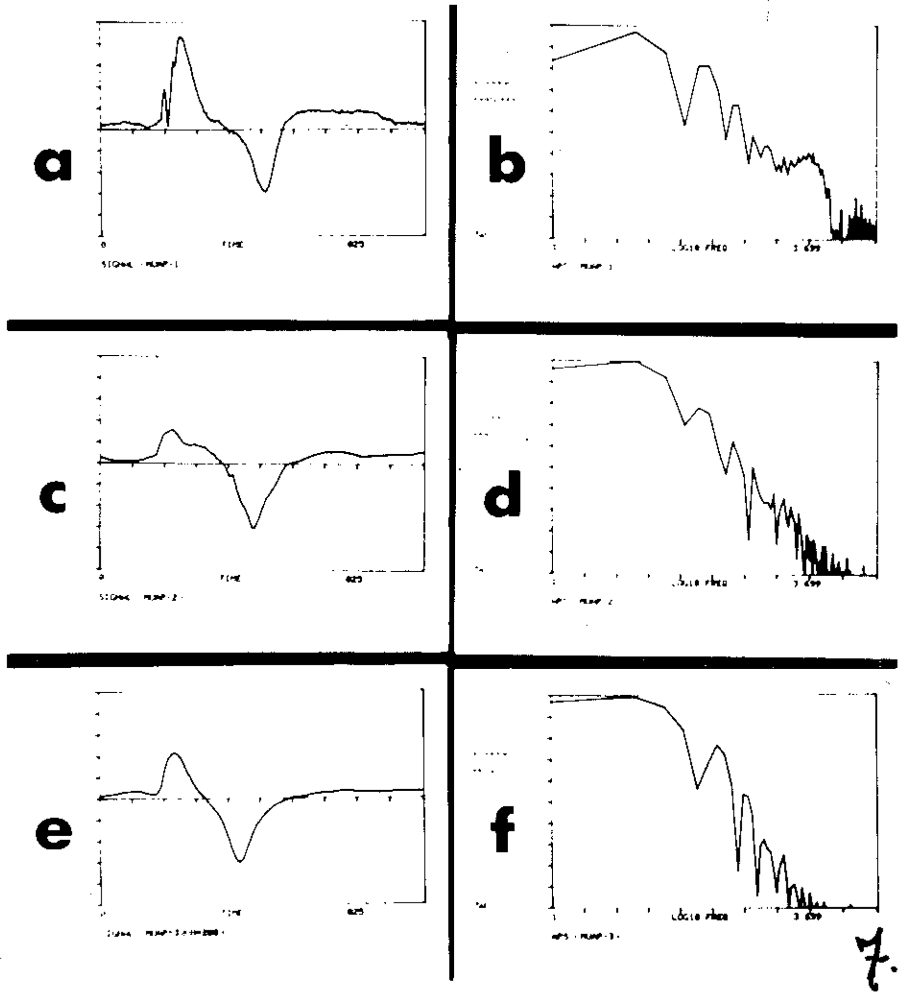
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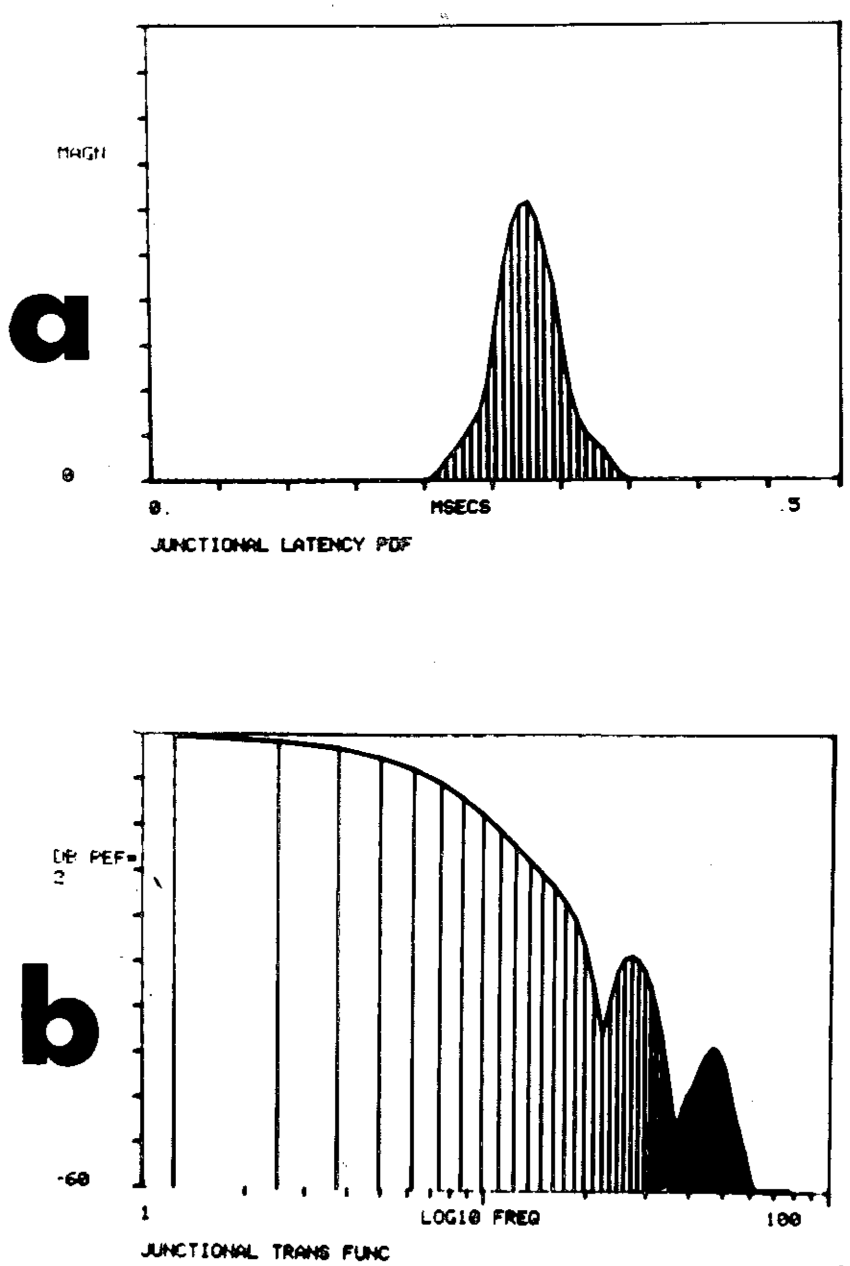
7.


a
MAGN
0
0.
MSECS
.5
JUNCTIONAL LATENCY PDF
b
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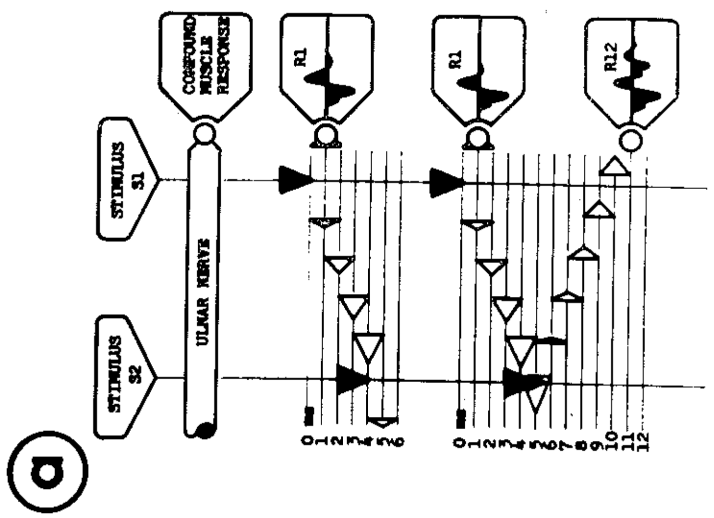
LOG10 FREQ
100
JUNCTIONAL TRANS FUNC

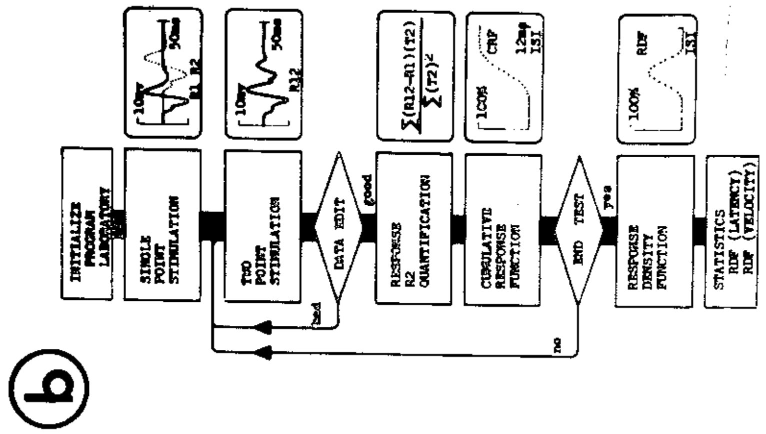
8.

a



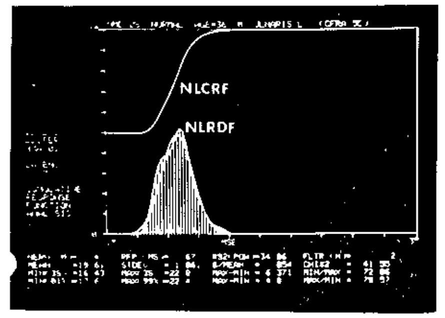
9.

b



10.

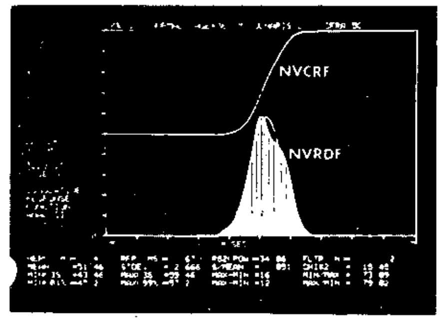
a



NLRDF (MSEC/M)

NERV (M)= .4	RFP.(MS)= .67	RS2(POW)=34.86	FLTR.(N)= 2
MEAN =19.61	STDEV. = 1.061	S/MEAN = .054	CHI*2 = 41.55
MIN(3S) =16.43	MAX(3S) =22.8	MAX-MIN = 6.371	MIN/MAX = 72.06
MIN(01%)=17.6	MAX(99%)=22.4	MAX-MIN = 4.8	MAX/MIN = 78.57

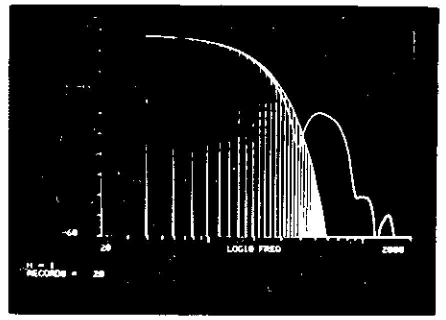
b



NVRDF (M/SEC)

NERV (M)= .4	RFP.(MS)= .67	RS2(POW)=34.86	FLTR.(N)= 2
MEAN =51.46	STDEV. = 2.666	S/MEAN = .051	CHI*2 = 15.45
MIN(3S) =43.46	MAX(3S) =59.46	MAX-MIN =16.	MIN/MAX = 73.09
MIN(01%)=45.2	MAX(99%)=57.2	MAX-MIN =12.	MAX/MIN = 79.02

c



11.

FIGURE CAPTIONS

1. Conceptualization of the motor unit (MU) as target organ of our studies. There are subunits which contribute to the performance and anatomy of the MU. The supra-unit is the integral function of a population of MUs as expressed in a whole muscle.

2a. Bioelectric signals associated with each stage in generation of the electromyogram. EPSP = excitatory post synaptic potential; IPSP = inhibitory post synaptic potential; threshold = the level at which a post synaptic action potential is initiated.

2b. Simplification of signal flow by substitution of Dirac Delta function for axonal and junctional potentials.

3a. Structural parameters of the peripheral motor system which we would like to know.

3b. Functional parameters of the peripheral motor system which are of interest in their own right, or needed to determine structural parameters.

3c. Levels of observation for study of the motor unit and generation of the EMG with parameters of the system which may be obtained.

4a,b. Schematic representation of the proposed EXPERIMENT. Electrode 1 is the electronics system ground reference. Electrode 2 is a surface reference electrode for differential amplification of the implanted monopolar wire macro electrode 3. Bipolar needle electrode 4 is for single muscle fiber recording.
Primary signals are the single fiber spike train and the summated EMG from electrodes 4 and 3. The MUAP is derived by averaging the macro electrode signal following each occurrence of a SFAP belonging to the same unit. Signal analyses are based upon linear systems theory and utilize the Fourier Transformation for auto and cross power spectrum estimation. Using the results from these analyses and assumptions regarding the signal generators, the peripheral motor system may be quantitatively simulated. Where reasonable confidence in the simulations has been established, the results of the simulation become predictive of actual system parameters. This is especially needed for estimation of unobservable structural parameters.

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4d. Symbology for the basic signal flow diagram defining generation of the electromyogram. Each letter represents the ensemble of time functions actually comprising that branch in the flow diagram. C = command inputs to the motoneuron; A = axonal conduction latencies; J = junction conduction latencies = terminal axonal conduction + neuromuscular junction delay + post synaptic activation delay; F = muscle fiber action potentials; U = motor unit action potentials = spatio-temporal summation of single muscle fiber action potentials; M = integrated whole muscle electromyogram = spatio-temporal summation of motor unit action potentials.

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5b. The auto power spectrum $M^{v}_{max}(jw)$, obtained by Fourier Transformation of 10 epochs of EMG as shown in 6a.

5c. An example of the M-responses, $M^{s}_{max}(t)$ obtained by stimulation at electrodes 5 and 6. In this figure both evoked potentials have been shifted to zero delay for visual comparison and for computation of their autopower spectra. Each waveform is the average of 10 evoked responses.

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THIRD INTERNATIONAL CONGRESS OF ELECTROPHYSIOLOGICAL KINESIOLOGY

ABSTRACTS OF COMMUNICATIONS

PAVIA - 30th AUGUST, 4th SEPTEMBER 1976

THIRD INTERNATIONAL CONGRESS
OF ELECTROPHYSIOLOGICAL KINESIOLOGY

ABSTRACTS OF COMMUNICATIONS

- SELECTED PAPERS -

(by alphabetic order)

PAVIA - 30th AUGUST, 4th SEPTEMBER 1976

WELCOME TO THE 3RD INTERNATIONAL ISEK CONGRESS.

In 1961 the University of Pavia had the privilege of offering hospitality to about one hundred neurologists and neurophysiologists coming from all Countries. Their scientific lectures, comunications and discussions were published as proceedings of the 1st International Congress of EMG; the EMG Pavia committee worked out the bases of Methodology and Thermi-nology in Electromyography; now the EMG committee is an important permanent staff of the Scientific Board of the International Society of EEG and Clinical Neurophysiology.

From that time, an enormous progress has been made. The insertion of EMG on the basic field of kinesiology represents a necessary clearer specification of the purpouse towards which the electrophysiological devices are to be applied in association with the biomechanical methodology. A similar purpouseful achievement has been realised with the association of electromyographical and morpho-hystometric, genetic and clinical investigations in the field of neuromuscular physiopathology.

Therefore EMG, if from some points of view tends to lose its primary autonomy, it has on the other hand greatly vitalized the study of both the normal branch of executive behaviour represented by kinesiology and the pathological one represented by the neuromuscular diseases.

These are the reasons why Electromyography, starting from the field of neurophysiology, was developed not only by neurologists, but also by the most engaged kinesiologists. Prof. Basmajian, Prof. Johnson, Prof. Ladd, Prof. Simard and many others became just after the 1st EMG Congress, the pioniers of a very fruitful and promising branch of the human Sciences: Electrophysiological Kinesiology. This Science expanded later also in bioengineering enriching its methodology with the contribution of the averaging thechniques and automatic statistical analysis: the muscles maintained « alive » by Basmajian were thus mimicked by very complicated and fine models. Among these scientists we were lucky to meet the Chief of the Neurophysiological Laboratorium of Göteborg Prof. Ingemar Petersen, the ISEK President.

Considering the close links that Kinesiologic Electromyography has developed between Neurology and Physical Medicine, we are glad that this ISEK 3RD International Meeting takes place just in an Orthopaedic Institut, that the Administration of the Hospital of Pavia and the Scientific Director

Prof. Mario Boni, have liberaly set at your disposition. While we are really satisfied with the promising and very efficient work going on in Electrokinesiology with the leading supervision of ISEK'S Scientific Committee, let us from this old University, to welcome all of you and wish you success in the reports and discussions that will engage us in these three days for the profit of many chronic patients; the opportunity will be also given to develop new ties of friendship between people coming from so many Instituts of different Countries in the world.

Pavia 31st August 1976

The General Secretary
Paolo Pinelli

The Local Secretary
Antonio Arrigo

INDEX OF AUTHORS *

* The names in capital letters are the first ones of each paper; coworkers are in small letters.

PRELIMINARY CLINICAL RESULTS OF UPPER EXTREMITY FUNCTIONAL ELECTRICAL STIMULATION APPLIED TO HEMIPLEGIC PATIENTS

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INTRODUCTION

Past work in the field of Functional Electrical Stimulation (F.E.S.) of the upper extremity of hemiplegic patients mainly described the orthotic value of this technique (Long et al. 1963, Rebersek et al. 1973, Merletti et al. 1975). It was the purpose of this work to carry on a clinical evaluation of upper extremity F.E.S. as an aid or as a partial substitute for traditional rehabilitation techniques for hemiplegic patients. As a result of such evaluation we expected to be able to define some basic criteria for patient selection and treatment.

MATERIALS AND METHODS

This preliminary evaluation was carried on in three clinical centers and involved 55 patients with rather wide ranges of age, time from lesion, spasticity and density of treatment (51 had etiology of cerebrovascular accident, 3 of trauma, 1 of intracranial tumor).
Stimulation was applied by means of cutaneous round or rectangular electrodes with respective surfaces of 7 and 12 cm^2 placed a) on the motor points of the muscles extensor digitorum communis and extensor carpi radialis (or ulnaris) or b) on a motor point of the triceps and on the radial nerve just above the elbow.
The electrodes had a sponge dampened with tap water to insure good electrical contact. The stimulator produced monophasic rectangular voltage pulses with 0.3 msec width, frequency of 30 Hz and amplitude adjustable between 0 and 90 volts.
The movements obtained were extension of the wrist and fingers with the electrodes in position a) and extension of the elbow and opening of the hand with the electrodes in position b). The stimulation, and therefore the movement, was controlled by a therapist or by the patient himself by slow changes of pulse amplitude obtained by means of a potentiometer. Each

evaluation consisted of 30 to 60 stimulated movements performed in 15 to 30 minutes.

The patients were treated for average of 26 times in 2.6 months (average of 10 times/month or once every three days for each patient).

At the end of the observation period the results were classified according to the following five classes:

W : (worse) worsening of either voluntary control or spasticity of any muscle of the arm or hand.

N : (no result) no change in spasticity and voluntary control.

P : (poor) improvement of the stimulated movement and/or of subjective feeling of the limb but no clear clinical improvement.

G : (good) clearly observable temporary (at least one day) decrease of spasticity and/or improvement of voluntary control.

E : (excellent) clearly observable permanent or long lasting (at least three days) decrease of spasticity and/or improvement of voluntary control.

No patient could be classified in the W class. Patients in group N or P were classified as «unsatisfactory» while patients in group G or E were classified as «satisfactory». The degree of spasticity of each patient was quantified according to the following four groups: 0) flaccid or with normal tonus; 1) tonus above normal but allowing a passive full movement of the joint; 2) tonus quite above normal allowing a large but not complete movement of the joint; 3) very high tonus allowing a very limited or no passive movement of the joint. While a full description of the cases, as well as some electrodiagnostic results, will be reported elsewhere, the distribution of patients in the above classes will be presented here.

RESULTS

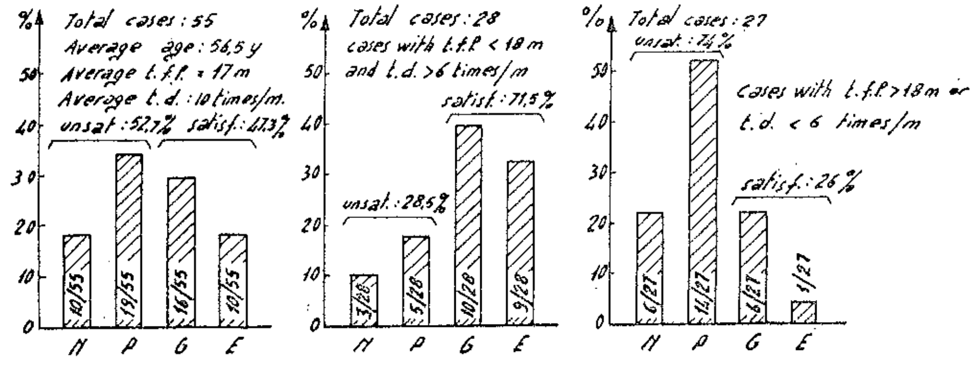
Fig. 1a shows the overall distribution of results but, because of the wide range of time from lesion (from 20 days to 10 years) and density of treatment (from 3 to 20 times/month) it is less interesting than Fig. 1b and 1c each representing half of the cases. The better results obtained from patients with lower time from lesion and more intense treatment are evident.

Fig. 2a shows the histogram of percentage of «satisfactory» cases as function of the time from lesion. Fig. 2b and 2c show the same percentage as function of the spasticity level and of the density of treatment.

These appear to be the three parameters with higher influence on the results.

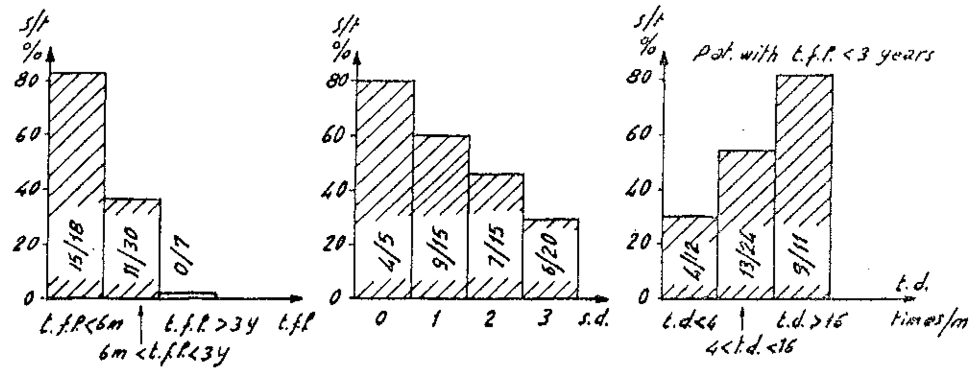
A better description of the correlation between parameters can be provided by three-dimensional plots; a simple one is reported in Fig. 3.

It appears to us quite important to report that over half of the patients classified in group P referred an increased psychological feeling of possession and control of the extremity and even if this was not objectively observable it may represent a first step leading to improvement to be obtained with further treatment. Beside this, a number of patients classified in class G or E had been previously treated with traditional techniques with poor success.



a) Total results. b) Results from pat. with lower t.f.l. and higher t.d. c) Results from pat. with higher t.f.l. and lower t.d.

Fig. 1 - Histograms of results. t.f.l. = time from lesion; t.d. = treatment density. See text for definition of the classes N, P, G, E.



a) Satisf./total res. as function of the t.f.l. b) satisf./total res. as function of the s.d. c) Satisf./total res. as function of t.d.

Fig. 2 - Histograms of satisfactory/total results as function of t.f.l. (time from lesion), s.d. (spasticity degree) and t.d. (treatment density).

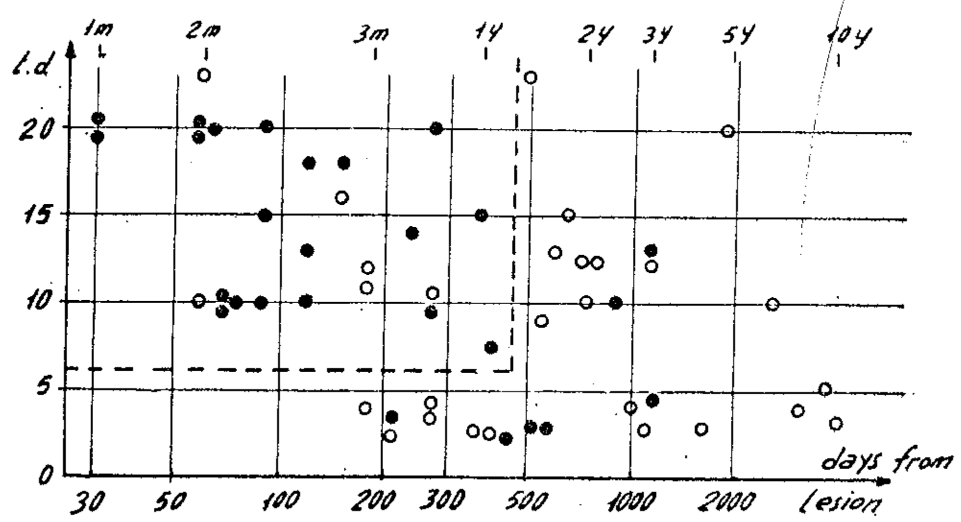


Fig. 3 - Plot of «satisfactori» (●) and «unsatisfactory» (○) results as functions of the treatment density (t.d.) in times/month and of the time from lesion (t.f.l.) in days. The percentage of satisfactory results within the dotted line is 77% while the overall percentage is 47%.

● = G+E
○ = N+P

This work also led to the conclusion that the stimulator, equipped with a shoulder transducer as described by Rebersek (1973) and Merletti (1975), could be used as an active orthotic device by about 8 to 10% of the patients, after proper training, at least for performing simple prehension tasks.

COCLUSIONS

The main conclusions from the above are the following:

a) on the basis of our experience, properly applied F.E.S. has no negative effect on hemiplagic patients and appears to be more important as a rehabilitation rather than as an orthotic technique.

b) clearly observable temporary or permanent improvement of muscle tonus and/or of voluntary control can be obtained in at least 70% of the patients treated more than 6 times/month for at least 2-3 months before the 18th month from lesion. This value rises to 80% for patients treated over twice a week and with time from lesion below 6 months.

c) the results are mainly conditioned by the time from lesion, the degree of spasticity and by the density of treatment. The patients should

be selected on the basis of these variables and treated at least twice a week.

The pocket size dimensions, the easy and foolproof operation of the latest devices make them suitable for use by the patient himself even outside the hospital personnel from routine clinical work. Our research on the upper extremity is continuing in order a) to evaluate this possibility, b) to study the effect of single, combined or sequential use of F.E.S. and traditional rehabilitation techniques, c) to define the optimal electrical parameters and conditions for maximal results, d) to investigate, on a larger number of cases, the significance of etiology.

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DECREMENT OF RECURRENT SINGLE MOTOR UNIT ACTION POTENTIALS. RECORDINGS FROM ISOLATED MUSCLES OF A NON-MYASTHENIC MAN

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A 21-year-old soldier was referred from a military installation for electromyography. He had been in excellent health except for long standing feeling of weakness in his right upper extremity which did not interfere with his duties as a clerk but limited his athletic activities.
Physical examination revealed a well developed, well nourished, young man in no distress. System review was unremarkable except for an absent right biceps reflex. Although no wasting was noted the right biceps brachii (musculocutaneous nerve C-5, C-6), the triceps brachii (radial nerve, C-6 to T-1), and the wrist extensors (radial nerve, C-6 to C-7) rated fair only, in motor strength. No sensory deficit was noted.

ELECTROMYOGRAPHY

In representative muscles of all four extremities, there was complete electrical silence during rest, no fibrillation or fasciculation potentials being recorded. During minimal volition discrete motor units of normal duration and amplitude were elicited in all muscles except the right biceps, triceps and the wrist extensors. With the exception of these muscles, normal recruitment occurred in all muscles and an interference pattern of normal density was noted on sustained voluntary contraction.
Marked, rapid decrement of amplitude could be observed after minimal volition in the right biceps, right triceps, and the right wrist extensors. The pattern of decay consisted of one or two motor unit potentials of normal amplitude followed by a rapid fall-off and even complete abolition of subsequent action potentials. After two minutes of rest, recovery of amplitude occurred, only to deteriorate after a few voluntary contractions.
Repetitive stimulation of right brachial plexus at Erb's point as recommended by Gassel (1964) with recording electrodes placed in the biceps, triceps, and wrist extensors, respectively, resulted in rapid decay of amplitude of the evoked potentials. Supramaximal stimuli of 0.3 msec. duration at frequencies of 3-10 Hz were employed. Recovery of amplitude occurred after two minutes rest, and decay was reproduced by renewed repetitive stimulation. Ten mg Tensilon^{R} (edrophonium chloride) was given intravenously, but no significant reversal of the decay in amplitude was observed.

In order to exclude myasthenia gravis, short subcutaneous electrodes were placed in the abductor pollicis brevis and abductor digiti quinti bilaterally and the medial and ulnar nerves were repetitively stimulated supramaximally up to 50 Hz without any evidence of abnormal muscle fatigability. Finally, the facial nerve fatigue test with recording electrodes placed in the orbicularis oculi was performed bilaterally with repetitive shocks up to 25 Hz, again without fall-off in amplitude of the evoked potentials.
The patient was seen three months after his initial visit.
Electromyography including above procedures were repeated and identical results were reproduced.

COMMENT

There is definite evidence of oscillating amplitude of motor unit action potentials in three isolated muscle groups of the right upper extremity of an apparently healthy, young man. Decrement was rapid and eventually to the point of complete abolition of the recurrent potentials. This result could be achieved both by minimal volition and repetitive faradization of the brachial plexus at Erb's point. Tensilon^{R} test was negative.
Repetitive stimulation up to 50 Hz of median, ulnar, and facial nerve failed to elicit a pathological fall-off of evoked action potentials recorded from the abductor pollicis brevis, abductor digiti quinti and the orbicularis oculi, a muscle involved early in myasthenia gravis. There was no facilitation or post-tetanic exhaustion, such as described by Desmedt (1957, 1966).
We have to assume that we deal in this case with a significant defect of neuromuscular transmission selectively limited to isolated lower motor pathways which have been damaged in the past.
Myasthenic-like states affecting lower motor neurons have been suggested by others, Simpson (1964). Buchthal and Høncke (1943) and later Pinelli and Buchthal (1951) observed unusual muscular fatigability of motor units manifested by rapid decrease in amplitudes during a sustained contraction as early as 10 days after onset of paresis. A similar fatigue of motor units was described by Kugelberg and Taverner (1950) in syringomyelia. Eaton and Lambert (1957) and Lambert, Rooke, Eaton and Hodgson (1961) reported their classical myasthenic syndrome associated with remote malignancies. Mulder Lambert and Eaton (1959) described several cases of amyotrophic lateral sclerosis. Baginsky (1967) reported a case of peripheral neuropathy displaying myasthenic EMG patterns.
The unchanging height of amplitudes of evoked action potentials by repetitive stimulation of nerves supplying muscles usually affected by myasthenia gravis precludes this systemic disease in our patient. We are dealing

instead with a selective neuronal affliction. His age and the steady status of his general health does not favor the diagnosis of amyotrophic lateral sclerosis, however, he apparently suffered damage to some of his anterior horn cells; most probably by poliomyelitis.

Hebb and Waites (1956) and McIntosh (1959) have suggested that acetylcholine and the enzymes acetylase and acetylcholinesterase are synthesized in the nerve cell and carried to the nerve endings via the axon. This would also explain the lack of adequate response to Tensilon[R] since too few quanta (if any) of acetylcholine and too few enzymes acting upon them are present at the myoneural junction.

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FUNCTIONAL TESTS OF A NEW PROPORTIONALLY CONTROLLED STIMULATOR IN HEMIPLEGIC PATIENTS

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INTRODUCTION

The Functional Electrical Stimulation (FES) of paralyzed extremities, as it is known, has been utilized usually for orthotic and rehabilitation purposes in hemiplegic patients with paralyzed extremities and damages of motor organization due to a variety of causes, such as cerebral stroke, injury, inflammation, tumour, etc. Remembering that the neurophysiological mechanism of the muscle contraction and muscle tone are the result of processes within the CNS, the FES is utilized to engage proprioceptive and musculocutaneous reflex mechanism which may have a strong effect on the organization at different levels of the neurophysiological systems.

The method in the present study has consisted of controlled stimulation applied to the paralyzed extremities for evoking the lost movements, utilizing also the residual voluntary capabilities of the patient. Moreover, some long duration effects of FES improve the reorganization of the basic neural mechanisms involving the movements. In using the FES as an orthotic device it is important to have the stimulation voluntarily controlled by the patient for realizing the desired movements.

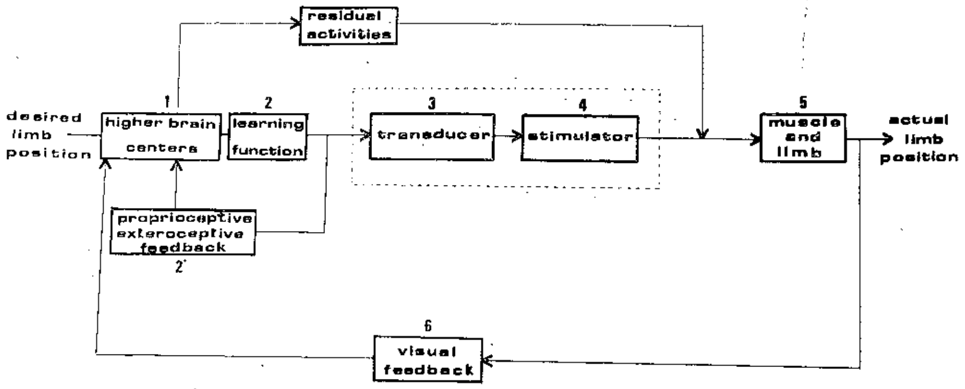
The block diagram of the orthotic methodology is reported in Fig. 1.

A series of patients suffering from hemiplegic conditions of traumatic or vascular origin have been submitted to functional electrical stimulation with a new kind of two-channel proportionally controlled stimulator. The main characteristic of this device is the transducer for controlling the stimulation, which results in the movement in relation to the residual volitional ability.

In principle, the transducer is a «Touch-Time to Amplitude Converter», which changes the duration of the time of touch in amplitude variation of the signal controlling the stimulation.

For correct use of the device a period of learning is required by the patient (block 2-6 of Fig. 1).

The presentation will describe the methodology used for evaluating the orthotic use of the stimulator on selected patients, and for determining the practical influence of the learning function on the possibility of restoring the impaired movement. A great deal of equipment has been developed for measuring biomechanical phenomena, in terms of the patients treated with FES, and this will be described as well.



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Facolta' di Ingegneria
1976

COMPARISON BETWEEN TRADITIONAL NEEDLE ELECTRODE EMG AND SURFACE ELECTRODE EMG SPECTRAL ANALYSIS

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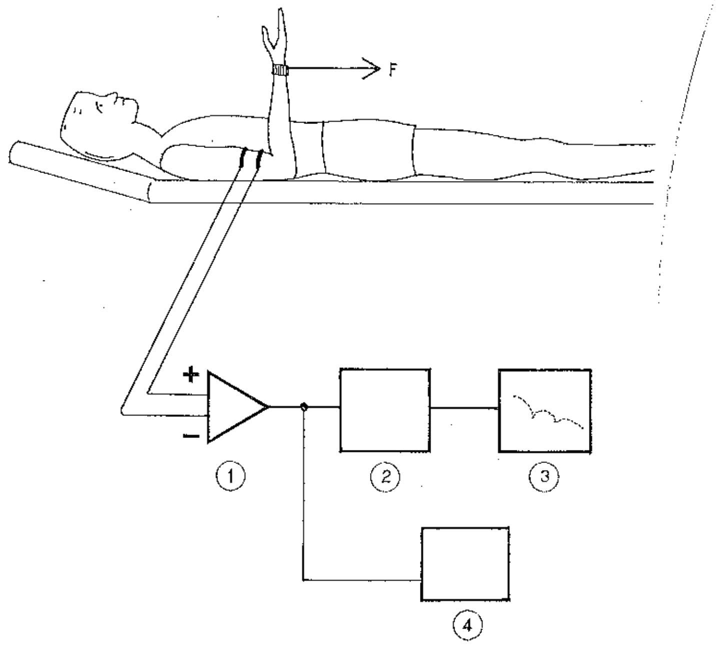
A method is proposed to characterize the electrical activity of the muscle. This method was based on the assumption, theoretically proved in an extremely simplified model, that the shape of surface electrode EMG spectrum should provide information about both «average» duration of the motor unit action potential (MUAP) and about «average» propagation velocity of the spike along the muscle fibers.

The theoretical model has shown that zero values of the single impulse spectrum are maintained in the spectrum of artificially produced surface electrode EMG. It was expected therefore that «dips» in the single MUAP spectrum should be maintained in surface electrode EMG spectra. To test this theory an analog frequency analyzer has been especially designed for use on-line. Fig. 1 shows the experimental procedure: the surface electrode EMG signal after amplification in a differential amplifier was fed into the frequency analyzer. The output was visualized on a memory oscilloscope.

The recurring shape of the spectra obtained for normal subjects has been shown in Fig. 2. A remarkable similarity can be seen with the shape of the single motor unit spectrum (surface electrodes) observed by Kadefors, Petersén & Broman (1973).

A few healthy young men volunteered as control subjects. The activity of the biceps brachii muscle under isometric contraction was measured both with needle electrode EMG and with surface electrode EMG. The surface electrode EMG was recorded during a maximal voluntary contraction, and was subjected to frequency analysis. A comparison was made between the results provided by the traditional metrod using Adrian concentric needle electrode and the ones obtained with the frequency analyzer. A good agreement was found as for the MUAP duration. The analysis has been extended to a few patients with both neurogenic muscle disorders and with myogenic disorders.

The results obtained with the patients will be presented and discussed.



(1) DIFFERENTIAL AMPLIFIER
(2) FREQUENCY ANALIZER
(3) MEMORY OSCILLOSCOPE
(4) MAGNETIC RECORDER

Fig. 1

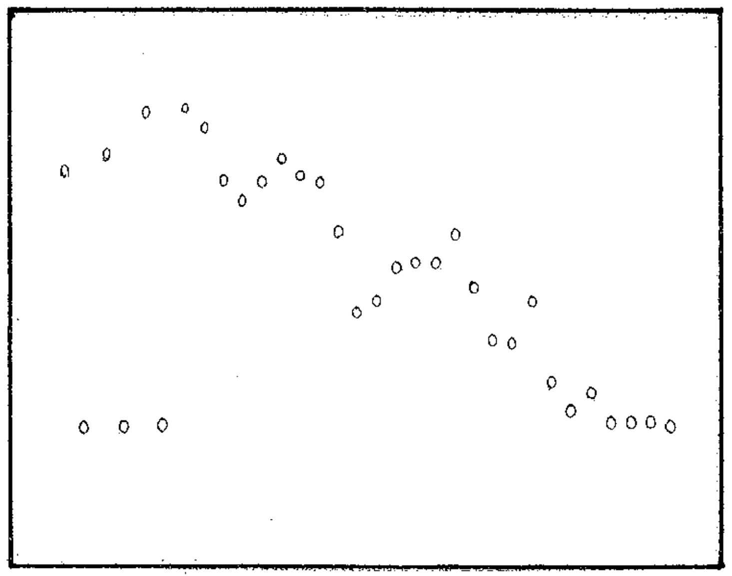


Fig. 2

EVALUATION OF GAIT IMPAIRMENTS BY MEANS OF VECTOR DIAGRAMS

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The deep analysis of the locomotory apparatus and the rigorous evaluation of the functional recovery constitute the fundamental basis of every correct rehabilitative programme. Along this line, the aim of this paper is to present a procedure for the evaluation of the human gait based on a suitable elaboration and representation of the ground reactions.

As known, during the stance phase the foot exerts a pression on the ground. The resultant force of such pression is named ground reaction. It depends on the dynamics of the whole body and represents a highly significant syntesis of the movement, provide that its information content assumes a form which allows its easy reading.

The representation chosen is a vector diagram where each vector represents the projection on the sagittal plane of the ground reaction with its magnitude, inclination and point of application in istants of time uniformly discretized.

The equipment used is a force platform (Kistler type 9261 A) with four piezoelectric transducers which, through suitable charge amplifiers, provide the voltages proportional to the force components acting on them. A special hybrid computer, made for this purpose, elaborates the above signals and forms on a persistent image display the vector diagram.

The subjects walked at a constant speed across a 17 m long track. The measure is taken at two-thirds of the way so that a steady-state deambulation is reached. In this way, it is possible to obtain on the display and to photograph the vector representing the ground reaction in sampled instants of time respecting the actual geometric relationship with the platform (amplitude, angle of slope and point of application).

In order to point out the validity of this procedure, many tests have been performed on many subjects both normal and pathological with different kinds of impairements to the locomotory apparatus.

The conclusion presented here also take in account the results obtained in a previous work by Boccardi et al. 1976) and by analogous analysis performed by Cappozzo et al. (1973) with a different procedure on normal subjects.

In Fig. 1a, b the pypical vector diagrams of a normal subject are shown. The pattern of a) refers to the right foot while the subject walked (from the left to the right) at 88 steps/min.

The pattern b) refers to the left foot at the same speed.
From these diagrams, which can be assumed as typical of normal subjects although some secondary individual characteristics arise and make this vector diagram a kind of identity of each subject, the following conclusions can be drawn:

a) High repeatibility of the diagrams obtained at different times from the same subject at the same speed. (This has always been verified even in the pathological cases).

b) A perfect similarity of diagrams obtained from the right and the left foot.

c) The envelope of the vectors presents a characteristic course without sudden variations with the shape of a half butterfly.

d) Except for the first few (heel-strike), the inclination of the vectors changes with continuity from the backward to forward.

e) The point of application of the vectors moves monotonically in the direction of progression and shows an evident plateau in the last of stance phase which causes a correspondent closeness of vectors.

The experiments performed on subjects presenting impairments of the deambulation following various pathological factors show that the reproducibility of the proof is more than satisfactory and, most importantly, the diagrams obtained are highly significant and clearly different from those concerning the normal walking. The variations observed allow for a valid interpretation both from a mechanical and kinesiological point of view.
In order to verify the existence of typical patterns, characteristic of various impairments of the locomotory apparatus, the following examples are shown.
The Fig. 2a, b, c, d, refer four subjects presenting after effects of poliomyelitis concerning the lower limbs, particularly one of the two where they wear a long orthesis. The diagrams refer to the stance of the orthesized limb. The shapes are completely different from those of normal subjects and very similar between them.
In particular the reduction of the reaction base (the distance between the points of application of the first and last vector) is evident.
The closeness of most of the vectors in the first part of the stance phase is a significant determinant. In fact the ground reaction changes from backward to forward following the rotation of the long stiff orthesis articulated in corrispondence of the shoe heel. Moreover the first vectors inclinated in the direction of motion, typical of the normal cases, are lacking and backward displacement of the application point in corrispondence of the closeness clearly appears.

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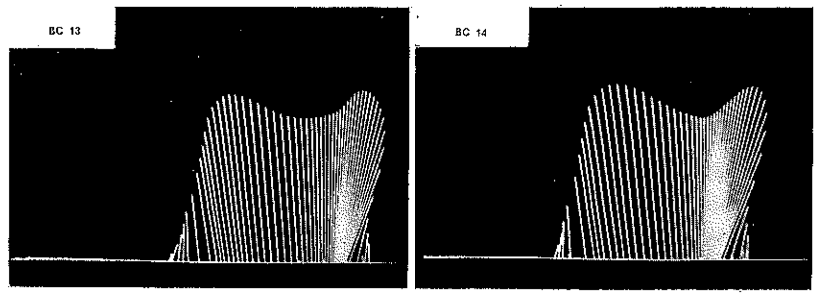
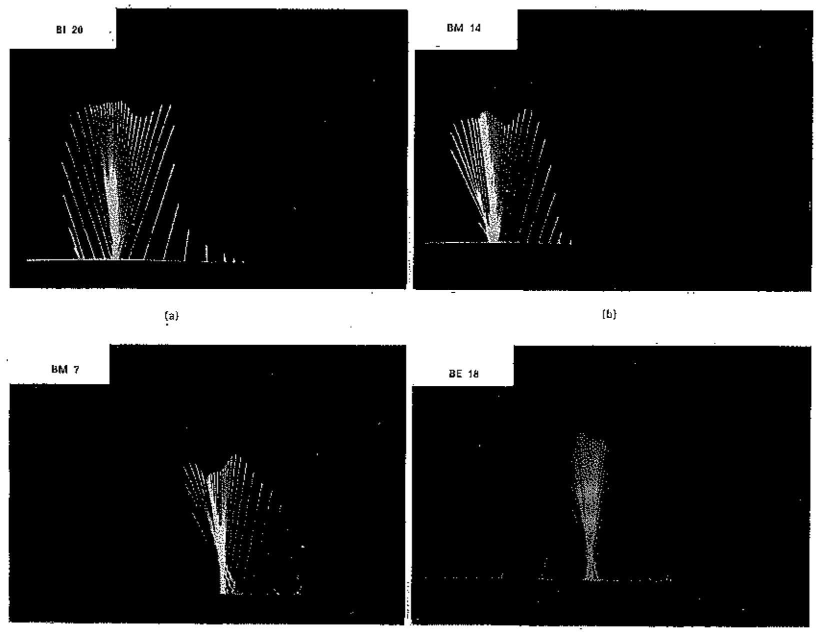


(a) Fig. 1 (b)



(a) (b)

(c) Fig. 2 (d)

Fig. 1 - Vector diagrams of a normal subject (R.R.) walking at 88 steps/min a) right foot, b) left foot. One vector every 15 msec

Fig. 2 - Vector diagrams of the orthesized limb of four different subjects presenting after of poliomyelitis (A.G., C.M., P.T., M.G. respectively). One vector every 27 msec

THE MOTOR UNIT ACTION POTENTIAL: STUDY WITH A STOCHASTIC SIMULATION MODEL AND DEVELOPMENT OF A NEW MECHANO-ELECTRICAL NEEDLE TRANSDUCER

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INTRODUCTION

A stochastic simulation model is proposed in which the motor unit action potential (m.u.a.p.) is considered as a summation of separate muscle fibre potentials. The spatial position and the diameter of the fibres and the dispersion (mainly determined by the situation of the motor end-plates) can be chosen according to certain probability density functions (p.d.f.) or in a deterministic way.

Furthemore the m.u.a.p. can be simulated as it will be recorded by a point electrode or a not-point-shaped electrode. With a not-point-shaped electrode the potential contribution of a certain fibre is calculated numerically by spacial integration with respect to the electrode surface.
This stochastic simulation model gives a more realistic representation of the generation of a m.u.a.p. than already described models. For instance the model of George (1970) in which the fibres are supposed to lie in a very regular way or the mathematical model of Ekstedt and Stålberg (1973) that only describes the action potential of *one* muscle fibre with a «not-point-shaped» electrode. With the model questions about the selectivity of different electrodes can be examined. Also diagnostical problems concerning the shape of a m.u.a.p. and the possible relation to pathological origins can be studied. For instance to what extent a polyphasic pattern may occur due to the loss of muscle fibres belonging to a single motor unit.

One of the reasons for constructing this model was to examine the possible relation between certain parameters of a m.u.a.p. and the motor unit type (e.g. slow, intermediate or fast). Therefore it is necessary that besides information about the m.u.a.p. information is also available concerning the type of motor unit that is examined.

Although there are several ways to fulfil this requirement it is interesting to start a development of a new type of needle electrode by which both electrical and mechanical activity *at the point* of the needle can be registrated *at the same time*. Although this project is still in progress some preliminary results of a first prototype will be given.

STOCHASTIC SIMULATION MODEL FOR A M.U.A.P.

The potential contribution of each muscle fibre with coordinates (x, y) to a point P (see figure 1) with coordinates (x_i, y_i, z_i) of the electrode is calculated according to the tripole model of Rosenfalck (1969) and the assumption that the amplitude increases proportionally to the square of the fibre circumference (Lorente de Nó, 1947; Håkansson, 1957).

$$\phi_P(r,z) = \frac{a^2 . I}{4\pi . \sigma} \{\alpha . \overbrace{\frac{1}{\sqrt{(z-s_1-z_i)^2+r^2}}}^{\text{pole 1}} - \frac{1}{\sqrt{(z+s_1-z_i)^2+r^2}} + \frac{1-\alpha}{\sqrt{(z+s_2-z_i)^2+r^2}}\} \qquad (1)$$

where : a represents the fibre diameter; a is chosen from a p.d.f.
r represents the distance from point P to a muscle fibre
$r = \sqrt{(x-x_i)^2+(y-y_i)^2}$; x and y are chosen from a p.d.f.
z represents the position of the tripole on the muscle fibre
$z = z_o+d-v.t$; the dispersion d is chosen from a p.d.f.
v represents the velocity of the action potential and t is time.
I, σ, α, s_1, s_2, z_o are constants.

If an electrode is not point-shaped the final contribution of each muscle fibre to the m.u.a.p. is for each point of time calculated by spatial integration with respect to the electrode surface A.

$$\phi_{el} = \frac{1}{A} \iint_A \phi_P\,(r,z)\;dA \qquad (2)$$

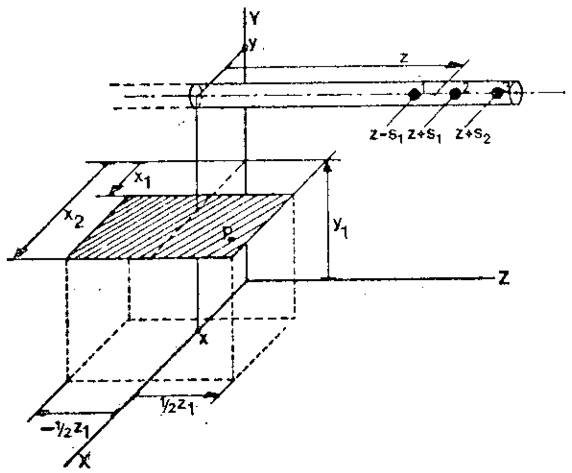

Fig. 1

To solve this equation the electrode surface was approximated by a rectangle, parallel to the muscle fibres. In figure 1 the position of the electrode and one muscle fibre is given in order to indicate the coordinates used.
The solution of the surface integral for pole 1 (see formula 1) is:

$$\phi_{el;pole\ 1} = \frac{1}{A} \iint_A \frac{1}{\sqrt{(z-s_1-z_i)^2+r^2}}\, dA \approx$$

$$\frac{1}{(x_2-x_1)} \cdot \frac{1}{N} \sum_{k=1}^{N} \ln \left| \frac{x_2-x+\sqrt{(x_2-x)^2+(y-y_1)^2+\{z-s_1-(-\frac{1}{2}z_1+k.\frac{z_1}{N})\}^2}}{x_1-x+\sqrt{(x_1-x)^2+(y-y_1)^2+\{z-s_1-(-\frac{1}{2}z_1+k.\frac{z_1}{N})\}^2}} \right| = g(-s_1)$$

thus $$\phi_{el} = \frac{a^2 \cdot l}{4\pi\sigma} \{\alpha \cdot g(-s_1) - g(s_1) + (1-\alpha) \cdot g(s_2)\}$$

In the model time t is increased with fixed steps (in fact $z = z(t)$ will decrease with fixed steps) and for each point of time, formula (3) will be calculated; the result is the contribution to the m.u.a.p. of one muscle fibre.

The contributions of the individual fibres are calculated successively and summated afterwards. For each calculation the characteristic values of the muscle fibre a, d, x, y are taken from p.d.f.'s. The model is programmed in Fortran.

Results can be given either in the time domain or by Fourier transforming in the frequency domain.

Figure 2 gives some simulated m.u.a.p.'s obtained with the model. The fibre configuration, the used parameters and three different electrode positions are indicated. In each position the m.u.a.p. is «measured» with a point-shaped (——) and a not-point-shaped electrode (– – –).
From these (and other) results the following can be concluded:

1. Within a motor unit: a not-point-shaped electrode gives a signal that is related more to many fibres of the m.u., whereas the point-shaped electrode gives a signal that is almost determined by the closest fibres.
2. At a distance of about 0.5 mm (C): there is hardly any difference between the signals of the two electrodes.

Used parameters:

a	:	30-50 μ	random
d	:	0-2 mm	random
x	:	0-1.5 mm	random
y	:	0-3 mm	random
x_2-x_1	:	0.3 mm	
z_1	:	0.6 mm	
y_1	:	0.85; 0.4; —0.5 mm	
v	:	4 m/sec	
α	:	0.8	
s_1	:	0.05 mm	
s_2	:	0.25 mm	
z_o	:	4 mm	
l/σ	:	arbitrary constant.	

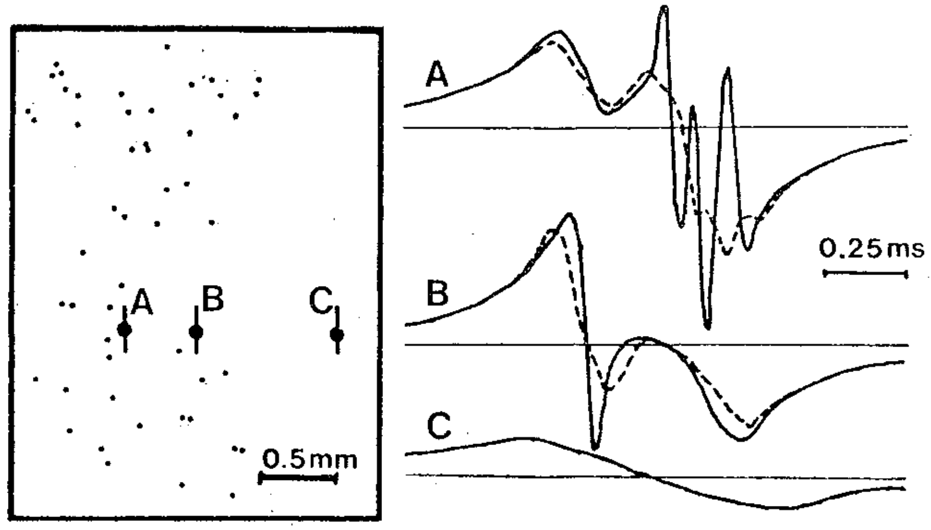

Fig. 2

MECHANO-ELECTRICAL NEEDLE TRANSDUCER

As already mentioned in the introduction it was interesting to develop a new kind of electrode in order to obtain specific information concerning the motor unit type. The usual techniques to obtain this information such as working with a homogeneous muscle or using the recruitment principle described by Henneman et al. (1965) or using a specific averaging technique described by Stein et al. (1972) are not applicable in a number of cases. For instance at studies of human back muscles.
A prototype of a mechano-electrical needle electrode is shown in figure 3a. It consists of a glass coated inner needle within another stainless needle.

The inner needle is suspended to an elastic membrane and can move rather freely. On top of the inner needle, inside a holder, a mirror is monuted. The distance between the mirror and a bundle of fibre optics is measured by means of reflected light by a so called Fotonic Sensor (type KD-100, Mechanical Technology Incorp.). In this way movements of microns (from 0-40 K c/s) can be registrated. The tip of the inner needle is not glass coated so it can be used as a normal EMG electrode. Figure 3b gives a registration of a double pulse. The tip of the inner electrode was just inside the muscle (musculus soleus of the rat). The elasticity of the membrane and the dry friction between inner and outer needle during deep intramuscular registrations are still factors to be improved. In figure 3b the intramuscular displacement during an isometric contraction can be observed. Also the force is given for a comparison between displacement and force.

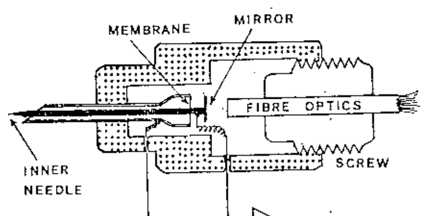

Fig. 3a

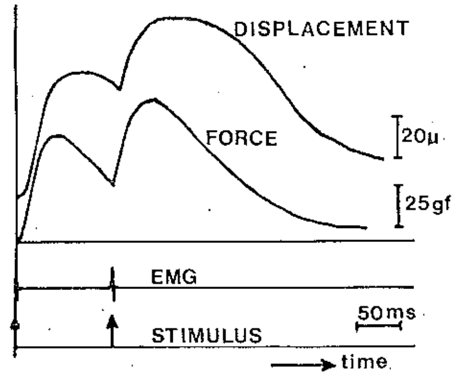

Fig. 3b

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THE EMG, AS A MEASURE OF THE ISOMETRIC FORCE OF INDIVIDUAL MUSCLES

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It is well known that in the intact organism the striated muscles are usually arranged in antagonistic pairs in relation to the articulation. In the simplest cases the movement can thus result from the action of a single group of agonist muscles. However, even in this case the force, or more precisely, the external torque is the sum of the torques exerted by the various muscles. Now in the intact human organism it remains practically impossible to measure directly the force exerted by each muscle, in spite of the prospects opened by the technique of Buchthal and Schmalbruch (1970). As for estimating this force by calculus, it requires, at least in the case of submaximal contraction, resorting to various hypotheses, some of which have not yet been verified, as is brought out in the analysis of Cnockaert et al. (1975). Thus the only possibility is presently to estimate the force exerted by an individual muscle from the EMG of this muscle. This procedure is generally adopted and is implicitly supposed that the EMG is the measure of the force. However, as far as we know, such an hypothesis has not been really confirmed. The present paper deals with this problem in the case of isometric contractions of elbow flexors.

TECHNIQUE AND PROCEDURE

The myoelectrical activities of the principal flexor muscles of the elbow were recorded *simultaneously*. The global myoelectrical activity of the biceps brachii was led off by means of two silver electrodes fastened to the skin. The global electrical activity of each of the other flexors (branchioradialis, brachialis, and pronator teres) was led off by means of wire electrodes. Bipolar platinum electrodes, similar to those previously described by Maton et al., [1969], were used. The records obtained were of the interferential type and thus expressed the global activity of the muscle.
The different EMG's were recorded on magnetic tape and then integrated by an analog-digital converter (Feuer, 1967). The integrated EMG was given in the form of pulses whose number was proportional to the area of the EMG.
The subjects were seated, with the right forearm and arm in the same horizontal plane. The forearm was fastened to a splint. The latter was

part of a mechanical system rotating about a vertical axis. A strain gauge, which was joined to the non-movable part of the apparatus, was attached at a point 25,5 cm from the axis of rotation. The signal of the force was both recorded on magnetic tape and displayed on an oscilloscope.
The tests involved exerting for 30 seconds by a constant force against the strain gauge in order to keep the spot of the oscilloscope on given marks on the screen. Six different levels of force whose value depend upon the maximum voluntary force of the subject, were successively maintained. An adequate rest period was provided between each exertion in order to avoid fatigue. The tests were carried out with the hand in a full supine position.

RESULTS AND DISCUSSION

The results were obtained from a group of 12 men and women between the ages of 15 and 59. They were either normal or without disorders of a neurological origin related to the neuro-muscular region under study. Each subject was examined once.
Whichever muscle is considered, a quadratic relation has been demonstrated between the integrated electromyogram and the force. These relations are lowly scattered, but their slope varies from one subject to another. In order to determine more precisely the form of the relations between the integrated EMG and the force, the values obtained for each of the muscles examined, for all of the subjects in all of the examinations, have been brought together. To do this the integrated EMG has been expressed as a percentage of the value (11 kg) corresponding to a single level of force maintained by all the subjects: thus neither the conditions affecting the reception of the signal nor the differences in maximum force among the individuals are taken into account. The activation level of each of the flexors, evaluated with the integrated EMG varies with the force according to a relation which, at first approximation, takes the form of $Q = aF^2 + bF + c$, where c depends only upon the integration threshold and can be considered negligible. It is, in fact, for the adyustment to this quadratic function that the correlation coefficients (r of Bravais-Pearson) are the highest (.91 for the biceps brachii, .92 for the brachialis, .94 for the brachioradialis, and .97 for the pronator teres). All of these values are of course very significant at the threshold of .01.
Because of this, it becomes possible to group on the same graph all the results obtained for the different muscles. It can be observed in Figure 1 that the points are distributed along one and the same curve. The relation between Q and F is expressed by $Q = 0{,}4\,F^2 + 9{,}9\,F + 17$.

This common relation express that the integrated EMG of each elbow flexor presents the same relationship with the external torque. Reciprocally, it can be said that the external torque is the sum of similar functions of individual integrated EMG's.
Consequently, taking into consideration that a) the external torque is the sum of the individual torques exerted by each muscle, b) the EMG of a muscle is a reliable index of the level of exitation of this muscle and of it alone (Maton et al., 1969; Bouisset and Maton, 1972), it follows that *each value of the integrated EMG of a muscle will give a value of the force exerted by this muscle, through a quadratic relationship.* This may be considered as generally valid, at least in the case of isometric contractions. It should be emphasized, however, that it is not a question of measuring the absolute value of the force and that, moreover, if the significance of the surface EMG changes (through a variation in length or through muscle fatigue) the measure of the force varies.

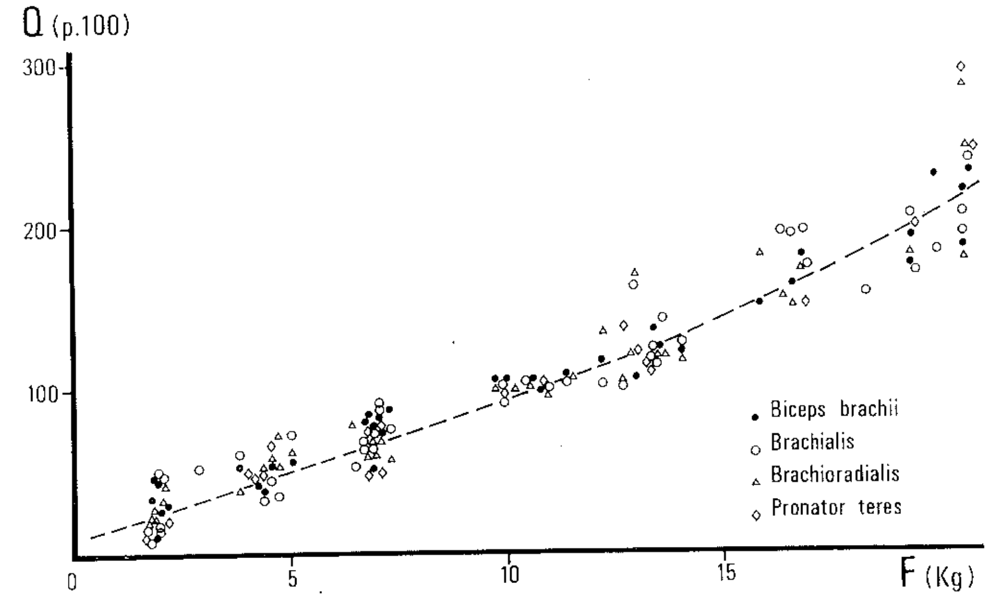

Fig. 1

Fig. 1 - Relation between the total integrated myoelectrical activity of the principal elbow flexors and the force.
Q: integrated electromyogram as a percentage of the value corresponding to a load of 11 Kg, the load maintained by all the subjects.
F: force measured at 25,5 cm from the axis of rotation of the elbow.
Note: For the sake of clarity the results of tests on only five subjects are shown.

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DETERMINANTS OF MYO-ELECTRIC POWER SPECTRA. PART II: EXPERIMENTAL VERIFICATION

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(see pag. 96)

STANDING WITH CANES AND OTHER WALKING AIDS AN E.M.C.G. STUDY I

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2 healthy young men and 1 woman have been studied in standing and walking with different types of canes and crutches. A standard program of EMG registration by the telemetry equipment of SFENA in Paris, has been used for each of them (Tables 1 and 2). The electric activity of each muscle mentioned in table 1 was registered and the EMG curve integrated and compared with the activity in each posture or type of gait listed on table 2. A photographic control has been made for each described posture. The first following conclusions can be mentioned:

1. The EMG of the «static» muscles of the lower limbs is not different in standing with normal weightbearing of both legs with and without canes.
2. The use of 1 cane reduces the electrical activity of the shoulder, trunk or leg muscles in monopodal weightbearing, in standing on tiptoes, on heels or in instable postures and bending forward.
3. This reduction of muscles electrical activity is much more important with 2 canes.
4. The best protection of the muscles activity of the lower limbs is assured by using the cane at the controlateral side (one cane left for helping the right leg).
5. The protective effect for the muscles is better with 2 short canes than with a too long one.
6. The most used muscle of the shoulder in using canes are, according to the importance of their electric activity: triceps brachii, posterior third of the deltoïdeus, trapezius and, muscles less important: latissimus dorsi, pectoralis major and biceps brachii. The deltoïdeus controls specially well the rotation movement of the cane.

The next part of this study is the use of the canes in walking and with the same research program in different types of limping.

Table No. 1

Recorded muscles

M. trapezius
M. deltoïdeus (posterior part)
M. deltoïdeus (anterior part)
M. latissimus dorsi
M. biceps brachii

M. triceps brachii
M. pectoralis major
M. rectus abdominalis
M. sacrospinalis
M. gluteus medius
M. gluteus maximus
M. rectus femoris
M. biceps cruris
M. tibialis anterior
M. triceps surae
M. peroneus longus.

Table No. 2

The various postures in which the muscle activity was recorded

1. Standing on both legs
2. Standing on the right leg
3. Standing on the left leg
4. Standing on the forefoot
5. Standing on the heels
6. Bending forward
7. Bendinb backward
8. Control in standing in normal free posture.

Table No. 3

Aids

One cane on the right side
One cane on the left side
Elbow crutches
Elbow crutches of exagerated length
Elbow crutches of insufficient length
Crutches
Frame
Tricycle.

HUMAN MUSCLE SPINDLE ACTIVITY DURING THE TONIC VIBRATION REFLEX: THE ABSENCE OF DEMONSTRABLE ALPHA-GAMMA CO-ACTIVATION AND ITS SIGNIFICANCE FOR THE TONIC STRETCH REFLEX

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In alert human subjects the activity of single muscle spindle afferent fibres may be recorded with microelectrodes inserted through the skin into appropriate muscle nerve fascicles. Previous studies have demonstrated that, with voluntary contraction, the fusimotor system in activated together with the skeletomotor system. Thus, at least in «isometric» contractions, the discharge rate of muscle spindle endings increases despite the small degree of shortening that results from muscle contraction (Hagbarth and Vallbo, 1968; Vallbo, 1970, 1971, 1974). A similar co-activation of the fusimotor system can be demonstrated during rapid alternating movements, unintentional postural adjustment, Parkinsonian tremor, Parkinsonian rigidity (Hagbart et al., 1975a, b, c) and even during the choreatic movements of Huntington's chorea (unpublished observations). Conversely, the responses to stretch of spindle endings in relaxed muscles are not altered by local anaesthetic blockade of the muscle nerve proximal to the recording site, so that it appears as if spindle endings in relaxed muscles are not subjected to significant fusimotor drive. Thus, when the skeletomotor system is silent, so also is the fusimotor system - there is no resting fusimotor tonus in relaxed muscles. Additionally, it appears that the motor pathways descending from supraspinal centres activate spinal skeletomotor and fusimotor systems in parallel.

Since muscle vibration is a potent stimulus for spindle endings, capable of evoking a tonic reflex contraction in normal human subjects, and since the spindle response to vibration may be increased by the action of both dynamic and static fusimotor fibres, at least in the cat (Brown et al., 1967), a systematic study of the sensitivity of spindle endings to vibration has been undertaken in man. The activity of afferent fibres was recorded in muscle nerve fascicles, usually of the peroneal nerve at the level of the fibular head. Vibration of amplitude 1.5 mm and frequency 20-220 Hz was applied to the tendon of the receptor-bearing muscle.

Provided that the vibration was applied to the appropriate tendon and that the receptor-bearing muscle was sufficiently stretched, all spindle endings (both primary and secondary) could be activated by vibration. The effects were more powerful for primary endings. If a spindle ending was responding submaximally to the vibratory stimulus, isometric voluntary contraction of the receptor-bearing muscle enhanced the spindle response, presumably a result of apha-gamma co-activation. However, contractions involving primarily neighboring synergistic muscles rather than the receptor-bearing muscle unloaded the spindle ending and decreased the response to vibration, thus confirming previous observation that the co-activated fusimotor outflow (Vallbo, 1970; Hagbarth et al., 1975b).

If a tonic vibration reflex (TVR) developed in the receptor-bearing muscle, spindle response to vibration decreased, as if the ending had been unloaded. On voluntary suppression of the reflex contraction the spindle response reverted to its previous level, as if the ending had been re-loaded. These findings suggest that in the tonic vibration reflex, alpha-gamma co-activation is either absent or inadequate to maintain spindle discharge in the face of the slight muscle shortening that occurs in the reflex contraction. The contrasting findings obtained with a voluntary contraction suggest that the descending motor pathways activated in a voluntary contraction are not used in the reflex contraction, and it is therefore suggested that the TVR is a spinal reflex rather than a cortically-relaying reflex. If the TVR can be assumed to use those reflex pathways activated by the tonic stretch reflex, it is further suggested that the tonic stretch reflex is not a cortically-relaying reflex, contrary to recent proposals (Marsden et al., 1973). Thus, like the phasic stretch reflex (Szumski et al., 1974; Hagbarth et al., 1975c), the tonic stretch reflex is probably an exception to the principle of alpha-gamma co-activation, a property to be expected if the tonic stretch reflex is to operate as an effective load-compensating mechanism.

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THE QUANTIFIED ELECTROMYOGRAM AS A PREDICTOR OF ACTIVE-STATE IN HUMAN MUSCULAR CONTRACTION

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INTRODUCTION

Under steady-state conditions the integrated electromyogram (IEMG) can be used to represent the degree of activity in human muscular contraction. As these conditions occur infrequently during human activity, it appears desirable to estimate the instantaneous active-state (AS) of muscle rather than an averaged value such as the IEMG. Unfortunately as no independent estimate of AS can be obtained, it can only be inferred from external force by means of the dynamic characteristics of a muscle model. As the estimation of the properties of a muscle model is affected by changing activity, some assumptions must be made concerning both the model and the time-course of AS.

A number of investigators have used a first order linear filtre to represent the model relating the EMG to AS. This estimation of AS has then been used to drive the force generator (F_m) of a linear model of muscle (Fig. 1) which is also represented by a first order filtre. Calvert and Chapman (1974) syinthesized electromyograms by computer and showed considerable agreement between force predicted from twitches and force predicted from the two filters in series. The time-constant (τ_1 and τ_2) of each recursive filtre were 25 msec (Millner-Brovn, Stein and Yemm, 1973) and 70 msec (Chapman, 1973) respectively. Gottlieb and Agarwal (1971) used the same two series filters to predict foot-torque from EMG of soleus by means of two identical time-constants of 105 msec Alternatively Coggshall and Bekey (1970) used time-constants of 10 msec. and 190 msec. for τ_1 and τ_2 respectively. The success claimed by Gottlieb and Argawal (1971) may have been due to the small percentage change of maximal force used, for Coggshall and Bekey (1970) stated that their own predictions were poor when the change of force was rapid and large.

METHODS

On the basis of the work of Houk (1963), time-constants were obtained from an exponential fit to the rise of torque in a maximal isometric contraction of the elbow flexors. The value obtained was assumed to represent

τ_2 in the second recursive filter representing the mechanical model (Fig. 1). Bigland and Lippold (1954) obtained a family of force-velocity curves at different levels of IEMG, and on the basis of these results, Houk (1963)

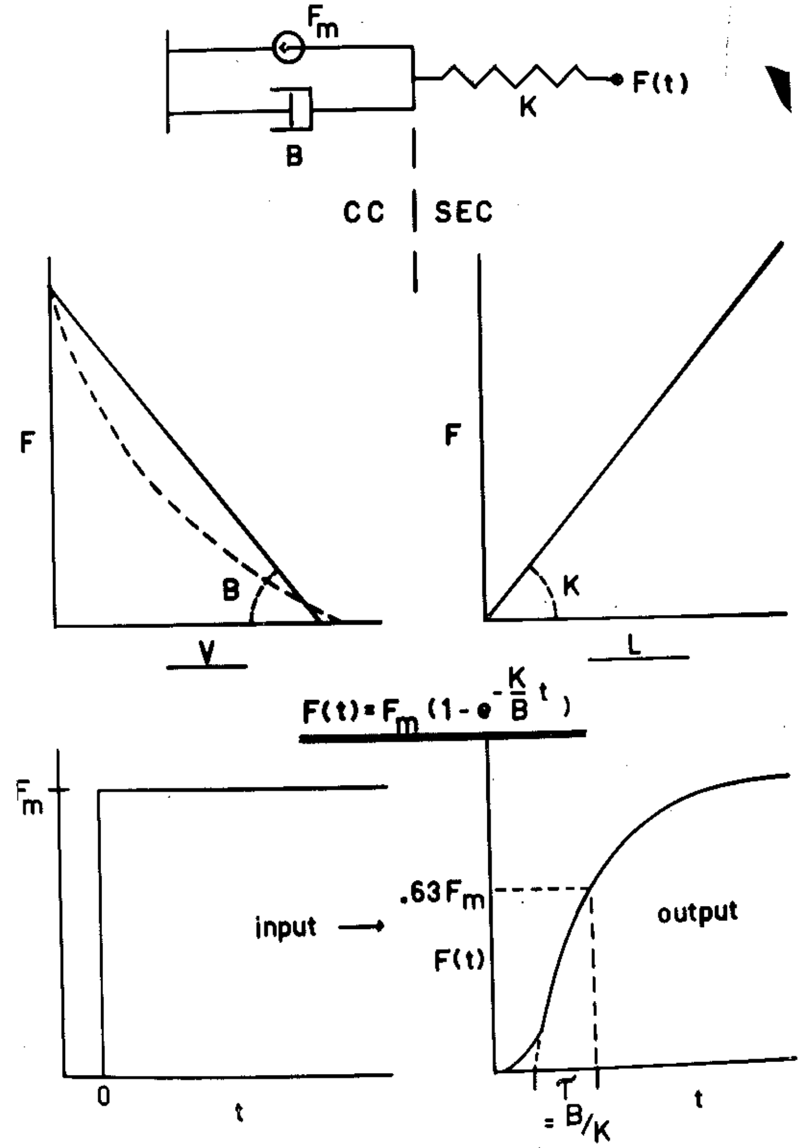

Fig. 1 - A linear model of muscle showing the mechanical properties of ist components. The equation describes the rise of force with respect to time (bottom right) on the assumption that the force generator (F_m) is activated maximally and instantaneously (bottom left).

deduced that both F_m and B were linear functions of the active-state A (see Fig. 1). Consequently the following formula:

$$A(t) = F(t)/(F_m - \frac{B}{K} \cdot \frac{dF}{dt})$$

allows calculation of the active-state (CAS(t) = A(t)) from a series of submaximal isometric contractions as shown in Fig. 2. EMG activity was recorded from biceps brachii during these contractions and the first filter (time-constant τ_1) was applied in order to obtain a theoretical measure of AS (TAS). By varying the value of τ_1 in the filter the closest agreement could be obtained between TAS and CAS (Fig. 2).

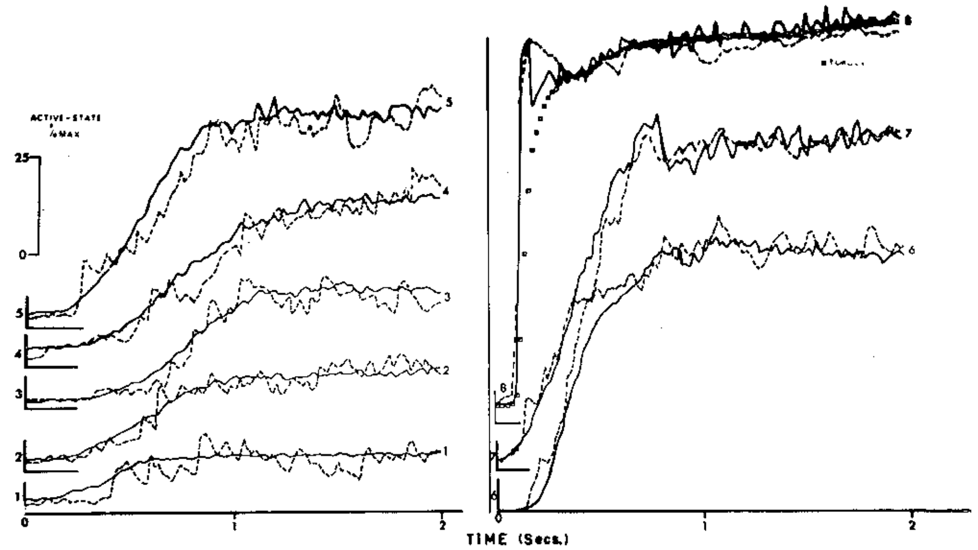

Fig. 2 - The computed active-state (C.A.S., solid line) and theoretical active-state (T.A.S., broken line) obtained with eight separate levels of voluntary muscular activation during isometric contraction. The time-constant used to obtain T.A.S. was in all cases 100 msecs. The dots in squares show torque produced by maximal effort on the eighth contraction.

The same technique was used for esimating TAS during contractions opposed with external oscillatory movement about the same angle used for isometric contractions. The results shown in Fig. 3 summarize the relationship between torque and angular velocity obtained from contractions performed at approximately 50%, 60% and 100% of maximal effort. The torque obtained at any given angular velocity during submaximal contraction was multiplied by the ratio

$$\frac{\text{maximal active-state}}{\text{TAS}}$$

in an effort both to elevate the submaximal torque-angular velocity curves to the maximal curve and to reduce the scatter of the relathionships (Fig. 3).

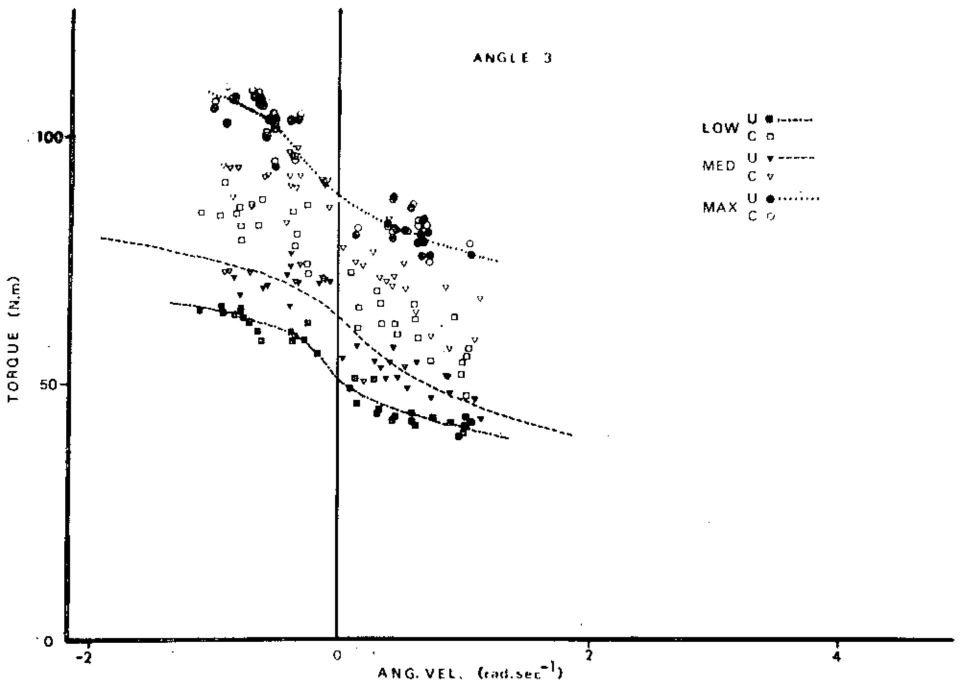

Fig. 3 - The relationships between torque and angular velocity obtained during contractions with maximal (max.) and submaximal (low. and med.) effort along with the corrections of the relationships based upon the electromyogram. Solid symbols represent uncorrected values and open symbols represent values corrected on the basis of the filtered electromyogram.

RESULTS AND DISCUSSION

On the assumption the τ_2 was estimated correctly, varius values of τ_1, were required to obtain the closest agreement between TAS and CAS.

The best compromise was obtained when τ_1 was chosen as 100 msecs (Fig. 2) although this produced a trace of TAS which was almost always more variable than CAS. Although the estimation of AS (TAS) obtained from the EMG was generally close to that calculated from torque (CAS) in isometric contraction (Fig. 2), attempts to use the value of TAS as a correction procedure in dynamic contraction proved fruitless (Fig. 3).
The problems encountered appear to centre upon the forced choice of single values of τ_1 and τ_2. Should the surface electrodes not see the true pattern of recuitment of fibres in different regions of the muscle, this type of analysis will never be successful on an instantaneous basis. Alternatively, the estimation of τ_2 from isometric contraction may be incorrectly applied as a representation of the muscle model under dynamic conditions. There is little doubt that improvements in this type of analysis will be the result of correct identification of the time-course of AS and mechanical properties of the muscles under the particular experimental conditions which are present.

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ELECTRICAL STIMULATION: A THERAPY FOR IDIOPATIC SCOLIOSIS

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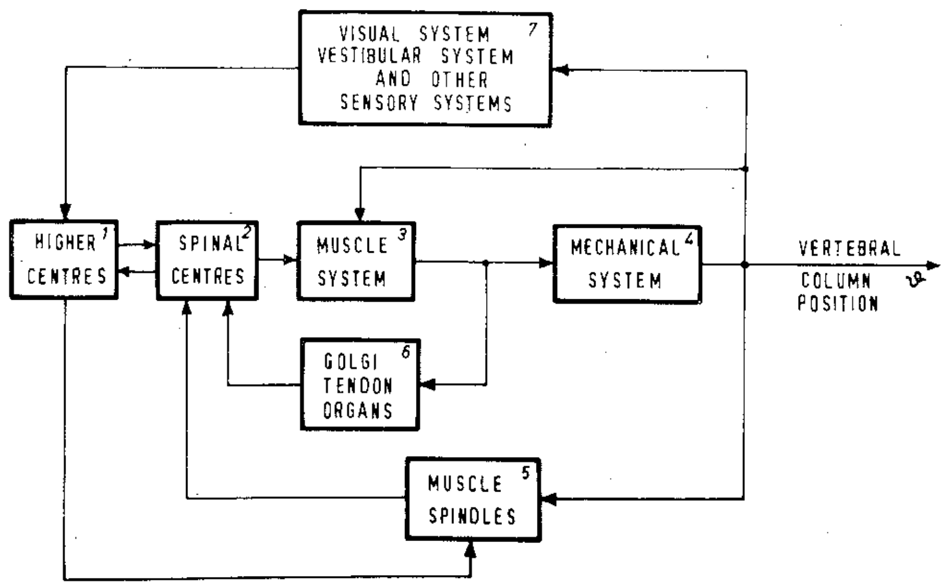
The problem of idiopathic scoliosis has been approached by many researchers who have achieved results which often were not satisfactory in spite of the efforts done, because they seemed unconnected or quite contrasting. A general vision of the posture system (Fig. 1) can allow the interpretation of all these phenomena (Dacquino et al., 1971). From this point of view this statement means that «scoliosis is due to an improper functioning of the postural control system» and the right therapy is to be based on the determination of the exact cause of the improper functioning. The problem has been how to proceed when the etiology cannot be stated or the present knowledge does not make it possible to act directly to remove the primary cause of the scoliosis. The current therapies consist in acting at block number 4 in Fig. 1 by means of external forces that oppose the deformation of the vertebral column through application of various methods (braces, traction, etc.).
On which other part of the system is it possible to act?
The muscle apparatus (block number 3) offers the best opportunity for new scoliosis therapies considering present knowledge and technology (Crivellini and Divieti, 1973). All known stimulation procedures act on the



convex side of the scoliotic curve in order to re-balance the system, but this method is not always successful (Bobechko, 1975; Gracanin, 1975). The points of stimulation have been chosen by means of the analysis of electromyographic records of the patient; as it is known that when the EMG activity is taken on both sides of the vertebral column it is symmetric for healthy people and asymmetric for scoliotic patients (Isch, 1963). The stimulation has been applied to the side which has low-level EMG activity (Crivellini et al., 1976).
The results of the stimulation procedure carried out on six patients have been shown in Fig. 2. These results are expected to be improved by determining the optimal procedure of stimulation for each patient.

Case	Sex.	Age	Scoliosis	Cobb Deg before stim. procedure	Cobb Deg after stim. procedure	improvement	Hump (mm) Before stim.	Hump (mm) After stim.	E.M.G. Before stim.	E.M.G. After stim.	Note
1	F	15	D	34°	25°	9°	15	5	A	S	
			L	41°	36°	5°			A	S	
2	M	14	D	60°	51°	9°	25	25	A	A*	A* < A
			L	44°	44°	—			A	S	
3	F	15	D	48°	45°	3°	22	17	S	S	EMG in foreward bending
			L	42°	37°	5°			B	S	
4	F	14	D	11°	8°	3°	4	2	A	S	
			L	13°	10°	3°			A	S	
5	F	12	D	30°	30°	—	15	14	A	A	Congenital scoliosis
			L	15°	15°	—			A	A	
6	M	15	D	25°	21°	4°	14	4	A	S	
			L	31°	28°	3°			S	S	

A = greater activity on the convexity side
B = " " " " concavety "
S = symmetrical activity

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A MODELLING APPROACH FOR EMG SIGNALS

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INTRODUCTION

The lack of proper description of the EMG signal is probably the greatest single factor which has retarded the development of electromyography into a precise discipline. Any useful description of analysis of the EMG signal must have physical and/or physiological significance. The schematic of Figure 1 displays the investigative approach that we have used for the past 8 years. During this time, any new insight in the EMG signal has always balanced between theoretical developments and practical applications.

GENERAL COMMENTS

In general, it is useful to obtain relationships between the observable EMG signal and the physical and physiological properties of muscles. However, the observable EMG signal that is subject to description is dependent on the characteristics of the recording devices as well as the properties of the muscle generating the signal. It is, therefore, necessary to consider, or at least be aware of, the characteristics of these intermediate effects displayed in Figure 2.
The Recording Equipment Transfer Function can be disregarded by choosing equipment of sufficiently wide bandwidth (0 to 3000 kHz) and a high fidelity so that it will not affect the properties of the EMG signal. The Electrode Transfer Function is dependent on the type of electrode being used: monopolar, bipolar, co-axial, wire type, surface, etc. This transfer function can be empirically obtained for any specific electrode if the particular experiment warrants a great accuracy. An example of a useful procedure con be found in De Luca and Forrest (1972), Kadefors et al. (1968). In general the amplitude of electrode transfer functions is directly related to the frequency. Hence, the lower frequency components of the EMG signal are relatively more attenuated than the higher frequency components. The electrical properties of the muscle, fascia, fat and skin tissues also affect the observable EMG signal. Their conductivity and frequency filtering characteristics will affect the EMG signal as it proceeds from the surface of the muscle fibers to the recording electrode(s). These latter

characteristics have not been completely determined, but are currently under investigation.
Given that the intermediate influences can be accounted for, the relationship between the observable EMG signal and the physical and physiological properties of a muscle may not be necessarily similar for different muscles, due to the large variation in the physiological and anatomical properties of muscles. In any case, when pursuing such a relationship, it is necessary to specifiy the specific contraction performed by the muscle. Muscle contractions can be categorized by the following properties:

force	*dynamics*	*time*
constant force	isometrics constant velocity	all contractions are a function of time
linear force	linear velocity	
non-linear force	non-linear velocity	

Any force and dynamics combination is possible, but all muscle contrac tions are time dependent.

Fig. 1

MODEL

We have begun our EMG signal modelling approach by considering the simplest muscle contraction, i.e., the constant-force isometric contraction. A specially designed needle electrode was used to record motor unit action potentials (MUAP's) for the complete time duration of constant force isometric contractions varying in discrete steps from minimal to maximal force levels (De Luca and Forrest, 1973). Several properties of the motor unit action potential trains (MUAPT's) were observed. In particular, the inter-pulse intervals between adjacent MUAP's of a particular MUAPT were measured and analyzed as a real continuous random variable. The distribution of the inter-pulse intervals was described by the Weibull probability distribution function with time and force dependent parameters. The statistical description culminated in the development of a model for the MUAPT's (De Luca, 1975). The model was based on the derivation of the autocorrelation and crosscorrelation functions of MUAPT's. These function were solved by using a Dirac Delta function approximation for the MUAP. This acceptable approximation sufficiently simplified the mathematics so that the model could be extended to EMG signals.
The EMG signal was modelled as a linear summation of the MUAPT's. Consideration was given to the effects of signal cancellation due to *superposition* of positive and negative phases of the constiuent MUAPT's

and to the effects of possible *synchronization* of the MUAPT's throughout a contraction. The model was used to derive time and force dependent expressions for the following parameters of the EMG signal: the mean rectified value, the mean integrated rectified value, the root-mean-squared value, and the variance of the rectified EMG signal (De Luca and van Dyk, 1975). For each parameter it was possible to relate various terms in the derived expressions to physiological phenomena that occur in contracting muscles. One of the parameters, the mean rectified value of the EMG signal, is shown in Figure 3. The variables have the followin description:

τ	Contraction time normalized with respect to the total time duration of the muscle contraction
ϕ	Constant force at which the isometric contraction is performed, normalized by the maximal contraction force
$m(\tau,\phi)$	The time and force dependent EMG signal
$E[\lvert m(\tau,\phi)\rvert]$	The mean rectified EMG signal
$\lambda(\tau,\phi)$	The *generalized firing rate* of the MUAPT's
$\overline{\lvert h_i(\tau)\rvert}$	The area of a rectified MUAP
$J(\tau,\phi)$	A variable used to account for the amount of cancellation due to superposition.

Fig. 2

The *generalized firing rate* is defined as the firing rate of a typical motor unit and has been defined analytically by a mathematical expression (De Luca and Forrest, 1973).

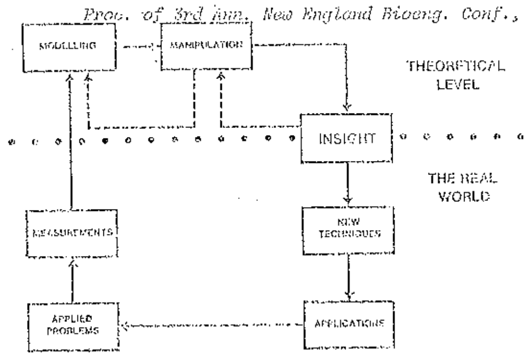
The particular expression in Figure 3 is the simplest of the derived expressions. The others contain additional terms reflecting the effect of MUAPT synchronization. By contrasting the various terms of the equations of the parameters it is possible to determine which EMG signal parameter most strongly reflects the effects of a particular physiological phenomenon. Experiments are presently in progress to compare the results of the model to empirically measure values of the corresponding EMG parameter. So far, there has been good agreement between the model and empirical measurements for the variance of the mean rectified EMG signal (Stulen and De Luca, 1975.

The model was derived without consideration of the intermediate effects described in Figure 2; however, these characteristics only affect the shape of the MUAPT, $h(\tau)$. Hence, they can be introduced in the model as transfer functions multiplying $h(\tau)$.

Although more work is required to prove that the described modelling approach is correct for the EMG signal constant force isometric contrac-

tions, our current results are encouraging. Accordingly, we have begun experiments and theoretical analysis to describe the EMG signal of the next hopefully describably muscle contraction, i.e., the isometric linear force-varying contraction.

Proc. of 3rd Ann. New England Bioeng. Conf., pp. 16-24, 1975.



Acknowledgement

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Figure 1. Investigative approach for EMG signal analysis and application.

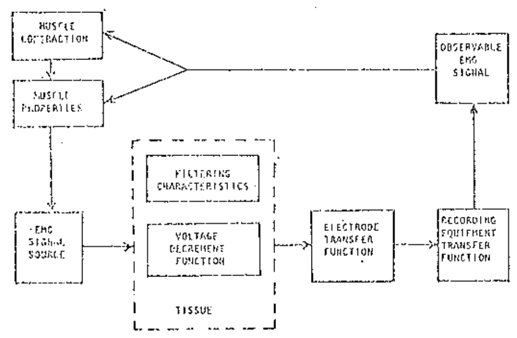

Figure 2. Relationship between the observable EMG signal and the muscle properties.

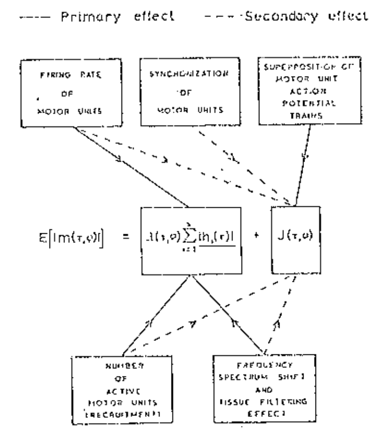

Figure 3. The expression for the mean rectified value of the EMG signal and its physiological correlates.

Fig. 3

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DIFFERENCES BETWEEN NORMALS, MYOPATHIES, AND POLYNEUROPATHIES OBSERVED IN PHYSIOLOGICAL TREMOR POWER-SPECTRA

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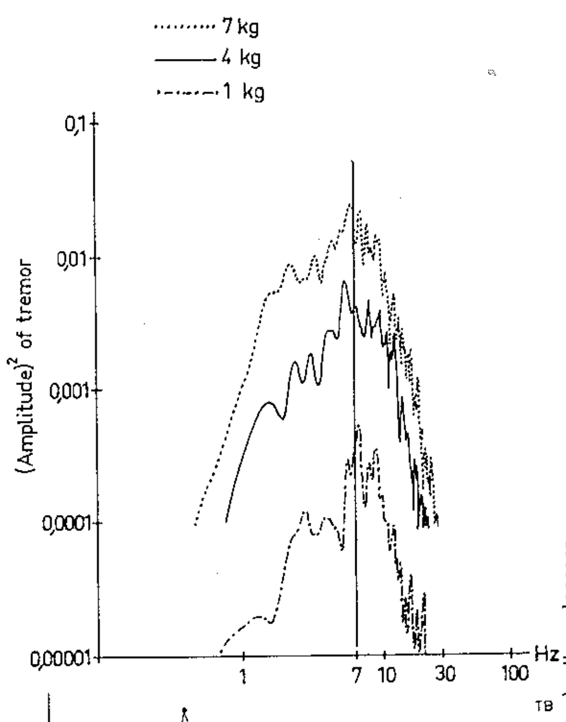
We have proposed elsewhere (Allum et al., 1976) that the shape of the physiological tremor-spectra in normals is solely the result of partially fused motor unit discharges firing between 6 and 25/sec. The characteristic fall-off of the power-spectrum for frequencies above 10 Hz and its changes with increased isometric force can, by this hypothesis, be explained on the basis of known recruitment and rate coding properties of motoneurons (Henneman et al., 1974). In addition, the tendency for a variation in tremor amplitude at the same force level in different subjects has been described as the result of a difference in the degree of synchronization between motor units (Dietz et al., 1976). When the discharge pattern of a motor unit is altered in a disease of the motor system, one would expect the tremor-spectra to show a corresponding alteration. Thus we have chosen to analyse the power-spectrum of myopathies and polyneuropathies and to compare them with those of normals. Our interest was firstly to improve and extend diagnostic possibilities of these disease, and second to obtain further evidence for the hypothesis described above.
The methods for recording and frequency analysis of physiological tremor have been described elsewhere (Allum et al., 1976). Basically, subjects exert a constant isometric force between their thumb and forefinger which is measured with a strain gauge bridge, high pass filtered to improve resolution between 5 an 30 Hz and then Fourier analysed on a digital computer.
Single power-spectra of a normal subject are shown in Fig. 1 for 3 different isometric force levels. They contain significant power for frequencies up to 25 Hz corresponding to the known maximum discharge frequencies of motor units (Freund et al., 1975). A peak at 7-8 Hz (the dominant tremor rate) is seen in the spectra at all 3 contraction levels. This frequency represents the firing rate of newly recruited, least fused motor units. Thereby, the mechanical effect of these larger twitch amplitude motor units is seen to be greater than those of earlier recruited smaller twitch amplitude motor units. Units which are recruited first have twitch contractions which become fused as isometric force increases. All motor units are recruited at the same discharge rate of 7-8/sec (Freund et al., 1975; Milner-Brown et al., 1973), hence the tremor peak remains constant throughout the recruitment domain.

Previous investigations have shown that average firing rates of motor units for a fixed force level are higher in myopathic patients than in normals (Dietz et al., 1975). Original tremor force recordings also show that the dominant tremor rate of 7-8 Hz is less pronounced in myopathies (see Fig. 2). Power-spectra of 2 typical mildly affected myopathies reveal these findings clearly. In Fig. 3 tremor-spectra from the 2 myopathies can be compared with the average spectra of normals for a force level of 1 kg. No clinical symptoms were observed in these patients' hand muscles. From the figure it is apparent that tremor amplitudes between 5 and 15 Hz are significantly reduced. Correspondingly amplitudes at higher frequencies (above 15 Hz) are relatively larger (if the value at 7.5 Hz is taken as a datum). Once myopathic patients exert greater muscular force (3 kg, for example) their tremor spectra have a more normal appearance. We conclude therefore that higher threshold units are less affected by the disease. Generally, we can distinguish the power spectra of a myopathy by its flat appearance and its relatively greater high frequency content.

Similarly, neuropathies can be distinguished by a characteristic shape to the power-spectrum. Again the changes in the shape with respect to normals are a reflection of the pathological symptoms of the motor system. From single motor unit recordings from patients with uremic polyneuropathies it is known that motor units at higher discharge rates have a greater tendency to discharge failure. A failure which is probably due to focal demyelination of peripheral nerves (Dietz and Freund, 1974). Power spectra from 2 neuropathies are compared with the average curve for normals in Fig. 4 (force level 3 kg). The neuropathic power-spectra are sharply pointed with a steep fall-off for frequencies above 15 Hz reflecting the failure of units to fire at high frequencies. In chronic neuropathies the peak of the power-spectra occurs at a low frequency with respect to acute cases. This is documented in Fig. 4 for a patient with peroneal palsy and another with a 3 week-old Guillain-Barré polyneuroradiculitis. These latter observations agree with those of Larsson (1975) who found for acute and chronic neuropathies differences in the spectra of the intramuscular EMG.

Pathological states of the motor system may be observed in the alterations to the properties of motor units. We have observed correlated changes in physiological tremor power-spectra. These changes lead us to believe that physiological tremor is solely the result of the motoneuron pool output and its resultant twitch contractions and not oscillations due to spinal or central reflex arcs. Characteristic changes in the shape of power spectra of myopathies and neuropathies, as compared to normals, which are apparent in the early stages of these diseases suggest that the physiological tremor pover-spectrum can be a valuable diagnostic tool. For the patient the method has the advantage that it is quick and painless; for the clinician it provides a quantitative technique with which long term observations may be followed.

Fig. 1



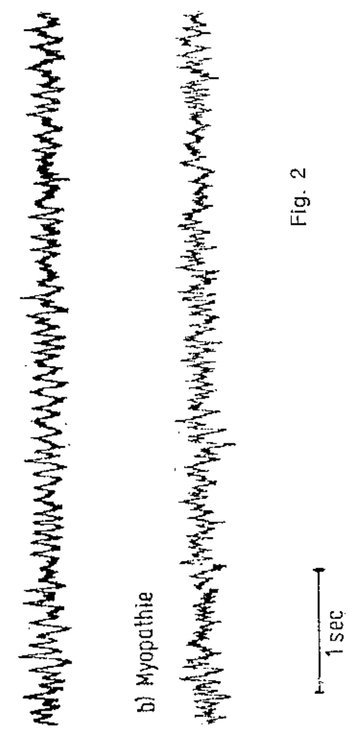


Fig. 2

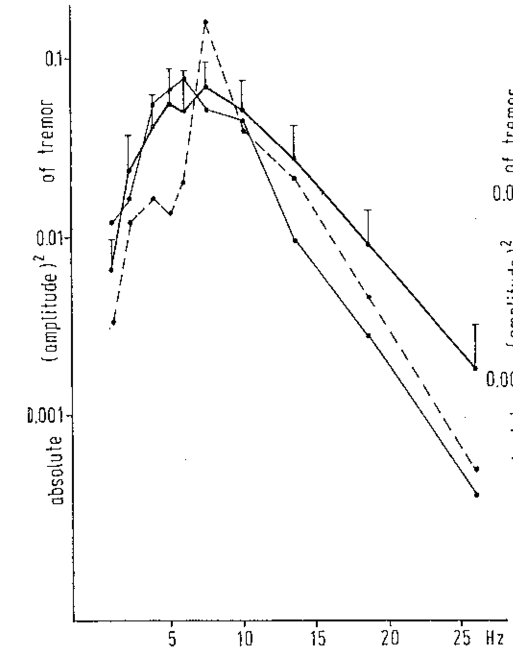


Fig. 3

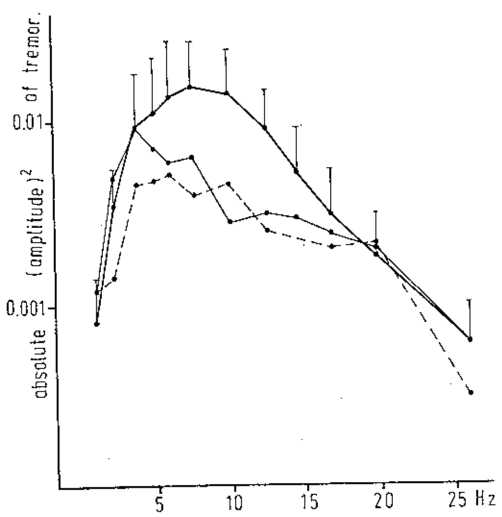


Fig. 4

Fig. 1 - *Original power spectra at force levels of 1,4 and 7 kg.* All are single spectra from the same subject. The effect of increasing force on the tremor amplitude is clearly seen. 0.1 on the ordinate corresponds to 160 gm².

Fig. 2 - *Physiological tremor recordings* a) from a normal and b) from a myopathic patient.

Fig. 3 - *Average tremor spectra for 7 normals compared with 2 myopathies* (force level 1 kg). The average curve is the thick line and the patients the dotted and thin lines. Single spectra from each subject were calculated, then averaged at several frequency bins. The centre of the bin is at each plotted point, and edges are equidistant between points. Standard deviations are shown by the vertical bars. 0.1 on the ordinate corresponds to 9.2 gm².

Fig. 4 - *Average tremor spectra for 7 normals compared with 2 neuropathies* (force level 3 kg). Patients are shown by this and dotted curves. Otherwise, rotation is as for figure 3.

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ELECTROPHYSIOLOGICAL AND PATHOLOGICAL STUDY OF THE MOTOR AND SENSORY NERVE FIBERS IN EXPERIMENTAL NEUROPATHY

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The purpose of this study is the neurophysiological and histological evoluation of organophosphorous insecticides chronic poisoning in rats.
These compounds were given by mouth, incorporated in the diet, or by inhalation in a continous controlled manner. A dose of 8-6-4 mg/kg administered 5 times a week was used. The motor and sensory conduction velocity were serially measured in vivo.
This investigation was performed with a simple method on motor and sensory fibers in rat's tail. Repetitive nerve stimulation was also performed for detecting defects of neuromuscular trasmission.
At the end of the neurophysiological investigation in all the experimental animals, the dorsal nerve of the tail was examined histologically with osmium tetroxide method. The diameter of every nerve fiber was measured. The fiber density and the fiber diameter distribution of the myelinated fibers in each nerve were calculated.

A PHYSIOLOGICALLY ORIENTED MODEL OF THE PACINIAN CORPUSCLE

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Pacinian corpuscles as receptors of mechanical events are thought to play a relevant role in motor control in general, and particularly for reflex regulation. From many years, due to its easy accessibility and its relatively large dimensions, Pacinian corpuscle has been rather thoroughly studied by several authors (see, e.g. Loewenstein and Skalak, 1966; Hubbard, 1958; Ilyinsky, 1965) and many salient features of ist behaviour have been evidenced; nevertheless, at present, some physiological aspects of the mechano-to-neural transduction have not been completely understood. An approach based on a detailed and quantified description of the relationships characterizing the Pacinian corpuscle behaviur is believed to be a constructive step to a better understanding of the matter and moreover it can provide useful suggestions in order to clarify its role as transducer of the dynamical aspects of mouvement.

An investigation of the electrophysiological features of the mechano-to-neural transduction, based upon the determination of a mathematical model, is presented in this paper.

For this purpose, the knowledge of the morphological structure of the system and the analysis of outputs in response to known inputs, besides the anologies existing between Pacinian corpuscles and other similar receptors have been taken into account following the classical identification methods of the Systems theory.

As shown in Fig. 1, the displacements of the outermost lamella and the pressure on the core have been taken as input and output of the mechanical part, respectively. This part of the system has been described at a linear system (see, e.g. Grandori and Pedotti, 1975), and therefore represented by means of the transfer function G_1 (s).

By analysing the outputs of the intact corpuscle and the decapsulated nerve terminal, in response to various kinds of mechanical stimuli, a typical non-linear, directional dependent phenomenon is found (Nishi and Sato, 1968; Ilyinski, 1965); therefore in the model of the sensory ending a non-linear transfer characteristic has been introduced.

The derivative and G_2 (s) point out the dynamic properties of the nerve terminal, as clearly appearing in the experimental results of the decapsulated terminal (see for instance, the Panel II of Fig. 2).

The threshold-like phenomena and the stochastic properties concerned in the generation of the on-off responses were accounted for by means of a

time-varying threshold block S with a zero mean valued gaussian noise superimposed.

The model has been implemented on a digital computer in FORTRAN language and usual experimental tests were simulated. Some results of the simulation are summarized in Fig. 2.

Panel I of Fig. 2 shows on the right the responses, in terms of receptor potential generated by the model of an intact Pacinian corpuscle compared with the experimental results of Nishi and Sato (1968), reported on the left. In each figure, lower traces represent the stimulus at the output, plotted in arbitrary units (the same for all the traces). Upper traces show the time course of the receptor potentials of the intact corpuscle, in μV.

As to the late depolarizing responses of the model these can not be compared with the above experimental results where the latter are not reported; but they are however present in Ilyinsky's data (1965) representing the everall response.

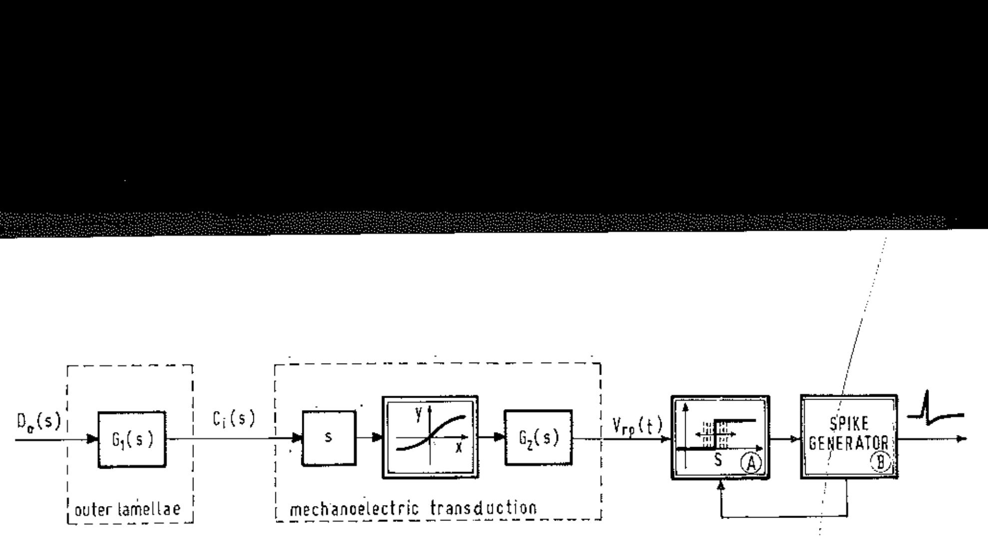
Panel III shows on the right the response of the model of an intact corpuscle compared with the experimental results by Ilynsky (1965), reported on the left. In this case the rate of increasing of the stimulating waveform is greater than the rate of decreasing and therefore the late response is weaker than the early depolarizing response.

At the light of the modeling, this phenomenon appears to be an obvious direct consequence of the shape of the stimulus and not a peculiar property of the corpuscle, as misinterpreted by Nishi and Sato (1968), pag. 389.

The structure of the model permits the simulation of the receptor potential generation following the direct stimulation of the decapsulated corpuscle. On the right part of the Panel II of Fig. 2, the results of this simulation are reported and compared with Nishi and Sato's experimental findings on the left.

The results achieved by implementing the complete model confirm that the appearance of a spike is determined by the depolarization of the nervous ending and, apart from the influence of an inherent biological noise, the mechanisms of generation of the off-responses are the same as those of on-responses.

Some of these results are given in Panel IV of Fig. 2. From the top to the bottom of the figure there can be seen: the mechanical stimulus delivered at the input of the model; the receptor potential and two sequences of discharges obtained in response to two different presentations of the same stimulus.



$$D_e(s) = \mathcal{L}\left[d_e(t)\right] \qquad G_1(s) = \frac{\mu_1(1+sT_1)}{(1+sT_2)(1+sT_3)}$$

$$C_i(s) = \mathcal{L}\left[c_i(t)\right]$$

$$V_{rp}(s) = \mathcal{L}\left[v_{rp}(t)\right] \qquad G_2(s) = \frac{\mu_2}{1+sT_c}$$

Fig. 1 - Block-diagram of the mathematical model.

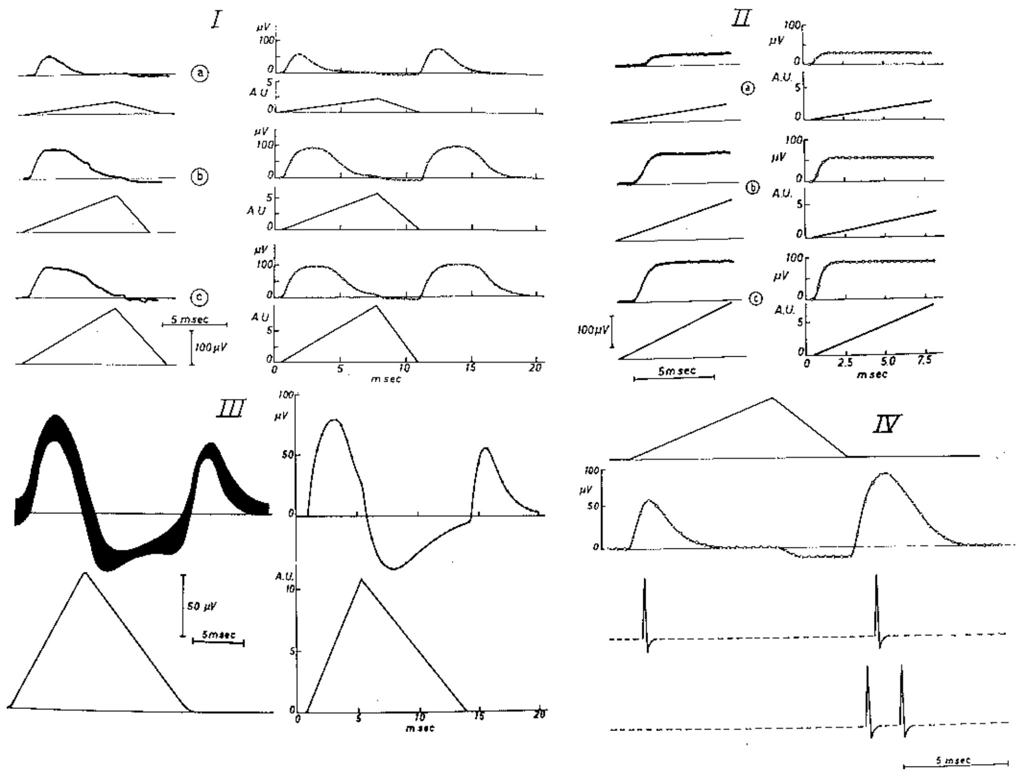


Fig. 2 - Some results of the simulation (see text).

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CONTINUOUSLY AND INTERMITTENTLY DISCHARGING MOTOR UNITS IN MAN: FLEXIBILITY OF THE VOLUNTARY RECRUITMENT ORDER

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The discharge pattern and recruitment order of individual motor units on voluntary contraction have been studied in electromyographic recordings. The anterior tibial and the short toe extensor muscles were tested (Hannerz 1974, Grimby and Hannerz 1976). Both muscles consist of type I and type II muscle fibres. The short toe extensor muscle fibres are type-grouped also in certain clinically normal subjects.

Electromyographic recordings with a selectivity permitting identification of individual motor unit potentials even on maximum contraction were used. In the anterior tibial muscle sufficient selectivity was obtained only with high impedance electrodes (electrode surfaces about 250 Kiloohm and 4 megaohm respectively) and sufficient stability only if the electrode was thin and bendable (100 μ silver wires) and equipped with a hook for fixation in the muscle. In the type-grouped toe extensor muscle, however, sufficient selectivity could be obtained with ordinary low impedance electrodes.

Certain motor units were found to discharge continuously at intervals as long as 100 ms and did not discharge at shorter intervals than about 30 ms on tonic voluntary contraction even though shorter intervals can be obtained on phasic voluntary contraction. It is suggested that these continuously discharging long interval motor units (CLMUs) have type I muscle fibres. Other motor units were found to discharge only intermittently and only at intervals shorter than 30-60 ms. It is suggested that these intermittently discharging short interval motor units (ISMUs) have type II muscle fibres. A considerable number of the motor units were, however, intermediate forms between CLMUs and ISMUs. The two criteria of motor unit type, viz. discharge endurance and discharge intervals, were partly independent of one another.

In slowly increasing contraction strength CLMUs were recruited before ISMUs and on prolonged tonic contraction only CLMUs discharged. This is in accordance with the Henneman principle (Henneman, Somjen and Carpenter 1965). In very rapidly increasing contraction strength, however, certain ISMUs were recruited before certain CLMUs. Selective activation of ISMUs was found to occur in contractions intended to be very short but powerful.

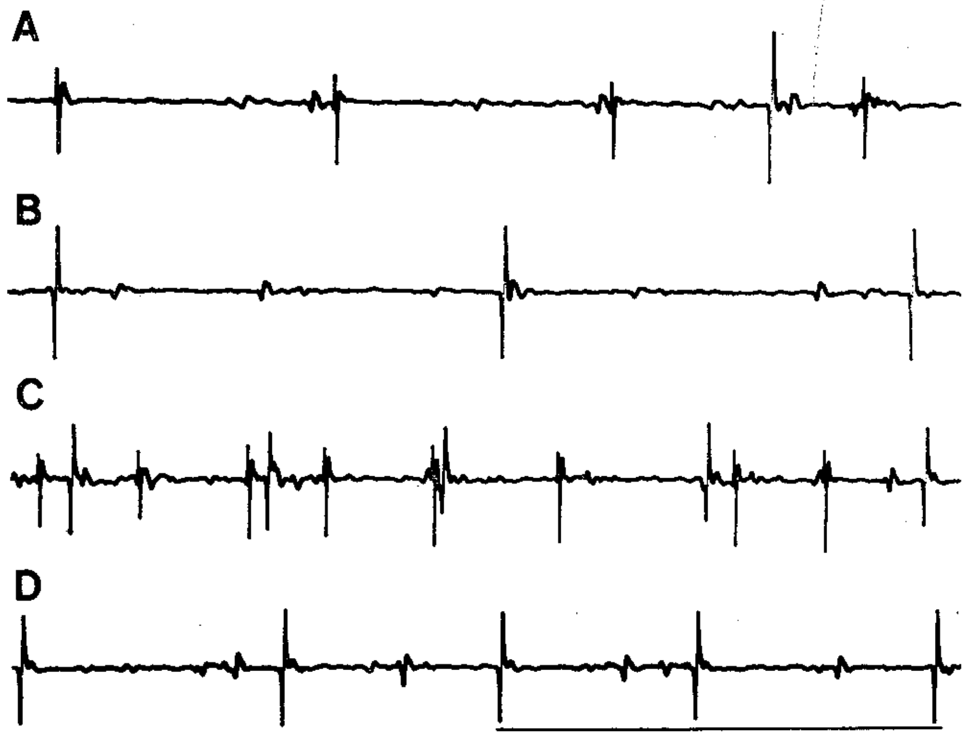

Fig. 1 - Four consecutive parts of a voluntary contraction. A) Rapidly initiated contraction. B) Weak maintained contraction. C) Rapidly increased contraction to maximum strength. D) Maintained maximum effort.
The potentials of one CLMU and one ISMU can be identified. On rapid initiation the ISMU in active 150 ms before the CLMU. On weak maintained contraction only the CLMU discharges. On acceleration towards maximum strength both units discharge but the ISMU at a higher rate than the CLMU. Finally on prolonged maintained maximum effort only the CLMU responds. Type-grouped short toe extensor. Low impedance electrode. Time bar 100 ms.

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THE IMPORTANCE OF EVALUATION OF THE LENGTH OF THE SUMMATED DEPOLARIZING ZONE IN ACTIVE MOTOR UNITS FOR THE ELECTROMYOGRAPHIC STUDIES

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It has been established that under different muscle conditions such as fatigue, ischemia, and temperature change there occur changes in the summated depolarizing zone (Gydikov and Kosarov, 1971, 1972). This process has been detected by the use of the method of vector-electromyography of action potentials from different motor units, and surface electrodes have been employed.

This paper is concerned with a new method which has been developed for precise measurement of the summated depolarizing zone. It has been found, for example, that under normal conditions the length of this zone in biceps brachii was 17.88 (± 0.36) mm. The changes in the length of the zone have been provoked, in turn, by different influences in muscle, as well as in the case of some muscle diseases. The length change was a result of both the change in the summary period of depolarization and repolarization at a given point along the motor unit, and the change in the propagation velocity of depolarization along the muscle fibres. The change in the length of the depolarizing zone was found to be accompanied, as well, by changes in action potential amplitude.

It has been determined through modelling that the extra-cellular potential amplitude in the volume conductor depends on the depolarizing zone, and the amplitude varies at different distances from the muscle fibres. In that the change in the length of the zone determines various changes in the potential amplitude at different radial distances, it was obvious that these changes were not due so much to desynchronization as to a change in the length of the depolarizing zones in different fibres. These results would suggest the value of measuring the summated depolarizing zone in electromyographic studies.

THE RELATIONSHIP BETWEEN THE AMPLITUDE DISTRIBUTION OF THE ABSOLUTE MYOELECTRIC SIGNAL AND THE MUSCULAR WORK PERFORMED

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INTRODUCTION

In ergonomic studies the registration of the myoelectric activity is commonly used to increase the information about the muscular work performed. It has previously been shown by Lippold (1952) that there is a relationship between myoelectric signal level and the force developed by the muscular contraction. A similar relationship has been reported to exist between the integrated myoelectric activity and the performed muscular work (Bouisset, 1968; Hagberg and Jonsson, 1975).

It has been demonstrated by Björkstén and Jonsson (in press) that there is a marked difference in muscular endurance between static and intermittent static contractions. The amplitude probability distribution function of the fullwave rectified and lowpass filtered myoelectric signal (in the following referred to as the *amplitude function*) has been shown to give a quantitative measure of static and dynamic components of the muscular work (Jonsson and Hagberg, 1974; Hagberg and Jonsson, 1975; Jonsson 1975).

The aim of the present study was to investigate how this amplitude function reflects the performed muscular work measured from the signal from a force transducer.

METHODS

The myoelectric activity was recorded from the belly of the right biceps brachii muscle by skin electrodes. The myoelectric activity was then linearly amplified and recorded on tape.

The investigation was performed with the subject in a sitting position. A forearm splint construction was set up in which the subject's right arm was fixed with 90 degrees of flexion in the elbow as well as in the shoulder joint. The forearm splint was directly connected to a force transducer. The force transducer signal was visualized in front of the subject for tension level control. The force transducer signal was also recorded on tape.

The experiments consisted of static and intermittent static contractions of the elbow flexors:

1. 5 minutes of continous static contractions at 5 and 10% of the maximal voluntary force of contractions (MVC).
2. 5 minutes of intermittent static contractions at 5 and 10% of MVC.
3. A complex series of intermittent static contractions with different degrees of contraction levels.

The used amplitude function (see introduction) was estimated by calculating the ratio between the detected cumulative time the lowpass filtered absolute myoelectric signal was lower than or equal to the mean amplitude of the absolute myoelectric signal corresponding to different levels of MVC, and the total detection time. The same analysis was performed on the signals from the force transducer.

RESULTS AND CONCLUSIONS

The used amplitude function (see introduction) of the myoelectric signal and the force transducer signal showed a fairly high degree of similarity. This may indicate that the used amplitude function of the myoelectric signal closely reflects the work performed by the muscle.
The amplitude function of the myoelectric signal and the force transducer signal will be influenced by the time constant used when the absolute signals are lowpass filtered. The most suitable time constant in this respect appears to be in the magnitude of 100 ms.

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MUSCLE POTENTIATION ANS INTRACELLULAR REGULATION OF THE ACTIVE STATE INTENSITY

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When a skeletal muscle is directly stimulated, or indirectly through its motor nerve, the time course of the isometric twitch elicited, and the peak force developed have been discussed, for a long time, according to Hill's model of a two-component system, where the contractile elements are arranged in series with the elastic elements. In his model, it was further postulated that the active state rises rapidly to its maximum, and muscle potentiations were therefore generally related to a prolongation of the active state. We have clearly demonstrated (Desmedt and Hainaut, 1968) that, in staircase potentiation recorded from a normal human muscle «in situ», the active state intensity is not maximum and therefore concluded that Hill's model is not applicable to mammalian muscle at body temperature. Our suggestion was a two-component active state where the intensity of activation of the myofilaments would be regulated by the intracellular Ca^{2+} kinetics.
In a previous paper presented at the 4th International Symposium on Biomechanics (Penn State, USA, 1973), we reported that this potentiation could neither be related to a lowering of the electro-mechanical threshold of the isolated frog muscle fibre. This was another evidence that some step in the intracellular processes regulating the electromechanical coupling was intensified during staircase potentiation. These findings gave further support to the two-component activation model suggested earlier (Desmedt and Hainaut, 1968).
In recent experiments, we have analysed the intracellular Ca^{2+} movements during standard membrane depolarizations, using the photoprotein «Aequorin» as intracellular Ca^{2+} indicator, and ^{45}Ca micro-injections were made directly into the cytoplasm of giant muscle fibres to follow Ca^{2+} movements across the outer membrane. Our results indicate that it is indeed possible to intensify the active state, even in a single twitch, by a direct action on the intracellular Ca^{2+} movements (Hainaut and Desmedt, 1974_a; 1974_b; Desmedt and Hainaut, 1976). On basis of these experimental results, and others, we will rediscuss the different types of muscle potentiations.

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NEURONAL MECHANISMS OF THE OPHTHALMOMIMIC MUSCLES (POLYGRAPHIC EMG STUDY)

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Ophthalmomimic is the interaction between the extra-ocular eye-muscles and the mimic-muscles (1-4). Beside their original assisting function in looking and hearing the ophthalmomimic muscles serve in psychological communication. Some special mechanisms of synergistic coinnervation are polygraphically analyzed by the help of the emg-needle.

a) BELL's phenomenon:
   The facial nerv induces the oculomotor nerv.
   Psych.: devotion
   Proverb: to turn a blind eye to

b) «Anti-BELL»:
   The oculomotor nerv induces the trochlear nerv.
   Psych.: attention
   Proverb: to be all eyes

c) Oculoauricular Phenomenon:
   The abducens nerv induces the facial nerv.
   Psych.: caution
   Proverb: to prick up one's ears

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EMG TO FORCE PROCESSING UNDER DYNAMIC CONDITIONS

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INTRODUCTION

Electromyography has found widespread and fruitful application in kinesiology, however mostly in a rather qualitative way. This in spite of the existence of some well established empiric relationships concerning quantitative interpretation of the emg. The best known one is that the mean rectified emg is linearly proportional to the isometric muscle force under static conditions. The conditions under which this relationship holds have been widened in two ways:

a) In isometric contractions the rectified emg is, after suitable low-pass filtering, a measure for the time varying isometric force (e.g. Gottlieb and Agarwal, 1971).

b) In isotonic contractions with constant velocity the mean rectified emg is linearly proportional to the active state of the muscle (Bigland and Lippold, 1954).

We have combined and extended these two lines of development, in order to cover the general case of non-stationary emg and time varying muscle length, by assuming that the rectified emg of a muscle, after filtering with a low-pass filter whose impulse response has the same form as the active-state curve of a twitch, is linearly proportional to the momentary value of the active-state of the muscle. When this signal is used as an input to an electrical analogue of the Hill muscle model the muscle force can be obtained instantaneously for any kind of muscular contraction.

PROCESSING

The block diagram of the emg processor is represented in Fig. 1. It is based on the Hill muscle model (Jewell and Wilkie, 1958). Its inputs are the emg and the angular position of the joint on which the muscle acts. Measurements and parameter values refer to M. triceps surae (M. soleus and M. gastrocnemius cap. mediale and laterale), the prime movers of the ankle joint. For sake of convenience torques (M) and angles (ϕ) are used instead of muscle force and -length.

1. Preprocessing

Surface emg's of the soleus and of both heads of the gastrocnemius are amplified, bandfiltered (30-600 Hz, 12 dB/oct) in order to suppress artifacts and noise, summed with adjustable weighting factors and full-wave rectified. The other input, the joint angle ϕ, measured between lower leg and foot, is recorded by a potentiometer.

2. Active state

The active state M_o is obtained from the rectified emg by low-pass filtering with a third order averaging filter, the transfer function of which is

$$H(s) = \frac{2T^2s^2 + 120}{T^3s^3 + 12\,T^2s^2 + 60\,Ts + 120}$$

T is the averaging time. It has been shown (Kreifeldt, 1971) that this filter has the smallest ripple with emg-filtering at a given settling time. This advantage offsets the deviations from the active-state curve of the impulseresponse (Fig. 2a).

3. Torque-angle relation

This function (Fig. 2b) has been taken as

$$M(\phi) = M_o \cdot f(\phi) \quad \text{with } f(\phi) = \begin{cases} 1 & \text{for } \phi < \phi_2 \\ \dfrac{\phi - \phi_1}{\phi_2 - \phi_1} & \text{for } \phi > \phi_2 \end{cases}$$

($\phi 1$ and $\phi 2$ being constants) with a gradual changeover between these segments.

4. Hill relation

Rewriting the well-known Hill relation for muscle lengths shorter than resting length (Abbott and Wilkie, 1953) yields:

$$M(\phi,\dot{\phi}) = M_o \frac{f(\phi) - n\dot{\phi}_c/b}{1 + \dot{\phi}_c/b}$$

$n = a/M_o$ and b being the parameters of the Hill relation. This function is bounded by $0 < M(\phi, \dot{\phi}) < M_o\ f(\phi).\ (1 + c_1)$ and so covers both positive and negative shortening velocities (Fig. 2c).

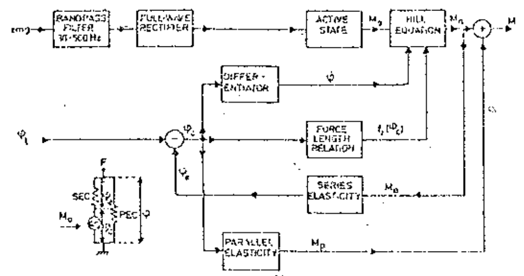

Fig. 1 - Block diagram of the processor.

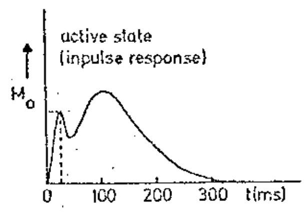

Fig. 2a

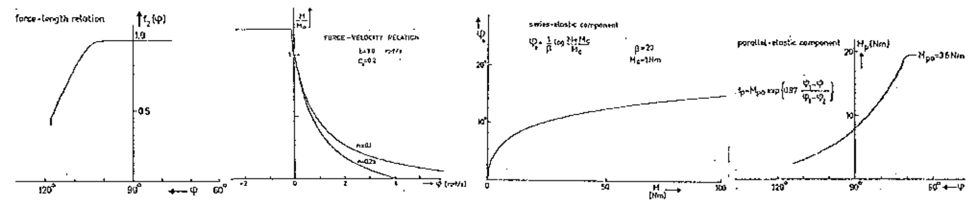

Fig. 2 b c d e

Fig. 2a/e - The components of the muscle model.

5. Series-elastic component

The SEC has been given a logarithmic characteristic (Fig. 2d).

$$\phi_e = \frac{1}{\beta} \ln \frac{M + M_c}{M_c}$$

With M_c (= 1Nm) serving only to make $\phi_e = 0$ at $M = 0$.

6. Parallel-elastic component

This component has been included to represent the mechanical properties of the passive muscle. Direct measurements show that it can be written as an exponential relation (Fig. 2e).

$$M_p(\phi) = M_p(\phi_2)\ \exp\left\{-\,0.87\ \frac{\phi - \phi_2}{\phi_1 - \phi_2}\right\}$$

RESULTS

In order to check the system performance, model output was compared to the measured foot torque. The subject was standing with one foot

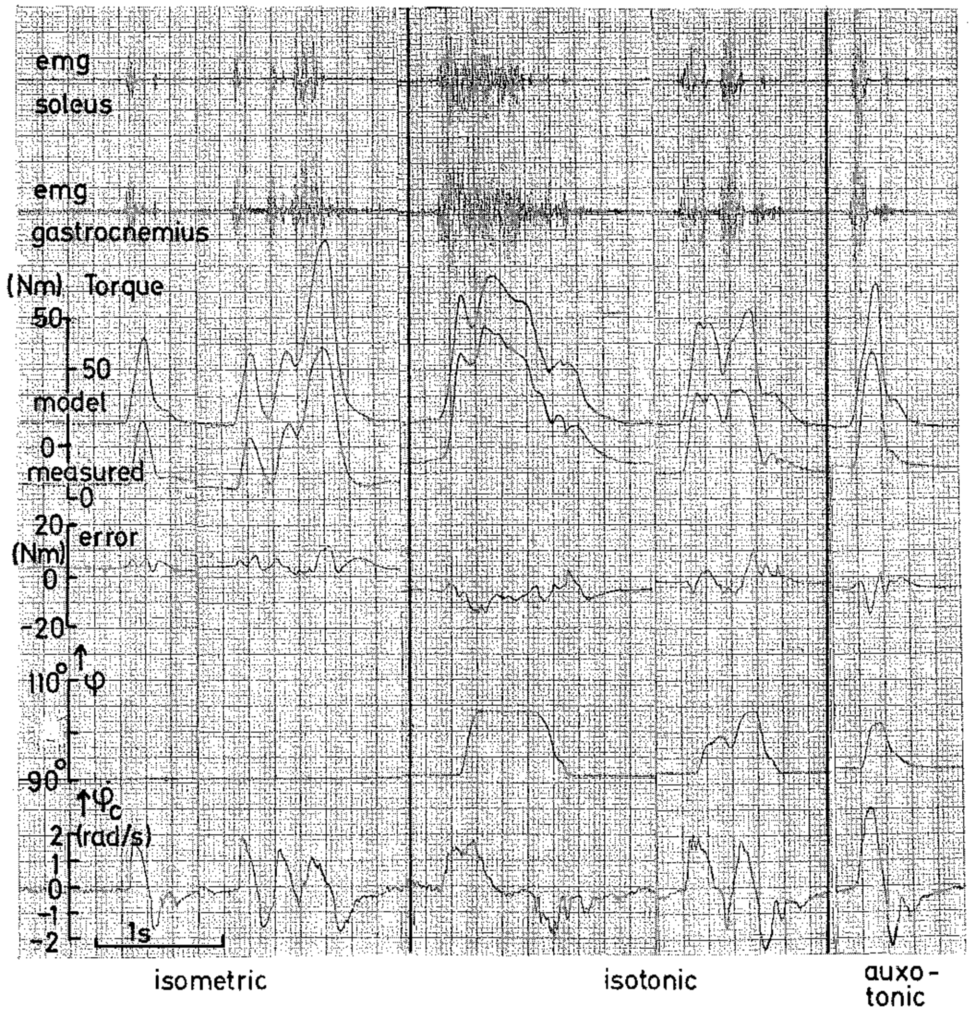

Fig. 3 - Model input, output and measured foot torque.
The values of the model parameters for this subject were:

active state	:	T	=	105 ms
M - φ	:	ϕ_1	=	116°
		ϕ_2	=	88°
PEC	:	M_p	=	10 Nm
Hill	:	b^r	=	1.0 rad/s
		n	=	0.1
		c_1	=	0.2
SEC	:	β	=	40/rad

fastened on a pedal whose axis of rotation was coincident with that of the ankle. The pedal could be fixed, for isometric contractions, or loaded with special Tensator springs or normal coil springs for isotonic and auxotonic contractions respectively.
The results will be discussed. An example can be seen in Fig. 3. The parameters have been found mainly by trial and error. The «error» trace, M(model) - M(measured), indicates very few -if any- systematic errors, only fluctuations due to the stochastic character of the emg remain.

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ELECTROPHYSIOLOGICAL STUDIES OF THE PERONEAL NERVE IN SYSTEMIC LUPUS ERYTHEMATOSUS

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In the last two years, 46 patients with the diagnosis of Systematic Lupus Erythematosus (SLE) were studied by clinical examination, laboratory and EMG procedures in the Hospital Universitario de Caracas. This rather high incidence and the first results of EMG examination were so impressive that it was decided to explore systematically all these patients by the same electrophysiological techniques. The peroneal nerve was chosen because, in our own experience and as referred by others – Dinapoli (1966), Codisch (1971), Cadilhac (1973), Noël (1973), Shahani (1975) – the involvment of this nerve seems to be more frequent, early, and pronounced in systemic pathological condition such as diabetes or chronic renal insufficiency than that of other peripheral nerves.

MATERIAL AND METHODS

The group of patients under study was formed by 44 women and 2 men, with a mean age of 27 ± 4.6 years. In all patients the peroneal nerve was investigated on both sides; 1-3 months after the first examination, 2 patients had a bilateral control and one patient had a unilateral control. In one nerve it was impossible to calculate the motor conduction velocity because of the absence of an evoked muscular potential. A total of 97 explorations of the peroneal nerve were performed.

Motor conduction velocity and distal latency were determined by electrical stimulation of the nerve with superficial electrodes and by monopolar recording needle electrodes, located in the Extensor Digitorum Brevis. The duration of the Evoked Muscular Potentials (EMP) was measured from the onset of the first deflection to the return to the isoelectrical line. The EMP with more than 5 phases were noted as abnormal. With the same needle electrodes we explored the Insertion Activity and the Maximum Voluntary Activity in 34 muscles (EDB).

These findings in patients were compared with those obtained in 23 peroneal nerves of normal subjects.

No significant differences were noted between the right side and the left.

RESULTS

Electroneurographic Findings

	SLE	Normal
Number of Nerves	97	23
Motor Conduction Velocity (m/s)	47.8 ± 6.3 #	57.7 ± 9.3
Distal Latency Time (ms)	4.6 ± 1.2 #	3.4 ± 1.0
Duration of the EMP (ms)	42% > 12	< 12
Number of Phases of the EMP	56% > 5	≤ 5

# $p < 0.001$

Electromyographic Findings

Number of Muscles (EDB)	34
Normal Insertion Activity	8.8%
Hyperirritable Insertion Activity	55.9%
Spontaneous Activity	35.3%
Normal Voluntary Activity	8.8%
Reduced Interference Pattern	17.6%
Discrete Voluntary Activity	70.7%
No Voluntary Activity	2.9%

DISCUSSION

Our electrophysiological findings clearly indicate the presence of a complex defect in the peroneal nerve of the SLE patients, pointing to a peripheral neuropathy, which is of the mixed type (fiber degeneration and demyelination). The interesting fact is that, in these patients, the clinical findings are relatively poor. Only 30% of the cases reported paraesthesiae in the distal lower extremities; in 13% of the patients it was possible to put in evidence objective sensory changes. The muscular strength was slightly reduced in 8.7%. Muscles of the leg were painful to pressure in 13% and the reflex responses were diminished in 22% of the cases.

These rather low percentages of clinical abnormalities contrast very much with the high incidence (more than 90%) of electrophysiological changes encountered in our series, specially at the level of the neuro-muscular system. It seems that the electromyographic abnormalities are more striking than the changes of the nerve conduction because of the vascular or/and inmunological process which affects the very distal part of the peroneal nerve. As an hypothesis, we could suggest that the peculiar involvment of the peroneal territory which occurs in the peripheral neuropathies is due partly to his vertical position in human life and consequently to a special status of blood circulation in the distal part of the territory. This type of peripheral neuropathy, generally sub-clinical as we shown above from our data, is possibly more frequent than it is expected from neurological examination and could be detected in many cases only by electrophysiological procedures similar to those employed in the present study.

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METHODOLOGICAL ASPECTS ON CONTRACTION LEVEL ESTIMATION FROM MYOELECTRIC SIGNALS

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INTRODUCTION

In electromyographic kinesiology there is often a need to use the myoelectric output in order to get a quantitative measure of the force of contraction or to facilitate quantitative interindividual comparisons of the results. To allow this the EMG output must be calibrated or normalized. Such levelling or calibration procedure consists of comparing the myoelectric output, usually the mean absolute amplitude of the myoelectric signal, to the output recorded during one or more defined test contractions. These test contractions can be of different types, e.g.;

1. A maximal contraction.
2. A submaximal contraction with a specific load.
3. Resisting the gravitional forces on the body segment.
4. A series of contractions at different levels of the maximal force of voluntary contraction (MVC).

It is important that the muscle is not in a state of fatigue when the test contraction is performed, and that the muscle does not become fatigued during submaximal test contractions. The latter can be avoided by keeping the recording time of the test contraction short. On the other hand too short a recording time may increase the error of measurement of the EMG level.

The first three test contraction types are based on the assumption of a linear relationship between EMG level and contraction level, whereby the calibration procedure is simply a normalization. If this relationship is exponential a calibration to only one test contraction is questionable.

The present study aimed at comparing some different methods for calibration of myoelectric signals versus contraction level.

MATERIALS AND METHODS

Five male healthy subjects were examined on 20 different occasions. In all cases the myoelectric activity from the common belly of the right biceps brachii muscle was recorded with surface electrodes.

The test contraction studied, 5 s in duration, were 10%, 20%, 30%, 40%, 50%, 60%, 70% and 100% of MVC. The analysis of the myoelectric signals consisted of detecting the time means of the amplitude of the full wave rectified and low pass filtered signals. In this study the mean absolute EMG amplitudes recorded at 10%, 30%, 50%, 70% and 100% of MVC were used for establishing the EMG – force relationship, and those recorded during 20%, 40% and 60% of MVC were used as test signals. Three different ways for calibration were tested (figure 1);

1. Normalization to 100% of MVC assuming a linear relationship between EMG and force from 0-100% of MVC.
2. Normalization to 30% of MVC assuming a linear relationship between EMG and force at least from 0-30% of MVC.
3. Using a calibration curve for the relationship between EMG level and force output obtained by the least squares exponential curve fit of the absolute EMG amplitude levels acheaved from recordings at load levels corresponding to 10%, 30%, 50% and 70% of MVC.

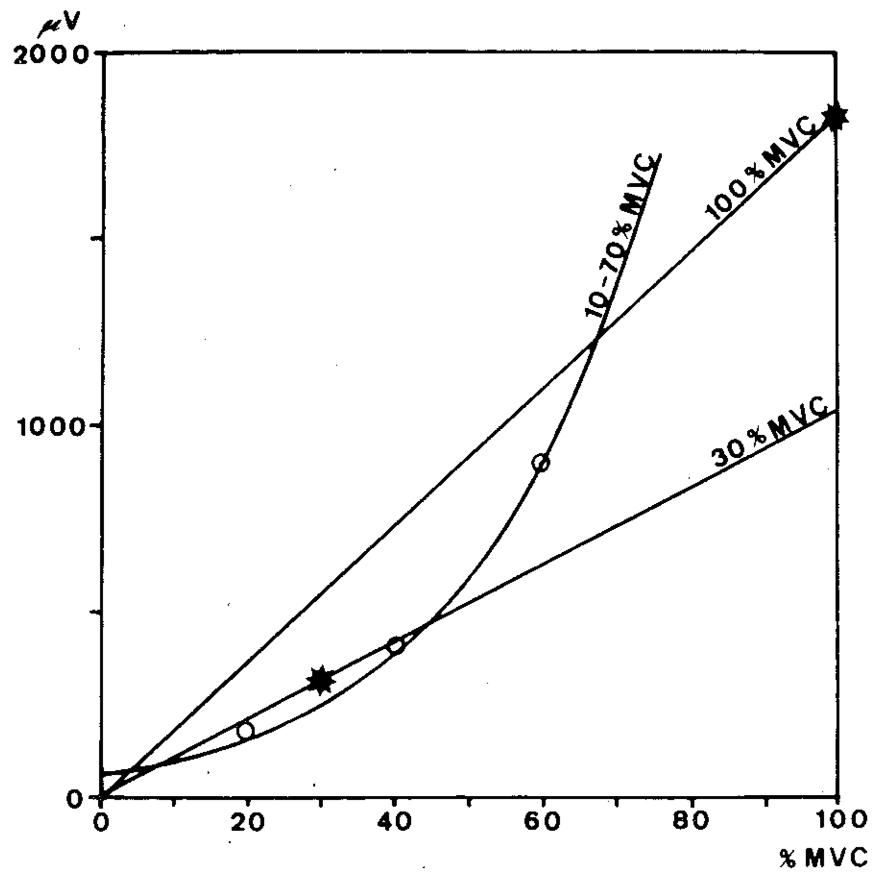

Fig. 1 - Absolute myoelectric level versus force for three different test contractions (0) to (1) the linear function through 100% of MVC, (2) the linear function through 30% of MVC, and (3) to the exponential function obtained from 4 test contractions at different levels of MVC.

RESULTS AND CONCLUSIONS

Calibration by normalization to the absolute EMG amplitude at 100% of MVC proved to be the least exact method for normalization. This method gave very large errors of force estimation at all three levels tested.
Normalization to a test signal at 30% of MVC proved to give almost exact values of the test contractions for 20% and 40% of MVC. Above the latter level this method for calibration gives too large errors of force estimation to be acceptable.
The use of the exponential function obtained from several test contractions proved to give the smallest errors of force estimation of all force levels tested.
Of the three tested methods for contraction level estimation by the use of myoelectric signal amplitude parameters the most exact method thus seems to be to perform a series of test contractions at force levels between 0-70% of MVC and from the results obtained calculate the curvilinear regression function in the least squares sense, y = b exp(mx), for the relationship between EMG and muscular force. Contraction level estimation should then be made against this calibration curve.

MOTOR NEURONE DISEASE OF LEAD POISONING

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Gombault (1) described lead poisoning in humans 1880 and emphasised that the main pathoanatomical feature of lead neuropathy was segmental demyelination. The results of the animal experiments made by Fullerton (2) were in agreement with old Gombault's opinion. The results of investigations and our own experience favour the diagnosis of motor neurone disease against neuritis. In 1907 Wilson (3) emphasised already the similarity between the clinical signs and symptoms of lead poisoning and amyotrophic lateral sclerosis. Aran (4) described in the original paper the disease, which is called by his name, in eleven persons. Three of them were exposed to lead and two had early clinical signs and symptoms of lead poisoning. Campbell and collaborators (5) studied 74 patients with signs and symptoms of amyotrophic lateral sclerosis. They found that 15% of these patients had been exposed to lead and seven were treated by chelating agents.

The authors analysed a large group (N:94) of factory workers and farmers intoxicated by lead. The diagnosis of lead poisoning was done at the Department for Occupational Diseases of Institute for Medical Research and Occupational Health.

In Electromyoneurographic laboratory the patients had complete neurological examination and electromyoneurographic analyses: electromyography of voluntary and spontaneous activity and also efferent and afferent conduction velocity measurements with evoked potentials analyses. The analysis was done by DISA three channel, Medelec two channel, Tektronix 565 Dual Beam osciloscope and Multistim DISA electronic stimulator.

The main results are presented in tables.

A great deal of analysed workers were coming from a factory working with enamel production. Some worked on a bridge burning off the old dye containing lead. They were young people with no sings of alcohol addiction. The most of them had electromyographicaly the loss of motor units and appearance of large motor units action potentials. The farmers drank the wine with lead which was coming in from the enamel of the ceramic pots and jars. Some of them displayed mixed polyneuropathy. The most of them were heavy alcoholics too. In few cases some syndromes were differentiated which were not related to lead poisoning.

Atrophy of the small muscles of the hand and feet was outstanding of neurological examination. The often described wrist and fingers extension

Table 1

GROUP	CLINICAL NEUROLOGICAL AND ELECTROMYONEUROGRAPHIC SIGNS AND SYMPTOMS	NUMBER
1	NORMAL CLINICAL AND ELECTROMYONEUROGRAPHIC FINDINGS	28
2	ONLY NEUROLOGICAL FINDINGS ABNORMAL	4
3	ONLY ELECTROMYOGRAPHIC FINDINGS ABNORMAL, LARGE MOTOR UNIT ACTION POTENTIALS	31
4	CLINICAL NEUROLOGICAL AND ELECTROMYOGRAPHIC FINDINGS ABNORMAL	
	a) POLYNEUROPTHY (MIXED TYPE)	15
	b) SPINAL AMYOTROPHY	10
5	SINGLE SYNDROMES PROBABLY NOT RELATED TO SATURNISMUS	
	a) LIMB GIRDLE MYOPATHY	1
	b) PERONEAL MUSCLE ATROPHY	1
	c) DELIRIUM, POLYNEUROPATHY	1
	d) PSYCHOTIC REACTION, POLYNEUROPATHY	1
	e) ENCEPHALOMENINGOPATHY, ALCOHOLISM	1
	f) CEREBROVASCULAR ISCHEMIC ACCIDENT, MALFORMATION OF LEFT CAROTID ARTERY WITH CONTRALATERAL HEMIPARESIS	1

palsy was found too. Sensory changes were found only in the patients that were heavy alcoholics.

The loss of motor units and appearance of large motor units action potentials was found in spinal amytrophy of amyotrophic lateral sclerosis. The conduction velocity studies were done in median, ulnar and deep peroneal nerve by bipolar needle electrodes. The motor conduction velocity and afferent conduction velocity with direct stimulation of the nerve were within the normal values. Only the afferent conduction velocity based on cutaneous stimulation in mixed polyneuropathy group were on the lower limit level. The second table displays the results of electroneurographic analysis of the median nerve.

Table 2

ELECTROMYOGRAPHIC ANALYSIS OF MEDIAN NERVE

GROUP	CONDUCTION VELOCITY m/s			ms/cm	µV	µV
	MOTOR	AFFERENT	SENSORY	DISTAL LATENCY	NERVE ACTION POTENTIAL	SENSORY ACTION POTENTIAL
1	M=58.6±7.2 N=21	M=65.5±7.3 N=21	M=52.5±5.3 N=16	M=3.8/6.7±1.1/0.9 N=21	M=18.8±12.4 N=21	M=7.8±9.1 N=16
2	M=54.6±6.2 N=4	M=68.8±5.5 N=3	M=59.3 N=3	M=3.6/6.8±0.5/0.8 N=4	M=15.7 N=3	M=5.7 N=3
3	M=61.8±6.7 N=21	M=71.1±8.4 N=21	M=51.6±6.7 N=19	M=3.8/7.3±1.8/1.0 N=21	M=21.2±14.0 N=21	M=8.0±5.7 N=19
4a	M=58.9±6.3 N=10	M=63.9±5.5 N=10	M=48.7±4.0 N=6	M=4.3/7.0±0.3/1.0 N=10	M=22.3±16.1 N=10	M=11.5±11.0 N=6
4b	M=58.9±5.6 N=8	M=62.4±7.1 N=6	M=56.7±6.1 N=4	M=4.0/7.0±0.3/1.0 N=8	M=20.8±10.9 N=6	M=11.5±9.0 N=4

We want to emphasise that the most of examined subjects had the electromyographic finding of motor units loss with large motor units action potentials and some of them had also clinical signs and symptoms of spinal amyotrophy. The conduction velocity studies gave completely normal results. We do not think therefore that the conduction velocity changes could have the value of exposure monitor.

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MUSCLE REINNERVATION AFTER BLOCKADE OF NERVE IMPULSES, A PRELIMINARY REPORT

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There have been reported many investigations about the action-mechanism of drugs inhibiting neuromuscular transmission, and about muscle destruction after local administration of anesthetics. There has been little work done to show if such substances could change the process of muscle reinnervation when injected in the vicinity of the injured nerve. The first experiments in these laboratories have suggested that some local anesthetics delayed reinnervation after axonotmesis. The purpose of the present research was to determine this effect more precisely, and to elucidate if the substances blocking neuromuscular transmission have the same influence.

MATERIAL AND METHOD

The n. ischiadicus was crushed unilaterally in rats. At various intervals after axonotmesis the operated extremity in the experimental group was injected every day with either marcaine, xylocaine or tubocurarine. The operated animals in the control group obtained the same volume of either isotonic saline solution or distilled water in the vicinity of n. ischiadicus as in the experimental groups.
The animals were under observation for four weeks. Electromyograms of the tibial muscle were done and walking ability was observed. In this time period excitability curves were defined for the tibial muscles, and motor conduction velocity of n. ischiadicus was measured.
The animals were sacrificed subsequently and the tibial muscles were removed. The dry weight of muscles was determined, as well as water content and calcium, sodium and potassium ion content. The healthy extremity served in all cases for comparison. For the histological examination m. gastrocnemius was used. The technique with OsO_4 and hematoxyline-eosine was carried out.

RESULTS AND CONCLUSIONS

The results have indicated that the reinnervation of rats which obtained marcaine differed significantly from those in the remaining groups. At

the end of the four weeks in the marcaine group numerous denervation potentials in tibial muscle had occurred, whereas in the other groups they were rare. The loss of dry weight of muscle and water content did not change significantly in all groups, but the calcium ion per cent decreased especially in the marcaine group.
The experiments are not finished yet but the results thus far suggest that the muscle reinnervation is restrained by the lack of nervous impulse transmission. Simultaneously the excitability curve of tibial muscle was shifted upwards and to the right.

TESTS FO MAKING USE OF THE PHANTOM PHENOMENON WHEN USING A PROSTHESIS

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The aim of this investigation was to verify whether the phantom feeling retained in the limb partially amputated can be helpful in teaching patients to use a prosthesis of the upper limb. Similar suppositions were suggested by Sielow, Moser and Eyged but concrete efforts of solving the problem were not untertaken. It was decided to test, thus, whether the appearance of the phantom phenomenon would be affirmed objectively by electromyographic examination.

The hypothesis was suggested that in patients with an active phantom phenomenon an imaginary movement of the amputated limb would evoke a pattern of the movement in the cortex of both hemispheres. This cortical pattern of movement may evoke, in turn, action potentials in muscles which should really cooperate in function of the limb. According to this hypothesis the imaginative movement of the phantom portion of the extremity could evoke a response of the healthy one in form of action potentials possible to be registered with the electromyographic method.

MATERIAL AND METHODS

All examinations were carried out on 30 persons with one upper limb amputated above the elbow and with a phantom feeling. In addition 20 healthy persons for the control group were tested electromyographically. Electromyograms were done with surface electrodes. The myoelectric activity was registered from the following muscles: thenaris, hypothenaris, flexores, extensores dig., biceps brachii and triceps brachii of the healthy limb during the imaginative movement of the amputated limb. We tried to evoke the imaginative movement in all examined patients regardless whether the appearance of a real phantom was confirmed by them or not.

In the control group the examinations were repeated during voluntary movements of the left and right limb alternatively, and the electromyograms were received from the inactive limb.

The sensory excitability curves were defined for all amputees. The curves were marked for the symmetric points for the healthy and for the amputated limbs at six points over the skin in the areas of the shoulder on the anterior and posterior surface, in the central part of the arm-pit on the

anterior and lateral surface of the arm, and on the end-surface of the stump. In six areas of the skin on both limbs marks for light touch examination were done in a similar way.

RESULTS AND CONCLUSIONS

In the control group the voluntary movement of one upper limb has resulted in the appearance of action potentials in the contralateral inactive limb, in varying amounts. A response of bioelectric activity in muscles of the hand was observed to appear rarely, while in the muscle of the upper-arm and forearm it occurred more often (Fig. 1).

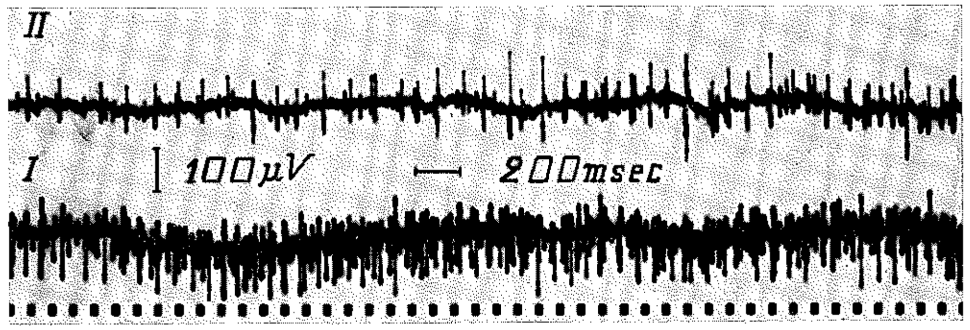

Fig. 1 - Electromyogram of the thenar (II) and hypothenar (I) muscles of the healthy limb, obtained while a movement of the amputated limb was being imagined.

The results obtained from persons with an amputation were found to be different. In this group action potentials were rarely observed during flexion and extension of the phantom limb at the elbow joint, but the response in form of bioelectric activity occurred more during an imagined movement involving carpal joint function and during finger abduction. In healthy controls the response was found to distinctly prevail in the part of the upper-limb, in amputees the response appeared in the distal part.

An imagined movement of the amputated limb was observed to often evoke non-voluntary myoelectric activity in the muscles of the normal hand. A response of this kind has appeared more rarely in the muscles of the forearm and upper-arm. In healthy controls the unvoluntary response during the movement of the contralateral limb was observed to occur most often in the muscles of the upper and forearm and only very seldom in the hand. The results of the electromyographic examinations of amputees were compared with the type of the phantom phenomenon. In nine patients the phantom of the amputated portion was very realistic. The rest demonstrated an unreal phantom.

Sensory excitability of the stump skin was less than in the skin of the normal contralateral limb, especially on the end surface of the stump (Figs. 2, 3, 4).

Excitability curves for amputated limbs were shifted upwards and to the right relative to the contralateral normal limb. The decrement in excitability was repeatedly observed and did not depend on the age of the patient or the kind of amputation. No connection was found between the sensory excitability and the length of period between the amputation and application of the prosthesis.

The results of light touch sensory measurements were more divergent than excitability designations, consequently the over-all differences between the obtained values from both limbs were not significant.

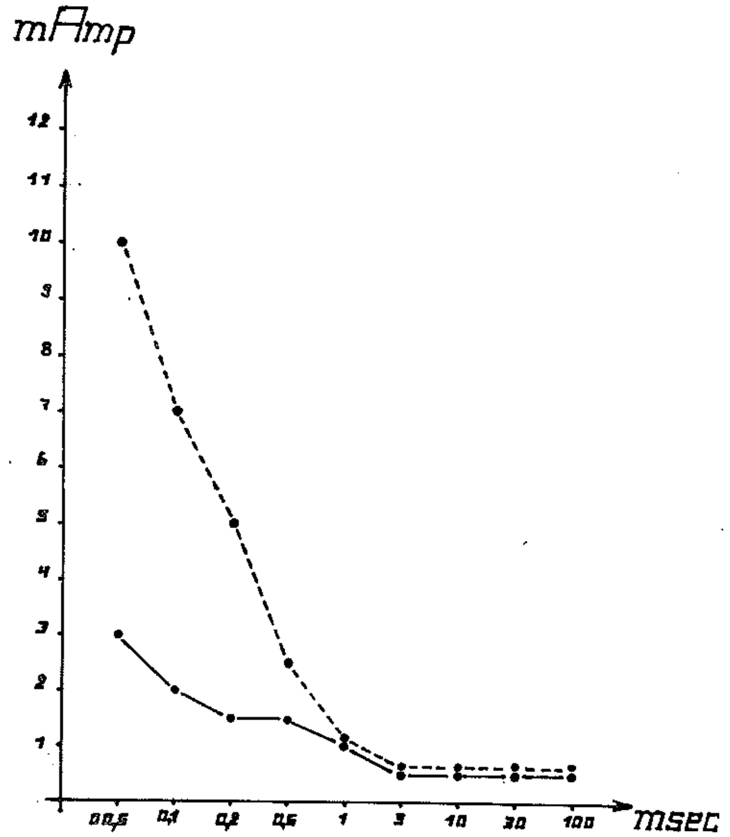

Fig. 2 - A comparison of the sensory excitability curves of the skin of the shoulder, the anterior surface of the amputated limb (dotted line), and of the normal limb (continuous line).

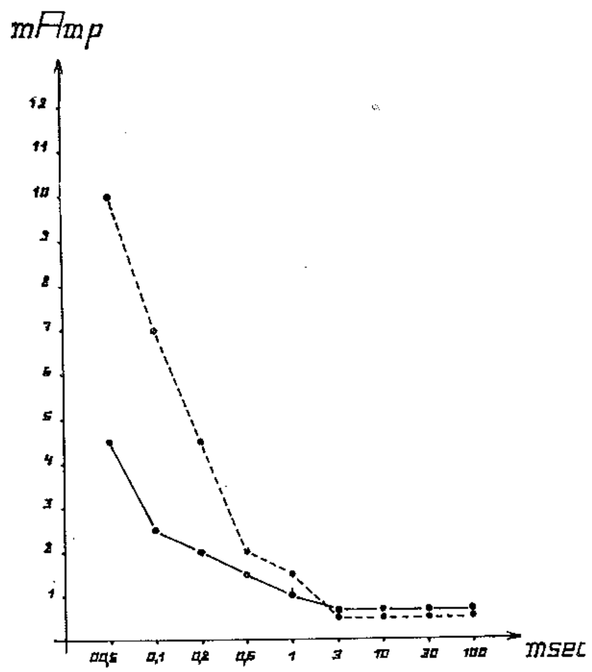

Fig. 3 - A comparison of the sensory excitability curves of the skin of the shoulder, the dorsal surface of the amputated limb (dotted line), and of the normal limb (continuous line).

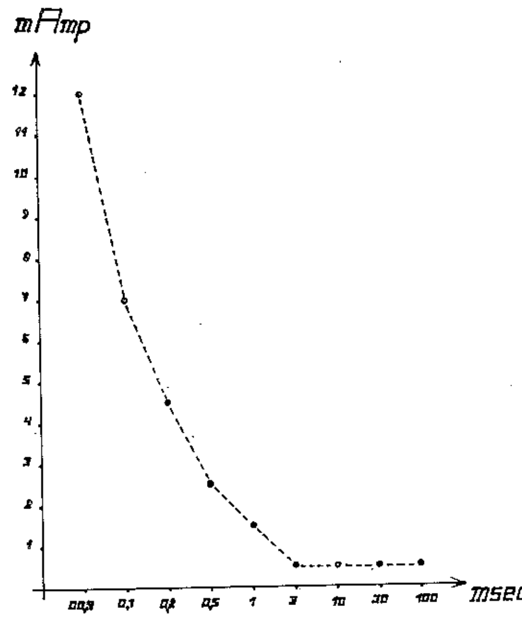

Fig. 4 - The sensory excitability curve of the skin at the end surface of the stump (patient S.J.).

ON-LINE COMPUTER APPLICATION IN CLINICAL QUANTITATIVE ELECTROMYOGRAPHY

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This paper concerns the quantitative on-line analysis of EMG data by means of a mini-computer (ANOPS). The computer was programmed to measure particular parameters of the EMG signal in the form of histograms. These measurements were taken during the electromyographic examination, and were fed, in turn, into the output of a Disa Electromyograph. In weak voluntary muscle contraction three histograms were obtained; giving the distributions of single action potentials in terms of 1) duration, 2) number of phases, and 3) amplitude. In strong voluntary contraction two histograms could be obtained; giving the distributions of 1) amplitudes and 2) intervals (discharge frequencies). The histograms were displayed on the computer screen, and their mean values were used for the automatic quantitative analysis of the EMG data. The obtained results of such analyses were sufficiently accurate, in comparison to more conventional methods, to differentiate between pathological and normal EMGs.

The method has been based on the assumption that the examiner will obtain technically the best possible EMG measurement, that is; the isolated action potentials of a single motor unit which are detected as near the source as possible. This assumption is most important because the computer counts and selects all potentials whose absolute amplitude exceeds 100 µV. This is the only criterion that was used for the inclusion of 1024 single action potentials in one histogram.

In terms of the parameter of potential duration another important issue has been considered, the level of amplitude at which real duration is most accurately estimated. The duration of the potentials was measured at the 20 µV level crossing points. This is in agreement with Lee (1973), and duration measured in this way with the computer from normal subjects agrees within ten per cent with the values reported by Buchthal (1957) and Kopec and Hausmanowa-Petrusewicz (1974).

The second parameter which has been taken into consideration is the number of phases in a specified potential. In certain cases we have observed fairly long polyphasic potentials with some phases not always crossing the zero line. A single phase has been defined, thus, as any inversion of polarity which exceeds 40 µV from the point of inversion. This method of determining the number of phases in the single potential is somewhat different from that of other authors.

The third parameter to be estimated was amplitude. It is known that the use of this parameter has been questioned, mainly because it depends considerably on the conditions of recording. It was found that with the histograms of the amplitudes of single action potentials this parameter became particularly important, e.g.; in differentiating between polyneuro pathy and other horn cell diseases.

The second stage of our work was the processing of the histograms of the EMG data measured during maximal contraction. It would appear that the most suitable method for analysis of such data has been suggested by Willison (1964). The method which has been used in this study was based, thus, on assumptions similar to those of Willison, but our basic unit was variable. In that the analysis of the EMG data has been done on-line and the computer is connected directly to the electromyograph, we have conceptualized the computer as a device reading the photographed EMG records. It was assumed that the accuracy of the amplitude reading by the computer during maximal contraction was one millimeter per one address, and the accuracy of amplitude estimation depends on the measurements procedure. For example, the sensitivity of 1000 µV per centimeter of the EMG amplifier corresponds to the computer accuracy of 100µV per address.

The density of the EMG response during maximal contraction was defined in terms of the time interval between negative peaks. The intervals were measured with an accuracy of 1.28 ms, and could be expressed in Hz as well.

The results of this study have been obtained with 20 control subjects and 125 patients. The results will be discussed, first, in terms of the range of variation within the three parameters for normal muscle, myopathy, peripheral neuropathy and anterior horn cell lesion; and, second, in terms of the correlation of these data with accepted quantitative norms as determined electromyographically.

ELECTROMYOGRAPHIC STUDY OF ENGLISH PRONUNCIATION

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The purpose of this study is to analyze English consonants from the electromyographic point of view, and to make the analyses available for teaching the pronunciation of English as a second language. Sounds should be studied synthetically in terms of articulatory organs, but this study is focused mainly on muscles around the mouth, in order to perform the experiment easily and to make it broadly applicable to teaching English as a foreign language.

METHOD

1. Subjects: i) Two native speakers aged 19 and 37, ii) Three Japanese aged 18, 28, and 30 who are well-trained in English pronunciation, and iii) Twelve Japanese aged between 18 and 39 who are untrained in English pronunciation. The five subjects out of the untrained twelve have been trained to pronounce English correctly, and learning process has been recorded.
2. Muscles tested: Orbicularis oris (OO), Depressor labii inferioris (DLI), Digastricus venter anterior (DVA), Levator labii superioris alaeque nasi (LLSAN), Masseter (M), and Temporalis (T).
3. Recordings: Electromyograms were recorded by an 18-channel multipurpose electroencephalograph and a 6-channel electromagnetic oscillograph. Surface electrodes used were 5 m.m. in diameter. Sounds were put into the microphone, movements of the lips were filmed by 16 m.m. movie cameras (32 f/sec., front and side views), and the vertical movements of the mandible were recorded suitably by the electrogoniometry. The sound waves, the signal pulse corresponding with each frame of the film, and the goniogram were simultaneously recorded with the electromyogram.
4. Words tested: The four-paired English words (right - light, she - see, vase - base, thank - sank) were selected because of the following reasons; 1) they include consonants peculiar to English, and 2) they are very difficult sounds for the Japanese to distinguish. In order to examine various ways of pronunciation, subjects were requested to repeat each word approximately a hundred times.

RESULTS AND DISCUSSION

Figure 1 shows the electromyograms of the native speaker, the well-trained Japanese, and the untrained Japanese, which were recorded during the pronouncing of «right» and «light».

1. [r] and [l] as pronounced by the native speaker:
   There appeared almost no difference of the discharge patterns in DLI and DVA, but a remarkable difference of the discharge pattern was observed in OO. The strong discharge of OO started before [r], continued till the middle part of [r], and later showed a tendency to disappear. Such discharge pattern was not the case with [l]. The discharge patterns of OO, DLI, and DVA which appeared before [r] indicated that the lips were slightly pouted, as was fully observed from the frames of the film. During the pronouncing of [r] and [l], the strong discharge of DLI continued, and the burst of DVA appeared in the latter half.

This indicated that during that period the lower lip was kept spread down, and in the latter half the mandible was lowered.

2. [r] and [l] as pronounced by the well-trained Japanese:
   The discharge patterns showed almost same tendency as in the native speaker, but the strong discharges of DLI and DVA appeared also before the pronouncing of [r], and they showed a co-contraction with OO, which did not appear in the native speaker. This discharge pattern indicated that the well-trained Japanese pouted the lips rather strongly as was observed from the frames of the film. During the pronouncing of [r] and [l], more remarkable discharge of DLI and stronger discharge of DVA appeared than in the case of the native speaker. This indicated that the Japanese subject depressed the lower lip and lowered the mandible rather actively, which might result in remarked discharges in M and T which are antagonists of DVA.

Figure 2 shows the electromyograms of the untrained Japanese which were recorded during the pronouncing of [r] and [l]. Record I was obtained when the subject was not trained to pronounce the two sounds correctly, record II after the subject was trained once a week for three months at a language school, and record III after the subject was trained personally and intensively by an English instructor, emphasizing the electromyographic differences obtained from the native speaker. In record I, both of the electromyograms recorded during the pronouncing of [r] and [l] showed an almost identical pattern. This was similar to what was recorded when the subject pronounced Japanese [ra]. In record II, [r] and [l] the subject pronounced could not be distinguished by the English instructor, and the electromyograms did not yet resemble what was obtained even in the well-trained Japanese. In record III, as for [r], an emphasis for

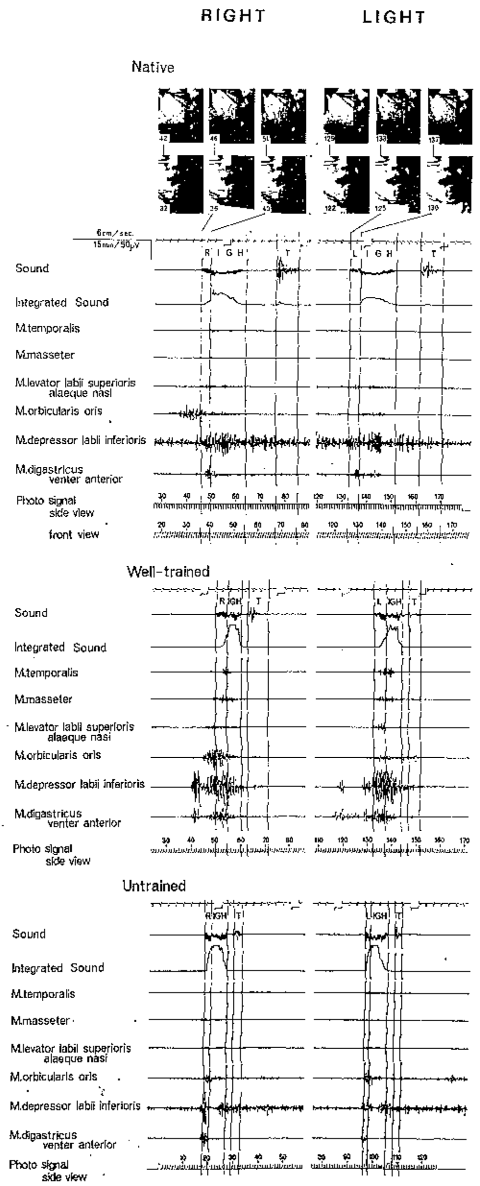

Fig. 1

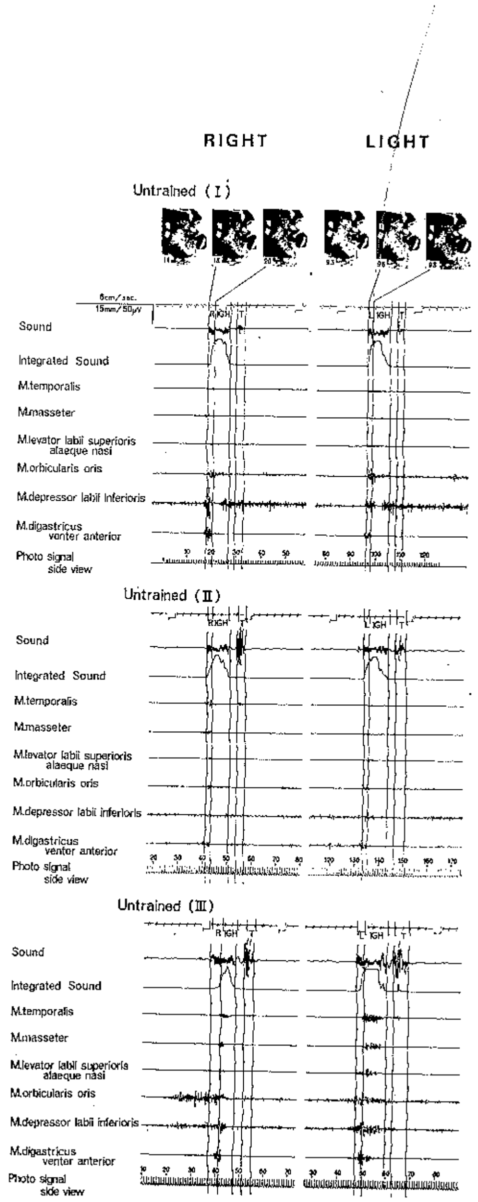

Fig. 2

trainings was put on the lips pouting; while for [l], the spreading down of the lower lip as well as the tongue position was emphasized. The result indicated that the electromyograms resembled what was observed in the native speaker and the well-trained subject. The electromyograms were recorded after [r] and [l] the subject pronounced could be distinguished by the English instructor. In this period, the discharge patterns approached those of the native speaker and the well-trained subject. After [r], the strong and continuous discharge did not appear supposedly because the depressing of the lower lip was not emphasized in teaching.

In the cases of «she - see», «thank - sank», «vase - base», the results obtained were briefly summarized as follows; there appeared the stronger discharges of OO when [ð], [θ], and [v] were pronounced by the native speaker and the well-trained Japanese subject than in the case of [s] and [b]. The strong discharges of DLI were observed when both of the subjects pronounced [ð], [s], [θ], [v], and [b]. As for [ð], [θ], and [v], the lifting up of the lower lip was emphasized; for [s], the depressing of the lower lip; for [b], the closing of the lips. After these trainings, the Englsh instructor concluded that the subject improved much in pronouncing each consonant correctly; while the electromyograms approached those of the native speaker and the well-trained subject.

The results obtained have made it clear that there are, more or less, specific and characteristic ways of using muscles around the mouth in pronouncing English consonants, and that this study would be very uselful for teaching pronunciation of English as a second language.

DETERMINANTS OF MYO-ELECTRIC POWER SPECTRA. PART I: THEORY

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The power spectrum is a highly effective and useful description of signals in general. Since spectral analysis is applicable to repetitive as well as aperiodic events, it is not surprising that this method has a growing position as a tool in the quantification of EMG. Changes in myo-electric power spectra are known to correlate with clinical findings although the details of casuality are in general unknown. In order to interpret such spectra a physico-mathematical model describing the propagation and summation of the myo-electric signals has been developed by Lindström (1970, 1973), Broman (1973), Broman and Lindström (1974), Lindström and Broman (1974) and Lindström and Kadefors (1974). In the model the shape of the power spectrum is determined by a set of physiologically and anatomically well defined parameters. Since the results are expressed in terms of power spectra, the features of the model are closely related to those of EMG spectral measurements. The model is started at the level of single fiber signals and is gradually built up to yield expressions for motor-unit signals and whole-muscle signals as well.

Consider an action potential propagating with velocity v along a cylindrical muscle fiber of radius a surrounded by a volume conductor. Due to the cylindrical geometry, the power spectrum $W_f(\omega)$ of the electric field outside the fiber will have a radial (h) dependence expressed by the following equation:

$$W_f(\omega) = W_a(\omega) K^2_0(\omega h/v)/K^2_0(\omega a/v). \qquad (1)$$

Here, ω is the angular frequency ($\omega = 2\pi$ times the frequency), $W_a(\omega)$ is the power spectrum of the action potential at the fiber surface and $K_0(\ )$ is the modified Bessel function of the second kind and order zero. The quotient between the Bessel functions in eq. (1) can be regarded as a *filtering function of low-pass type with a cutoff frequency which is inversely proportional to the distance* h. The spectral dependence according to eq. (1) can be more or less modified due to non-ideal geometry, e.g. innervation discontinuity and finite fiber length which factors are also accounted for in the model.

Due to the innervation configuration there is a temporal dispersion of standard deviation σ_τ of the individual fiber contributions to the motor unit signal. At low frequencies the original signal coherence is preserved and the spectral contributions add linearly. With increasing frequency the coherence gradually decreases through a transition region to a region at high frequencies where the contributions are added in square. The shape of the power spectrum in the transition region will depend on the distribution of time delays. In the case of a Gaussian distribution the observed motor unit power spectrum is

$$W_{mu}(\omega) = W_f(\omega)\{N + N(N - 1)\exp(-\omega^2\sigma_\tau^2)\}, \qquad (2)$$

where N denotes the number of fibers in the motor unit. *The filtering effect of the summation of dispersed signals is that of a low-pass filter* with cutoff frequency

$$\omega_0 \lessapprox 1/\sigma_\tau \qquad (3a)$$

and an increasing rolloff up to a frequency

$$\omega_1 \lessapprox \omega_0 \sqrt{1n\ N} \qquad (3b)$$

above which the value of this particular filtering function reaches a constant value. A measure of possible variations in the motor unit power spectrum is given by the variance according to

$$\begin{aligned}\mathrm{Var}\{W_{mu}(\omega)\} = W^2_{mu}(\omega)[N(N - 1)(1 - \exp[-\omega^2\sigma_\tau^2])^2\\ (1 + 2[N - 1]\exp[-\omega^2\sigma_\tau^2] + \exp[-2\omega^2\sigma_\tau^2])]. \end{aligned} \qquad (4)$$

It is to be noted that the motor unit signal summation displays large variations in the high-frequency region, where the normalized standard deviation tends towards unity.
The motor unit contributions to the whole-muscle signal are assumed to be uncorrelated and are thus summed as power densities weighted with the distance dependent filtering function. For extramuscular lead-off the following expression for the whole-muscle power spectrum is valid

$$W_{total}(\omega) = A[H(h_{min}) - H(h_{max})]. \qquad (5)$$

Here,

$$A = W_a(\omega)[1 + (N - 1)\exp(-\omega^2\sigma_\tau^2)][(\omega a/v)K_0(\omega a/v)]^{-2}, \qquad (6)$$

and

$$H(h) = \tfrac{1}{2}(\omega h/v)^2[K_1^2(\omega h/v) - K_0^2(\omega h/v)]. \qquad (7)$$

In eq. (5) h_{min} denotes the shortest distance between the electrode and the muscle and the difference $h_{max} - h_{min}$ is the muscle thickness.
For intramuscular lead-off additional parameters such as distances between active motor units and the degree of their overlapping enter into the equations describing the whole-muscle power spectrum.
The electrodes used for lead-off have a pronounced filtering effect on the signal. Calculations on the configuration of *coaxial needle electrodes yield transfer functions of high-pass character* with cutoff frequencies between 100-200 Hz and slopes in the low-frequency region varying between 0 and 6 dB/octave depending on insertion depths and cannula radii.
The transfer function $G(\omega)$ of bipolar surface electrodes lined up in the direction of the muscle fiber is

$$G(\omega) = \sin^2(\omega d/v), \tag{8}$$

where d is half the electrode plate separation. *Bipolar surface electrodes introduce a differentiation at low frequencies and so-called dips* in the power spectrum at certain higher frequencies such that

$$\omega d/v = n \cdot \pi, n = 0, 1, 2, ... \tag{9}$$

Using eq. (9) we are able to calculate the action potential conduction velocity from the position of the dips.
The model outlined here is a tool in the interpretation of EMG findings in general. More specifically, the model has been used for experimental parameter identification yielding quantities of biological significance. The model has also been applied to localized muscle fatigue studies, studies on motor unit synchronization phenomena, optimization of electrodes for EMG control of externally powered prostheses and evaluation of antispastic drugs. Extensions of the model are descriptions of spectral moments of various orders and their relationship with a number of methods for quantitative EMG analysis.

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DETERMINANTS OF MYO-ELECTRIC POWER SPECTRA. PART II: EXPERIMENTAL VERIFICATION

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In this communication attention will be drawn to the identification of the different parameters used for describing the myo-electric power spectrum. First, it will be shown that the model derived in Part I gives an accurate description of the outcome of a set of test experiments. These experiments range from the single fiber to the whole-muscle levels. Second, the use of the model in the interpretation of experimental data will be demonstrated.

Unless otherwise stated all power spectrum and transfer function estimates were obtained by means of the fast Fourier transform implemented on a PDP 15 digital computer. The parameter identification was performed digitally. Values from theoretical expressions well matched by parameter variation to fit experimentally observed data. Parameter values yielding minimum mean squared error – as calculated on the logarithmically scaled data – were accepted as descriptors of the experimental myoelectric signals.

As can be seen in eq. (1) of Part I, the power spectrum at the cell surface enters as a source factor of the power spectrum outside a muscle fiber. There is no way of measuring this source factor. Thus, in order to test the validity of eq. (1) we have to proceed in the following way. Consider the power spectra obtained at two different distances h_1 and h_2, say. The ratio between these two spectra can be written:

$$\frac{W_f(\omega, h_2)}{W_f(\omega, h_1)} = \frac{K_0^2\,(\omega h_2/v)}{K_0^2\,(\omega h_1/v)} \tag{10}$$

Ratios such as this, or rather their square-roots, may be referred to as transfer functions, and describe the filtering of a source at h_1 as measured at h_2. By performing a volume conduction measurement (Stålberg, 1966) it is possible to calculate a series of transfer functions with unknown distances to the fiber, but with a known relation between these distances. In this way one may estimate the actual distances from the recording surfaces to the fiber center and observe the differences between actually observed transfer functions and those predicted by eq. (10). The outcome of one particular experiment is plotted in Fig. 1. In this experiment, the recording

surface closest to the fiber was estimated to be 82 μm from the fiber center, which is a quite reasonable value in such an experimental situation. Furthermore, the average deviation between estimated and predicted transfer function was approximately 2 per cent, indicating a reasonably good fit.

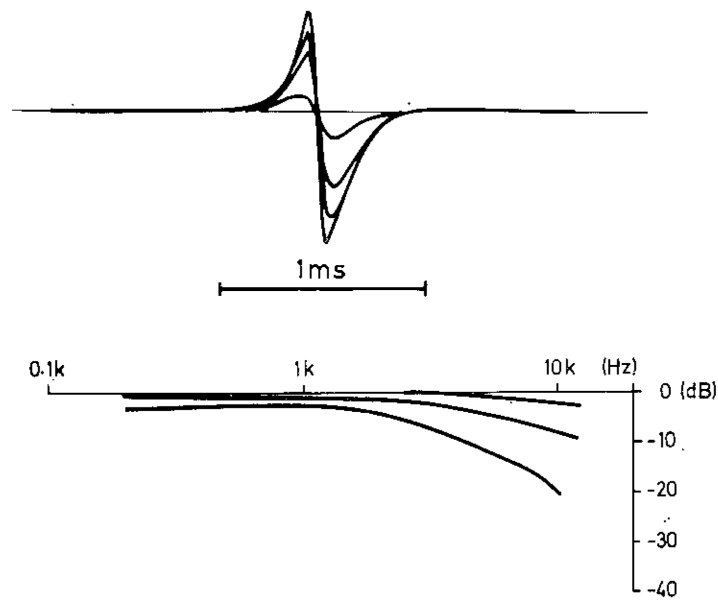

Fig. 1 - Single fiber action potentials from four adjacent surfaces in a multi-electrode needle (above). The three transfer functions according to eq. (10), when the surface closest to the fiber is regarded as the source (below).

The next test on the validity of the present model of EMG power spectra was performed on a single motor unit action potential, derived by means of a differentially coupled pair of surface electrodes. These electrodes introduce dips in the power spectrum, cf. eq. (8) of Part I. For a motor unit displaying small interfiber velocity dispersion, the average conduction velocity can easily be obtained from the location of such spectral dips. In order to identify other parameters of anatomical and physiological significance, it is necessary to have a low-noise recording of the motor unit action potential. This can be accomplished by means of averaging a succession of discharges from a specific unit. An example of such an experiment is plotted in Fig. 2.

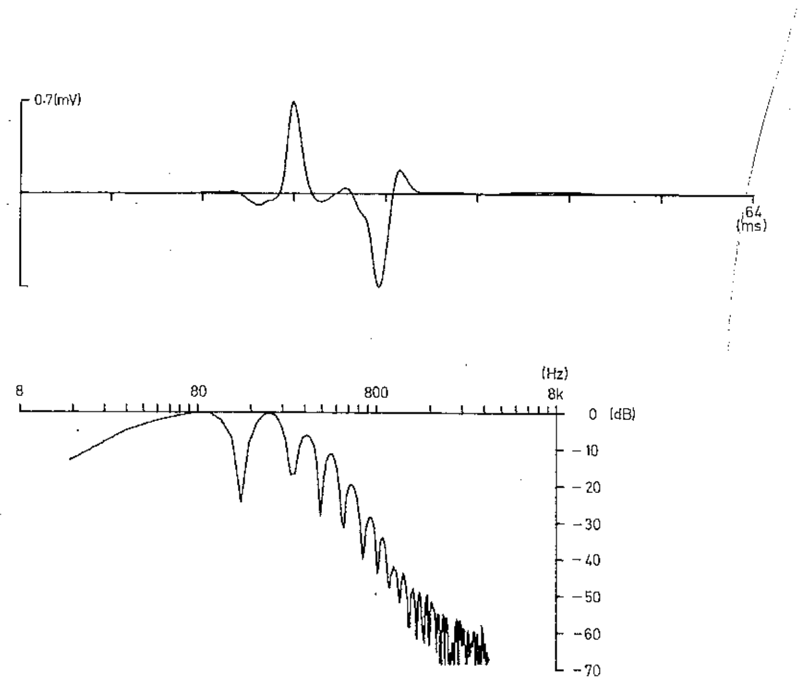

Fig. 2 - Averaged motor unit action potential (above) and its energy spectrum (below).

For one particular motor unit in the long head of the m. biceps brachii the following parameter values were obtained: conduction velocity 3.9 m/s, distance to the skin 1.9 mm, 160 fibers, and an arrival time standard deviation of 0.6 ms. This dispersion of arrival times implies that the distribution of synapses along the muscle fibers can be described by a standard deviation of 2.4 mm. Taking into account that the electrodes were pressed towards the muscle belly, thus lowering the muscle-to-electrode distance, and that the unit was recruited at a relatively low load on the biceps, all the parameter values seems to be in good agreement with data in the literature, obtained by other, more complicated methods.
The final test on the model to be reported here concerns whole-muscle signals. Due to the increased complexity of the model and to experimental difficulties, a rigorous verification is almost impossible. In this test the muscles investigated were m. biceps brachii dx and m. extensor carpi radialis longus dx. The power spectrum analysis was performed by means of an analog octave band analyzer. The data used is actually a subset of the data reported upon by Lindström, Magnusson and Petersén (1974). As the analyzing procedure determined the power in seven frequency bands only, a limited number of parameters can be identified. The parameters

chosen and their estimated values are listed in Table I. Some of these variables take values that seem to be somewhat questionable. For instance the muscle-to-electrode distance seems to be underestimated and the conduction velocity to be overestimated. The reasons for these deviations can be found in the analyzing procedure, octave band analysis yields biased spectra and a poor frequency resolution, in the fact that noise is added to the myoelectric signals, and in the fact that several parameters of the model have similar influence on the power spectrum and thus are very sensitive to imperfections in the estimated spectra. Further refinement of the model and of the analyzing procedure is needed.

Table I

ESTIMATED PARAMETER VALUES FOR WHOLE-MUSCLE SIGNALS
For notations see part I

	m. biceps brachii dx		*m. extensor carpi radialis longus dx*	
	mean	*SEM*	*mean*	*SEM*
h_{min} (mm)	2.5	0.7	1.9	0.5
σ_τ (ms)	0.66	0.05	0.42	0.08
v (m/s)	4.7	0.4	5.4	1.4

The present model for myoelectric power spectra has been used in the interpretation of experimental results in basic physiological terms. As is well known, the power spectrum displays characteristic changes during heavy contractions, static as well as dynamic: the high frequency content decreases and the low frequency content increases. Eqs. (1), (6) and (7) suggest that the argument of the power spectrum is essentially ω/v, and the phenomenon described could be caused by a decrease in conduction velocity v. That this is indeed the fact has been demonstrated in several experiments where dip analysis has been employed. For instance, as reported by Lindström, Magnusson and Petersén (1970), during a 30 seconds heavy isometric contraction the conduction velocity declined exponentially from 3.3 m/s to 1.9 m/s. In another experiment (Broman, 1973), on a single motor unit, it was demonstrated that the conduction velocity after such a contraction recovers exponentially, cf. Fig. 3. In the case studied the time constant was 170 seconds. Another example of the use of dip analysis for conduction velocity estimation is given by Broman, Ladd and Petersén (1974), who determined the average conduction velocity of the action potentials of the fibers of one motor unit with and without administration of an antispastic drug (Dantrolene Sodium). No change in conduction velocity was observed.

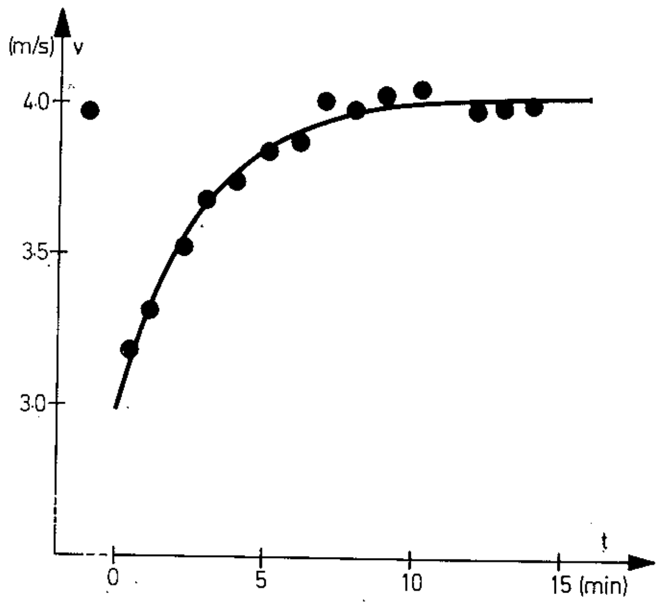

Fig. 3 - Average conduction velocity of fiber action potentials in a motor unit after a heavy isometric contraction as a function of time after the contraction. Note control value before the start of the heavy contraction. Solid line is the least squared error fitted exponential.

In conclusion, the model gives an accurate description of EMG-phenomena on the single fiber and the single motor unit levels. When whole-muscle signals are considered, the model becomes fairly complex. This complexity makes the parameter estimation procedure quite difficult and sensitive to imperfections in the experimental data. Thus, at the present level of knowledge it is not possible either to fully accept the model or to reject it.

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THE CLINICAL COMPUTERIZED EVALUATION OF PARAPARETICS GAIT IMPROVEMENT BY FUNCTIONAL ELECTRONIC BRACE

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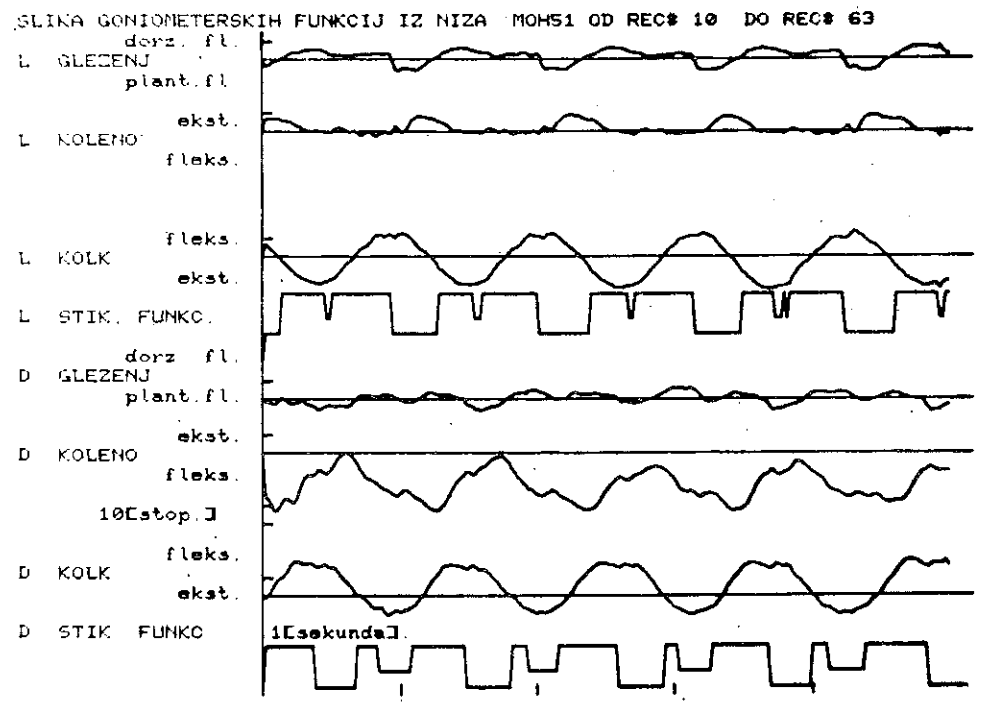
The principle of functional electrical stimulation (FES) found its confirmation in the routine use of functional electronic peroneal brace (FEPB - commercially available stimulator) for improvement of hemiplegics gait.
After some preliminary research studies about spinal cord injury patients and FES (Kralj, Grobelnik 1973), we have had some encouraging results with application of two FEPB-10s for bilateral gait stimulation of patients with paraparetic lower extremities after spinal cord injury. An orthotic device using FES makes use of the natural bone support and the lever system does not need any external weight-bearing or force transferring devices. Having in mind the results of the above-mentioned authors, it was the high power amplification of FES, which was very important.
Our work is new evidence to the hypothesis that spinal cord patients can benefit from FES from both therapeutic and rehabilitation points-of-view. The method of stimulation (ipsilateral, contralateral) for gait improvement and selection of optimal stimulus parameters has become much more sophisticated.
Beside objective recording with a goniometric system there is a great need to have a computerized model for on-line gait evaluation. Clinical laboratory test of the type being developed and applied in our research study should provide improved methods of identifying the optimal walking pattern.
To determine the efficiency of orthotic devices based on FES the computer oriented system has been developed with the following properties: accuracy, reliability, simplicity, low cost, simultaneous measurement of right and left leg movements, and usefulness in clinical routine practice for gait analysis. Different authors have designed and developed measuring equipment which can provide the data necessary for the quantitative gait evaluation. The problem of objective on-line and real time gait analysis programs has not been solved yet.

INSTRUMENTATION

The modular constructed program package for gait analysis has been written in Fortran and Assembler program language. The package is running on the mini-microprogrammable computer system consisting of a Hewlett Packard

2100 S processor with 24 K word core memory, disc HP 7900 with 2.5 M word mass storage, Tektronix 1412 graphic terminal with hard copy unit, digital to analogue converters, HP 2313 analogue to digital subsystem, and standard paper tape units. Teletype is used as a system console. In addition to the computer each experiment was monitored also by recording 6 goniograms of right and left basograms on a multichannel ultraviolet oscillograph.

The hip, knee and ankle angles of both lower extremities in the sagittal plane are recorded with a simple electrogoniometric system (Trnkoczy, Bajd 1975). Heel-on and toe-off contacts were measured by foot switches. Software system program package for gait analysis and diagnosis and for experimental control is operating under a Multiprogramming Real Time Executive (RTE) system.



An example of immediatelly available data sheet results is presented. The difference between unstimulated left (upper lines) and stimulated right lower extremity (lower lines) is obvious.

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DURATIONS BETWEEN WALKING PHASES FROM FILE: GRB31  STEP NO.= 14

                 times in [ms]
   STEP    STEP STANCE STANCE  SWING  SWING  STEP#
    R       L     R.     L.      R.     L.   R.&L.

   1700    1500   2480   2230    720    840      1
   1570    1450   2470   2350    550    740      2
   1640    1560      0      0      0      0      3
   1750    1350   2360   2070    740    780      4
   1500    1480   2270   2240    710    750      5
   1510    1660   2380   2480    790    700      6
-----------------------------------------------

 1611.7 1500 0 2392.0 2274.0  702.0  762.0  MEAN VALUE
   93.3    95.4   77.3  136.3   80.8   46.7  STAND. DEVIAT
WAM3--REALIZED BY VOJKO--NOVEMBER -1975
ENTER OP CODE
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On-line real time results of bazogram are immediatelly recognized and printed by Tektronix terminal display.

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THE BEHAVIOUR OF THE ELBOW FLEXORS DURING FATIGUE

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Some question has been raised in the literature as to what the prime elbow flexing muscles are. This research studied four of the more prominent elbow flexors, the short and long heads of the biceps, the brachialis and the brachioradialis. Six subjects were instrumented with fine wire electrodes in each of these four muscles and the EMG was recorded for later presentation from the four muscles. The subjects were then asked to hold a contraction about the elbow joint at approximately 50% of their voluntary maximum output torque. These subjects were asked to hold this contraction as long as they could and the EMG was monitored over the full length of the experiment. The results from this study show a rather interesting phenomena.

There was no consistancy across subjects as to the prime elbow flexor. In some subjects the prime or major flexor at the start of the experiment was the long head of the biceps, in other subjects it was the brachialis. In no subjects were the brachioradialis nor the short head of the biceps shown to be important elbow flexors at the start of the contraction. In all subjects there was a variation in the level of EMG's from the major elbow flexors. The major pattern showed the prime elbow flexor EMG level would decrease with time and the secondary elbow flexor would then increase to take over its function.

All subjects behaved differently across this experiment but all subjects traded levels of contractions within the muscles such that in the end, all the EMG from all muscles eventually increased. Data is presented on all six subjects to show the relative behavior of the EMG during the fatigue experiment up to the time where the subject could no longer maintain the contraction.

The significant conclusion to be drawn from this study was that when measuring the EMG from any specific muscle and relating this to the torque about the muscle, one must be very careful since the relative contributions of the various muscles to the output from at least the elbow joint changes depending on the physiological conditions of fatigue within that muscle group.

EGM MEASUREMENTS IN THE DETECTION OF THE MECHANISM FOR SUBLUXING PATELLA

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The use of EMG measurements of clinically normal muscle in order to diagnose the mechanism for joint pathology has been little used. In a previous study done with patients exhibiting recurrent subluxing patellae, the motor function associated with subluxing patella has been indicated. In the present study nine chronic patients, four normal athletes suspected of being in jeopardy of incurring a subluxing patella and five normal athletes not suspected of being in jeopardy of subluxing patella were evaluated electromyographically. The EMG was measured from the vastus medialis obliquus, vastus intermedius and vastus lateralis muscles of the quadriceps group. In addition, the reflex time between a patella tendon tap and the occurrence of the EMG in the vasti muscles was recorded and analysed by a PDP.8E off-line computer. The results have shown that in the chronic patients and in the apparently normal athletes who were deemed to be in jeopardy of subluxing patella, the ratio of stretch reflex times of the vastus medialis over the vastus lateralis was 1.09 while this same ratio in those athletes deemed not to be in jeopardy of any subluxing patella was .96. The examination of the EMG's from the three muscles during active flexion and extension in weight-bearing conditions has shown that the possible etiologies for subluxing patella to be either a hyperactive lateralis muscle, and/or a hypoactive vastus medialis obliquus muscle. It has been shown biomechanically that the only force applied to the patella to keep it from subluxing laterally is the force applied by the vastus medialis obliquus. If this muscle either contracts late and/or with not enough force, then either the lateralis and/or the intermedius will pull the patella laterally and cause a subluxation.

The indication from these results is that surgically something should be done to increase the effectiveness of the vastus medialis obliquus as a medial stabilizer of the patella, and in many of the patients a release of the vastus lateralis muscle should be accomplished to reduce the hyperactivity of this muscle and its extreme influence on subluxation of the patella.

KINESIOLOGIC CHANGES OF THE SPASTIC MUSCLE

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The exaggeration of the monosynaptic stretch response has been recognized as one of the most striking and fundamental aspects of spasticity. On the other hand, one is struck during the clinical examination of reflexes in spasticity not only by the exaggerated response of the muscle to stretch but by the briskness of its contraction as well.

In the present study, the contraction time of the quadriceps muscle was examined by compairing the simultaneous electrical and mechanical isometric response of the knee jerk in hemiplegic patients and in normal subjects.

The contraction time of the spastic quadriceps reflex response varied from 50 to 95 msec with a mean of 72 msec significantly shorter ($p > 0.001$) than the mean value of 91 msec (range 70 to 130 msec) obtained in the uninvolved quadriceps and 92 msec (range 80 to 115 msec) in normal subjects. The time to half relaxation was also shorter in the spastic quadriceps, though the difference with normal was less significant ($p > 0.01$).

It is evident from the results that the response of the quadriceps muscle to phasic stretch is not only exaggerated in spaticity due to the larger number of motorneurons that are reflexly activated, but it is also faster due to a shorter contraction-relaxation time. It is reasonable to speculate that the changes in twitch speed of the spastic quadricpes observed in hemiplegic patients may be caused by one or more of those factors known experimentally to affect and modify the speed of contraction of skeletal muscles, such as: relative disuse of the hemiplegic quadriceps secondary to lack or reduction of voluntary contraction; changes in pattern of activity impinging on the motor neuron pool, hyperactivity and predominance of phasic motorneurons.

ELECTROMYOGRAPHIC STUDY OF WALKING PATTERNS BEFORE AND AFTER TOTAL HIP REPLACEMENT

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Various types of total hip replacement operations have been satisfactorily performed for severe hip disorders, and many kinds of objective methods have been elaborated for evaluating the results. However, no electromyographical studis have been found in the literature.

The purpose of this study is to analyse walking patterns before and after the operation using the electromyographic method.

MATERIALS

Thirty cases operated on during the period from 1971 to 1974 were selected for the study. Twelve of the cases were operated bilaterally. The period of follow-up ranged from one year and four months to four years and eleven months, the average being two years and nine months.

All the cases were divided into four groups: the first consisting of seven cases with bilateral rheumatoid hips; the second, four unilateral cases; the third, five cases with bilateral osteoarthritic hips; and the fourth, fourteen unilateral cases.

METHODS

Each patient was instructed to walk at his natural speed on a level floor. The electrical activities of six muscles (rectus femoris, hip adductors, inner hamstrings, gluteus medius and maximus, and sacrospinalis) were relayed from surface electrodes into the amplifier. The signal was recorded on an eight-channel oscillograph.

Foot switches were placed on the plantar surface of the first metatarsal and the heel, and duration of the stance and swing phases was recorded on the same oscillograph. Furthermore, walking patterns were observed with a stroboscopic camera from anterior, posterior, left, and right sides.

RESULTS AND CONCLUSIONS

1. First group (bilateral rheumatoid hips)

Six of the seven cases could not walk at all before the operation. Although

the degree of pain was markedly relieved, the range of the hip motion was hardly improved. Consequently, walking ability was least improved in this group.

The mean walking cycle duration after the operation was 1.47 second (1.07 second in normal controls), suggesting slow walking. The mean percentage of the stance phase was 67.8 on the right and 79.2 on the left sides (60 in normal controls), suggesting increased duration of the double support period.

The hip muscles showed almost normal phasic activities after the operation, but the amount of activities was still greatly reduced compared to that of normal controls. However, activities of the muscles working in the stance phase (rectus femoris and both glutei) were revealed definitely. On the other hand, the sacrospinalis was active even in the stance phase, suggesting that the swing phase of the other side was not long enough.

2. Second group (unilateral rheumatoid hips)

Three of the four cases could not walk and the range of hip motion was markedly limited before the operation. Walking ability was improved only in one case with relatively good motion before the operation.

The mean walking cycle duration after the operation was 1.45 second, and the mean percentage of the stance phase was 69 on the right and 70 on the left sides.

Muscle activities on the operated side decreased generally compared to those on the non-operated side. The rectus femoris was active only in the stance phase, suggesting a stamping gait. Activities of both glutei increased after the operation but those of the sacrospinalis did not change. Furthermore, the appearance of activities of the inner hamstrings at the beginning of the stance phase and of the rectus femoris at the latter part of the swing phase suggested that the patient was walking more vigorously. This fact was found especially in the cases with good motion before the operation.

3. Third group (bilateral osteoarthritic hips)

Although the degree of disability was almost similar to that of rheumatoid hips, walking ability was markedly improved by the opeartion in this group. The mean walking cycle duration changed from 1.60 to 1.32 second. The mean percentage of the stance phase changed from 67 to 60.5 on the right and from 71.5 to 62.7 on the left sides, both of which were almost normal.

Abnormal phasic and quantitative activities of the muscles were generally found on both sides before the operation. The rectus femoris showed activities in both phases, and the swing phase activities suggested its stronger action in extending the knee. Both glutei and the adductors showed a slight decrease of activities, but the sacrospinalis showed increased activities especially in the stance phase. These abnormal findings were converted to almost normal ones by the operation.

4. Fourth group (unilateral osteoarthritic hips)

This group was further divided into three subgroups; A) with normal opposite hips, B) with painful and contracted opposite hips, and C) with ankylosed opposite hips.

Walking ability was generally improved by the operation, especially in subgroup A.

The mean walking cycle duration in subgroup A changed from 1.52 to 1.20 second, and the mean stance phase percentage, from 58.2 to 60 on the operated and from 62.5 to 61 on the normal sides. In subgroup B, the former changed from 1.46 to 1.31 second, and the latter, from 62 to 60.1 on the operated and from 64 to 61.5 on the other sides. As only two cases belonged to subgroup C, those values were not calculated in this group. As far as muscle activities were concerned, both glutei and the adductors in subgroup A showed marked decrease of activities on the operated and a slight increase on the normal sides before the operation. The rectus femoris did not participate in swinging the leg. In subgroup B, although it was difficult to find a distinct pattern, electromyographical findings before the operation were quite abnormal, depending upon the degrees of pain and contracture. In subgroup C, muscle activities before the operation were almost identical to those of the cases with bilateral osteoarthritic hips. Muscle activities in the three subgroups were almost normal in their phasic and quantitative aspects after the operation, suggesting that walking patterns were definitely improved, especially in subgroup A.

PHYSIOLOGICAL AUTOPSY IN THE STUDY OF ARTERIALLY DISEASED MUSCLE

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INTRODUCTION

Chronic ischaemia is known to produce changes in the histological appearance of voluntary muscle. We have now been able to study the concomitant functional changes in amputated legs maintained by perfusion for periods of up to 10 hours. The observations were concerned first with the gross biochemistry and mechanical properties of the muscles, and secondly with the histochemical and mechanical properties of their individual motor units.

METHODS

Legs amputated for reasons of arterial insufficiency were perfused with human blood diluted with acid citrate dextrose usually via the popliteal artery, as previously described by O'Donovan, Rowlerson and Taylor (1976). The effects of perfusion per se were evaluated by comparing biopsies taken during perfusion with those taken during surgery, just prior to amputation. Levels of ATP, CrP, and lactate were estimated as described by Edwards, Jones, Maunder and Batra (1975). Glucose and oxygen uptake, and lactate production were determined from arterio-venous differences. Fatigability of whole muscle was estimated by isometric recording of force at the tendon with direct muscle stimulation at frequencies up to 100/sec. Single motor unit contractions were studied by the technique of controlled intramuscular microstimulation, and force at the tendon averaged with muscle length at the optimum for whole muscle twitch (Stephens and Taylor 1975). A suitable unit once isolated was glycogen depleted by repetetive stimulation (Edstrom and Kugelberg 1968), following this sections were taken for histochemistry.

RESULTS

Pre-amputation biopsies give evidence of severe hypoxia. Thus, in 5 cases the mean levels of ATP and CrP were low (6.22 and 10.52 μmoles/g dry weight respectively), though there was great variation between patients and values were lowest peripherally. Lactate was less variable but consi-

stently high (52.8 μmoles/g dry weight). The relative efficacy of the perfusion was demonstrated by the improvement of these values during the experiment, in a number of cases.
Fig. 1a summarises the metabolic behaviour in one typical experiment. A surprising feature is that despite a reasonably normal rate of oxygen consumption, lactate production – a sign of anaerobic metabolism – is high. Whether this will prove to represent a real abnormality of chronically ischaemic muscle remains to be seen; however, it is clear that oxygen and glucose uptake are responsive to insulin.
In contrast to the remarkably steady biochemical state achieved it is evidence from fig. 1b that whole muscle isometric force of contraction generally declines continuosly. The site of this fatigue is not known, but in single unit stimulation experiments we have observed signs of failure at a point before the contractile element. Notwithstanding some degree of deterioration, the muscles are capable initially of producing considerable force, ranging from 42 kg for medial gastrocnemius-soleus to 2.14 kg for peroneus brevis.
In the unit contraction studies we have examined twitch times and strengths, and these have been briefly reported (O' Donovan, Rowlerson and Taylor 1976). Mean twitch times with ranges were, in medial gastroenemius (122, 100-150 msec), in soleus (92, 64-100msec), in extensor hallucis longus (77, 65-100 msec) and in peroneus longus (75, 38-100 msec). These twitch contraction times are longer than those reported by Buchthal and Schmalbruch (1970), who found mean times of 79 and 74 msec for fibre bundles in gastrocnemius and soleus, though as these authors recognised, fast fibres may have been dominant in determining contraction times in these bundles. It is to be noted that the present results relate to elderly patients (mean age 66.4 years) and a slowing of muscle contraction time with increasing age has been demonstrated in human extensor digitorum brevis muscle by Campbell, McComas and Petito (1973).
In the sample of motor units studied here the fastest units had contraction strengths within the same range as that of the slow units. This finding is contrary to experience with both normal human and animal muscles, where the fastest motor units are well known to have the largest contraction strengths (Milner-Brown, Stein and Yemm 1973, Stephens and Usherwood 1975, Stephens and Stuart 1975, Burke 1967).
In the case of two units, we have correlated histochemical and mechanical properties using the method of glycogen depletion. This involves repeated tetani for up to 2 hours, and the intramuscular microstimulation technique proved quite reliable for this purpose. A fast, small, highly fatigable unit was found to consist of small diameter fibres, rich in myosin ATPase and low in oxidative enzymes; and a slow, non-fatigable unit was found to be

low in myosin ATPase, and rich in oxidative enzymes. These correlations are essentially similar to those found in animal work, where speed of contraction has been shown to be related to ATPase activity (Barany 1967, Burke, Levine, Tsairis and Zajac 1973) and fatigue resistance to oxidative enzyme activity (Edstrom and Kugelberg 1968).
The absence of large fast twitch units in these muscles is reflected in the histochemistry of their parent muscles. Fig. 1c shows an example of a transverse section through medial gastrocnemius, stained for myosin ATPase (pH 9.4). It can be seen that the type II fibres are abnormally small. There is also evidence of re-innervation as shown by fibre-type grouping (Dubowitz and Brooke 1973). The reduction in the diameter of the type II fibres is probably a form of disuse atrophy, since these fibres presumably belong to high threshold fast twitch units, rarely recruited in these patients whose movements are necessarily restricted.
An alternative explanation would be denervation of type II units by degenerative neuropathy, though neuropathies are not known to be selective in this way (Dubowitz and Brooke 1973). Moreover, our results imply that these units are still accessible to microstimulation.
We feel that our artificial perfusion system constitutes a true physiological autopsy, and is capable of providing a detailed physiological interpretation of pathological processes. There is no reason why this approach should not be extended to other fields and so provide a valuable tool in the understanding of disease.

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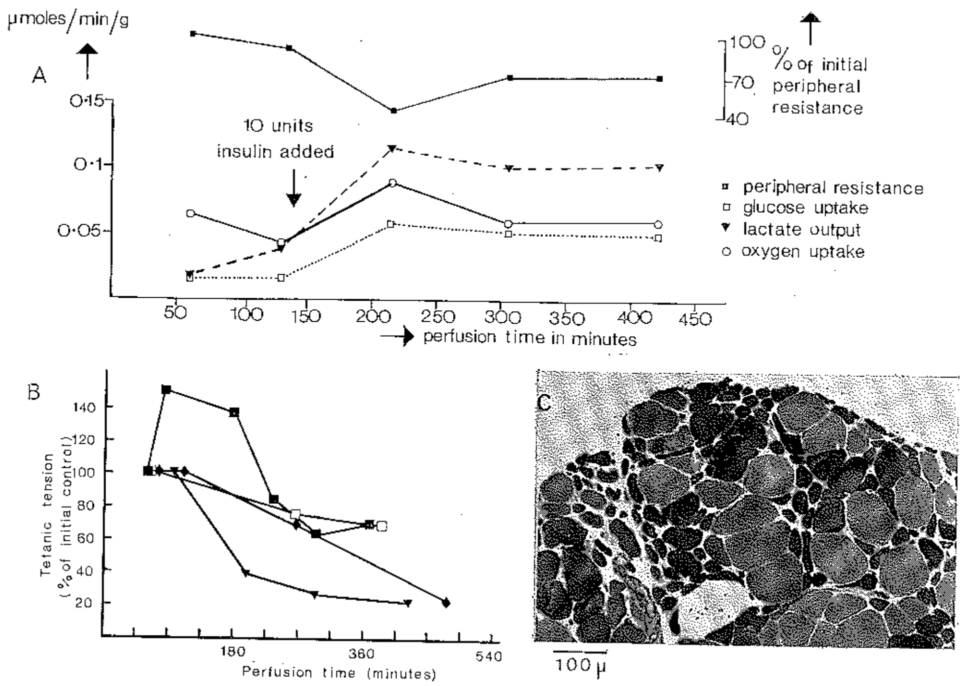

Fig. 1 - a) Relation between oxygen uptake, glucose uptake, lactate production and peripheral resistance, and perfusion time in a typical experiment. b) Relation between tetanic tension (40 pps) and perfusion time for four muscles in four experiments. ■ peroneus brevis, □ ▼ peroneus longus, ◆ gastrocnemius-soleus, c) Transverse section through human medial gastrocnemius muscle stained for myosin ATPase at pH 9.4.

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ELECTROMYOGRAPHY OF BIPEDAL STANCE IN PRIMATES

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Higher primate species have inherent capacity for maintaining bipedal standing posture. Meanwhile, they exhibit a wide variety of locomotor behavior, e.g. arboreal quadrupedalism, brachiation, knuckle walking etc. Comparative assessment of biomechanical features in the bipedality among contemporary primates might be useful for understanding the biological significance of human bipedality. This paper will describe EMG activities in dominant postural muscles during static bipedal standing of the chimpanzee, gibbon, baboon, macaque, and spider monkey. Arrangement of body segments and the line of gravity in the sagittal plane were recorded simiItaneously with the EMGs. Bipedality with the hip and knee joint flexed, a common observation among the species tested, brought about tonic and differed basically from those shown in the human half-rising. Interspecific difference was found in EMG magnitude in the extensors at the hip and the knee, which was interpreted in terms of biomechanical situation in each species as indicating specific locomotor adaptation.

MYOELECTRIC BACK MUSCLE ACTIVITY RELATED TO SPINAL POSTURE AND LOADING

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INTRODUCTION

Heavy loading of the back as a source of low back pain has been of great concern for many years, and considerable effort has been made to reduce the load on the back during labour (Brown, 1972). Evaluation of such measures requires methods to quantify the load on the back that can be used at a working place. To study the usefulnes of quantitative electromyography in that respect, myoelectric signals of back muscles have been recorded when the back was loaded in a series of static spinal postures (Andersson, Örtengren and Herberts, 1976).

METHODS AND MATERIAL

The myoelectric signals were picked up by means of bipolar surface electrodes. The electrodes were placed 3 cm lateral to the midline on both sides of the trunk at T4, T8, L1, L3, and L5 levels. At the level of L5, two additional electrodes were placed 6 cm lateral to the midline. The myoelectric signals were amplified and recorded on magnetic tape, and were subsequently analyzed for power content using linear-law detectors. Full-wave rectified and averaged values of the myoelectric signals were estimated for each recording period and expressed in microvolts (Andersson and Örtengren, 1974).

Fifteen healthy subjects, picked at random among students, participated in the investigations. Their ages ranged from 19 to 35 years with a mean age of 27 years. Mean weight was 72 kg (64 to 80 kg), and mean length was 182 cm (171 to 189 cm).

The subjects were placed in a reference frame in which the position of the trunk and the pelvis could be accurately controlled. Recordings were made in the following static postures: (1) forward flexion to 10, 20, 30, 40, and 50 degrees with a load of 100 N in each hand; (2) forward flexion to 30 degrees with a total load of 0, 100, 200, and 300 N; (3) upright standing and lateral flexion to 20 degrees – to the left and to the right – with a load of 100 N in one hand; (4) rotation of the trunk to 45 degrees, flexion, and

lifting of a load of 100 N one cm from a 30 cm high table. The load was placed at an angle of 45 degrees and 40 and 50 cm from the subject.
The analysis of the data consisted of calculations of mean values, standard errors of the mean, 95 per cent confidence intervals, and regression analysis (Snedecor and Cochran, 1967; Acton, 1959).

RESULTS

The myoelectric activity increased at all levels of the back when the angle of forward flexion increased. Regression analysis combined with analysis of variance showed that a linear relationship between the myoelectric signal amplitude and sine of the angle of flexion was statistically significant at all levels of the back. A linear relationship between the myoelectric signal amplitude and the angle of flexion itself was also statistically significant at all levels of the back. An exemple of the regression calculations is shown in Figure 1. The same type of analysis showed a

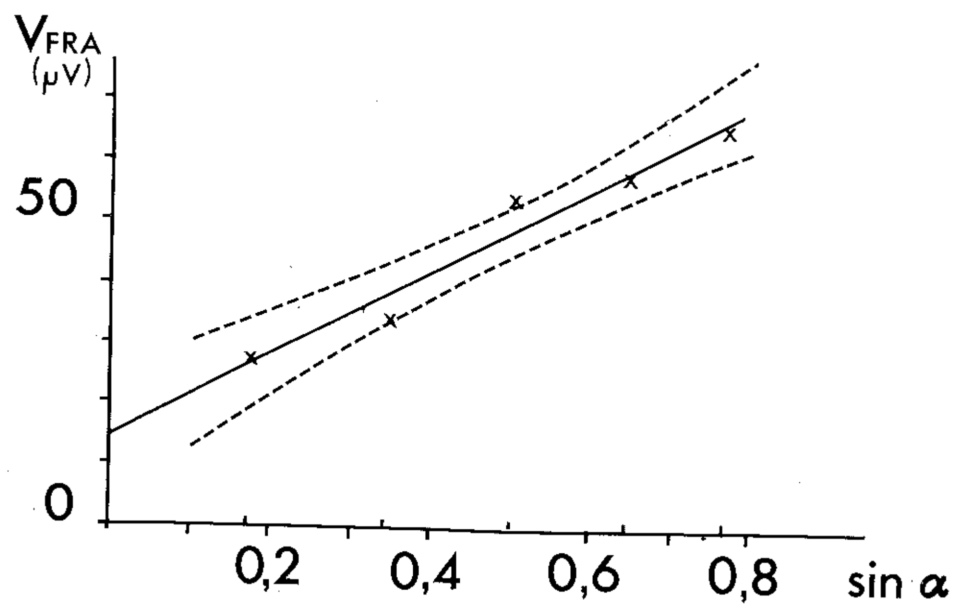

Fig. 1 - Mean values over subjects of myoelectric signal amplitude at L3 level plotted versus sine of the angle of flexion. The equation of the regression line is $V_{FRA} = 14.5 + 69.0 \sin \alpha$ (μV). The dashed lines indicate 95 per cent confidence limits of the regression line.

linear relationship between the myoelectric signal amplitude and the load which was statistically significant for all levels of the back. The regression coefficients for the angle relationship as well as the load relationship were larger in the thoracic region than in the lumbar region. Also the variance of the regression coefficients and the variance around the regression line were larger in the thoracic region for both relationships. Asymmetrical loading of the back always resulted in asymmetric activity patterns. When the recording sites were compared bilaterally, higher activity was found on the side ipsilateral to the load in the thoracic region and on the contralateral side in the lumbar region. The asymmetry was small in upright standing but large when loading was combined with lateral flexion. The largest asymmetry was observed in combination with rotation.

DISCUSSION

From a simple mechanical model of the gross behaviour of the trunk during forward flexion, the myoelectric signal amplitude of the posterior muscles of the back can be expected to vary rather as sine of the angle of forward flexion than as the angle itself.
From the same model, a linear relationship between myoelectric signal amplitude and load variations can be expected when the angle of flexion is unchanged. The interindividual variations in the data are considerable, but the mean values over subjects are close to the corresponding regression lines; analysis of variance shown that the total variation is almost entirely explained by the interindividual variation and the regression. Regression analysis show that the relation between the myoelectric signal amplitude and the angle of forward flexion can with equal probability be linear as according to a sine function. An explanation to this is that the sine function deviates only a few per cent from the best linear fit over the interval 10 to 50 degrees while the residual variation of the data around the regression curve is much larger. An extension of the range of angles beyond 50 degrees would cause difficulties in resolving the functional relationship as the myoelectric activity is known to cease when the trunk is fully flexed (Andersson and Örtengren, 1974).
The data show functional differences between the muscles of the thoracic and the lumbar regions. In the lumbar region the myoelectric response of the muscles follows the load variations closely. In the thoracic region the myoelectric response of the muscles follows the load variations with a larger variability, indication that they are also influenced by other factors. One such factor can be that parts of the trapezius muscles respond to arm and shoulder movements.

CONCLUSIONS

Quantitative electromyografy correlates well to posture and load in laboratory investigations. Further studies during work are needed to test the method under realistic conditions. Such studies are at present performed at the Volvo assembly line.

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ELECTROCORTICOGRAPHIC STUDIES OF THE INPUT SYSTEMS IN THE MOTORSENSORY CORTEX IN MAN

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Electrical stimulation of the precentral cortex in unanaesthetised man has been shown to evoke sensations at the periphery (Penfield and Boldrey 1937). Later experiments showed that the development of such sensations was indipendent of the integrity of the post-central cortex (Penfield and Jasper 1954). Thus the case for a motorsensory cortex (Ms1) was made (Woolsey 1958).

In 1970 Goldring et al., and in 1972 Goldring and Ratcheson, using evoked potential techniques and single unit recordings, showed that local potentials developed in the human precentral cortex following peripheral nerve stimulation. The role of the postcentral cortex in the development of such potentials was not clear from these recordings. However, the work of Broughton (1969), later verified by Allison and Goff (1974) and Papakostopoulos and Crow (1974) showed that the first deflection was of opposite polarity in pre- and post-central areas. Both Broughton and Goff and Allison interpret this as an electrical expression of a dipole generator buried in the central sulcus.

In the present study the nature of the pre- and post-central potentials has been investigated in order to clarify their dependence or independence. The methods include tests additional to electrical stimulation of the peripheral nerves and some of them depend on the active participation of the subject.

The recordings have been obtained from 10 alert co-operative neurologically normal subjects with chronically implanted cortical gold electrodes for therapeutic purposes (Crow 1973, Crow et al. 1961). Data collection and analysis was carried out with a multichannel recording system and a PDP-12 computer. Simultaneous recordings of 16 electrodes using aplifiers having a time constant of 6 sec and upper frequency cut at 2000 Hz were obtained (Papakostopoulos et al. 1974).

Electrical stimulation of the medial nerve at the wrist, contralateral to the electrodes, evoked potentials at both pre- and post-central cortex. The post-central responses started with a negative and the pre-central with a positive deflection. The latency of these deflections was consistently different for the two areas, the post-central deflection being earlier by about 1.5 msec (Fig. 1).

Similar differences in waveform configuration and latencies were observed in potentials elicited in the two areas with passive or externally paced

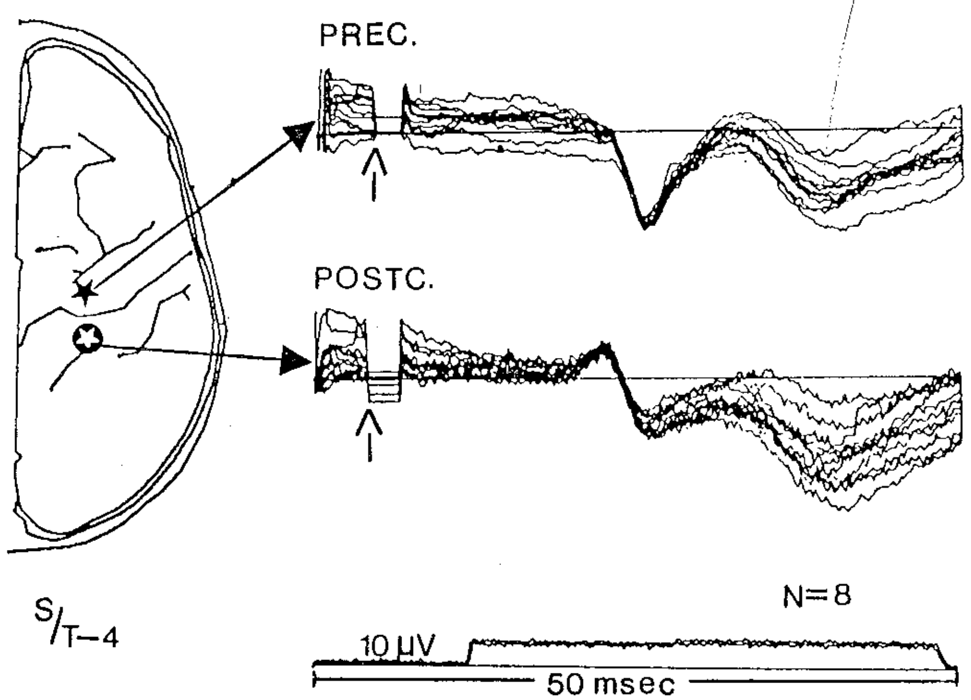

Fig. 1 - Superimposed, averaged evoked potentials from precentral (PREC) and postcentral (POSTC) right cortical areas. Electrical stimulus (open arrow) at the left medial nerve at the wrist. Note the different polarity and latency for the first component. Cortical electrodes all figures referred to average at 60 electrodes in frontal lobe.

displacement of the contralateral index finger (fig. 2). In view of the time differences it is difficult to see how such potentials could be explained by the hypothesis of a common generator in the pre- and post-central areas. However, the data could be explained by accepting that there are different afferent pathways for the two areas, as suggested by the original stimulation of the pre-central area. The recent data of Murphy et al. (1975), in animals, supports this hypothesis by showing that the input from the VPL to the pre-central cortex is independent of the classically accepted VPL pathway for the post-central cortex.

The independence of pre- and post-central electrical activity has been shown by two additional tests demanding the active co-operation of the subject. In the first the electrical activities of the two areas associated with self paced index finger displacement were compared with those following passive or externally paced displacement. In the second test electrical stimuli were applied at the medial nerve with the index finger at rest and during movement. On both occasions the post-central re-

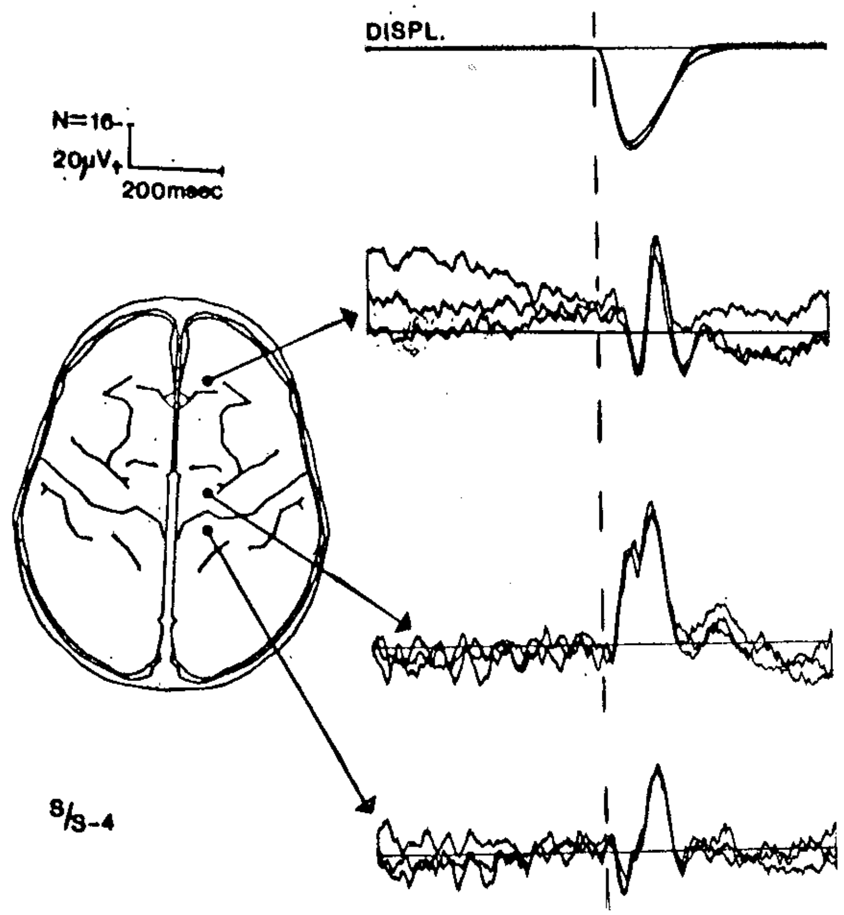

Fig. 2 - Superimposed, averaged evoked potentials from prefrontal, precentral and post-central right cortical areas to the left index finger externally paced displacement (DISPL). Note the consistency in waveform for the same area and the differences in different areas.

sponses were abolished or severely diminished while the pre-central response retained the amplitude for the majority of its components.

Figs. 3a and 3b show the pre- and post-central somatosensory responses, simultaneously recorded, with the electrical stimulus applied at various times after the beginning of the index finger movement. The upper traces show the potentials of each area with the finger at rest. The second traces show the potentials elicited by the stimulus when applied 50 msec after the initiation of the movement. It can be seen that the post-central response has largely diminished while the pre-central response has retained the amplitude for the majority of its components. At 400 msec after

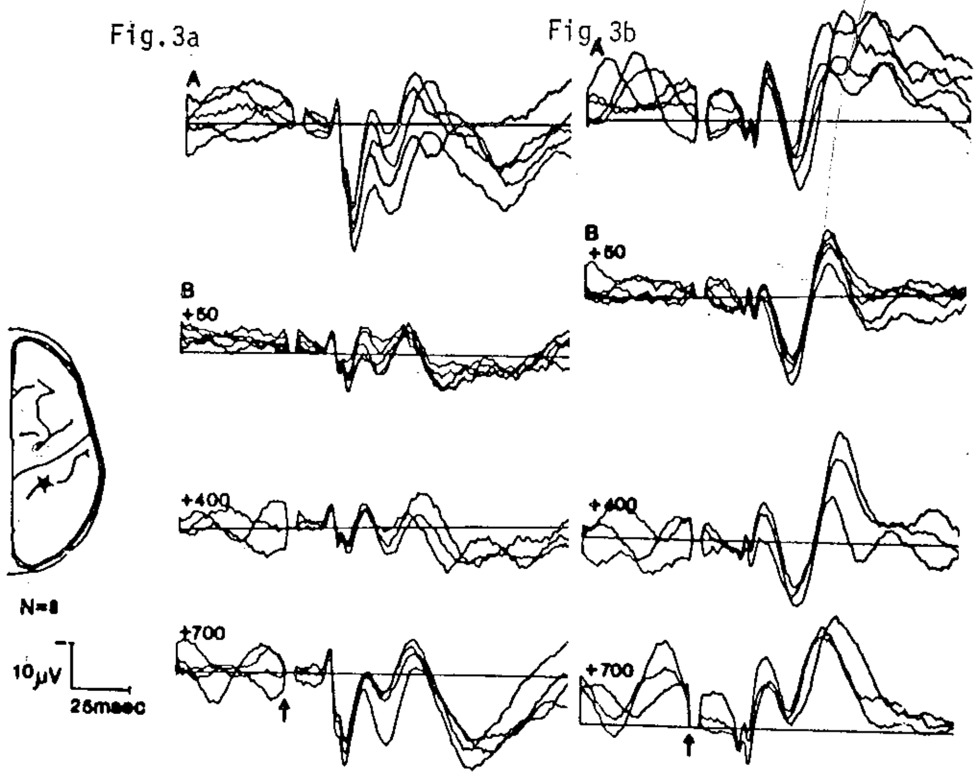

Fig. 3 - Superimposed averaged potentials from post central (Fig. 3a) and pre central (Fig. 3b) cortical areas evoked by stimulation (arrowed) of the left medial nerve with the index finger at rest (upper traces) and at 50, 400 and 700 msecs (lower traces) after the initiation of a self paced finger movement.

the initiation of the movement the pre-central response has almost fully recovered while the post-central response is still as 40 msec after movement.

Thus, because of the different waveforms, latencies and reaction to behavioral acts in the two areas, we conclude that the primary sensory input to pre-central cortex is independent from that to the post-central cortex and we suggest that these inputs are differentially gated by the conticofugal systems responsible for the initiation and execution of motor action.

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THE HUMAN ANAL REFLEX

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In addition to its general physiological interest, the anal reflex is of clinical value, especially in tests for lesions of the conus and cauda since it is the lowest, or at least one of the lowest, segmental spinal reflex, and it is generally assessed in neuro-urological examinations. Furthermore, the anal sphincter is the most easily accessible of the striated muscles of the pelvis floor, for example, for electromyographic recording, and with some limitation it can also reflect the activity of other parts of the pelvic floor.

Nevertheless, since Rossolimo in 1891 gave his brilliant description of the reflex only a few experimental investigations have been published. The need for more systematic knowledge of the anal reflex forced us to extend our routine examination, consisting of reflex elecitation by pinpricks in the perianal skin and recording of the contraction by electromyography from the external anal sphincter. In quantitative investigations of the reflex, including latency, electrical stimulation was necessary, but owing to the short distance between the stimulus electrode and the recording electrode the amplifier was saturated for several hundred milliseconds, during which time the recording of potentials therefore became difficult. This problem was, however, solved by incorporating a relay in the amplifier. This relay is activated just before the stimulation and can cut off the amplifier for a period of from 2-40 msec as desired. A cutting-off period of 20 msec is normally used, which renders the methods highly satisfactory.

The anal reflex can be elicited by a single stimulus or by a train of stimuli of a duration of up to 50 msec without changing the response qualitatively. We now routinely use a train of stimuli of a total duration of 9 msec consisting of square-wave pulses of a duration of 1 msec, separated by 1 msec. The electrical stimulation is applied by a Disa Ministim through a Grass constant-current unit, and the EMG is obtained by an intramuscular bipolar electrode (Disa, type 13K57) introduced from the perineum just lateral to the anocutaneous junction into the external part of the anal sphincter.

The latent period decreases with increasing stimulus intensity from about 400 msec to a minimum of 50-70 msec, which is found in most cases. On unilateral stimulation, EMG responses can be picked up from both sides of the anal sphincter, but with a preponderance on the stimulated

side. On elecitation of the anal reflex by perianal stimulation, a simultaneous reaction is observed in the external urethral sphincter. The electromyographic response of the reflex is comples ('flexor-reflex-like') and is increased with increasing stimulation.

The relationship between the flexor reflex of the leg and the anal reflex has previously been demonstrated in our laboratory, and our continued studies have shown that reflex activity of the anal sphincter can be elicited by stimuli ordinarily used for the flexor reflex, such as mechanical or electrical stimulation of the planta or by electrical stimulation of the posterior tibial nerve behind the medial malleolus. Both healthy individuals and patients have been studied, and the anal sphincter reflex could be elicited in all of them. On electrical stimulation distally on the leg, the anal response was found to have the lowest threshold – a finding which may be of interest in relation to the observation which Riddoch made as early as 1918, namely that the anal reflexes were the first to return after spinal shock – 'if ever disappeared'. This observation was later confirmed by others, and we have also been able to confirm it.

By increasing the intensity of the stimulation, reflex responses can also be picked up from the muscles of the leg participating in the flexor reflex, in our investigation usually represented by the tibialis anterior muscle and the biceps femoris. It is frequently seen that – at a certain stimulus intensity above the threshold – an early component may appear both in flexor muscles and in the anal sphincter.

At low stimulation, the latency of the anal reflex elicited distally on the leg ranges from 300 to 400 msec, but at increasing stimulation the latent period is usually reduced to 60-90 msec, which is approximately the same as the minimum latency of the flexor reflex from the tibialis anterior muscle. At this stimulation intensity, the rhythmic activity which is known from the flexor reflex is also seen in the response of the anal sphincter.

The possibility of eliciting the reflex reaction in the anal sphincter in two different ways, namely by stimulation in the perianal region and distally on the leg, gives an opportunity to perform a closer analysis of defect in the anal reflexes, since the two types or reflexes have the efferent, but not the afferent pathway in common.

AN ELECTROMYOGRAPHIC ANALYSIS OF MUSCULAR ACTIVITY CORRELATED WITH THE BIOMECHANICAL ASPECTS OF HUMAN LOCOMOTION

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A significant improvement in therapeutic and diagnostic procedures for many physically disabled subjects has resulted in a better knowledge of the locomotor system and its functioning. As system theory suggests, the first step of fundamental importance for a rigorous analysis of a system consists in the determination and in the study of the correlations of all the significant variables available.

In this respect the kinematics and EMG activity have been largely investigated by Basmajian and MacConaill (1969) and at the University of California (1953). Though some work has concerned the analysis of forces acting during deambulation, despite its evident relevance, a complete description of the dynamics of movement including inertial forces and torques on joints have been carried out in only a few cases (Bresler and Berry, 1951; Paul, 1972).

The present work has evaluated the EMG activity of most of the muscles involved in human locomotion correlated with the mechanical moments required at joints. As this analysis was performed subject by subject during the same test in the same experimental conditions, the correlations between these two sources of variables assume particular significance. The electromyograms of the muscles were recorded by using surface electrodes; the experimental and analytical procedure is described in complete form by Cappozzo et al. (1975, 1976).

A mathematical model of the lower limbs has been determined. As illustrated in the simple scheme of Fig. 1, the inputs of this model are the variables easy to measure by conventional instrumentation (kinematics, ground reactions), and the outputs are a set of variables which could not be measured. They are the muscular torques on joints and the lengths of various muscles computed as functions of time during a complete double step.

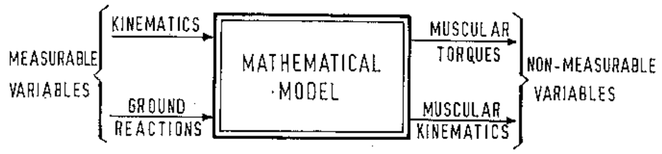

Fig. 1 - Scheme of the mathematical procedure used.

Although the same considerations can be drawn from all the seven subjects examined, the data of only one subject have been presented. He is a male normally walking at a speed of 104 steps per minute. The data have been divided joint by joint and the EMG records are directly reported without any filtering or further elaboration. In Fig. 2 the patterns of the angular displacement, torque and the EMG activity of muscles acting on the hip joint in the plane of progression have been illustrated.

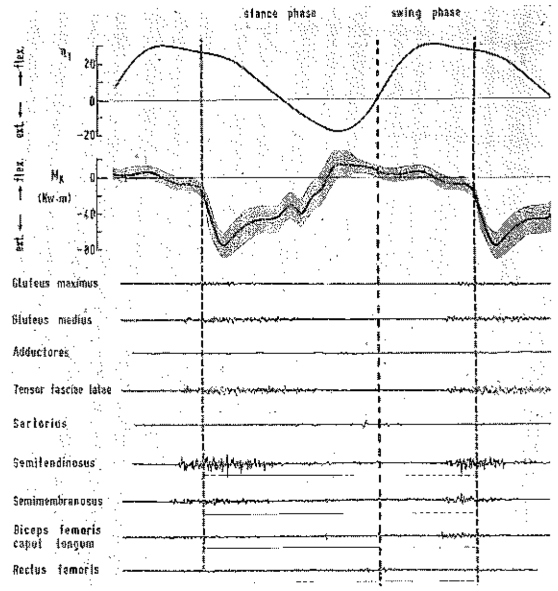

Fig. 2 - From the top downward: angular displacement, mechanical moment, EMG records of muscles acting on the hip joint. The continuous line below the EMG traces means a muscle shortening; the interrupted line an almost isometric condition.

The moment present a clear extensor phase beginning at the last part of the swing phase. It is a result of the activity of glutei, semitendinosus, semimembranosus, biceps femoris caput lungum muscles. The latter three exert at the same time the flexor action required at the knee joint as shown in Fig. 3, where the same data are reported referring to the knee joint. In turn the last part of this flexor torque required at the knee joint, when the activity of the hamstings disappears, is exerted by the gastrocnemius muscular activity. This simultaneously and together with the soleus muscle provides the large torque on the ankle joint at the middle part of the stance phase (see Fig. 4).

The flexor torque required on the hip joint at the end of the stance phase is realized by the iliacus muscular activity. In this pattern the EMG record

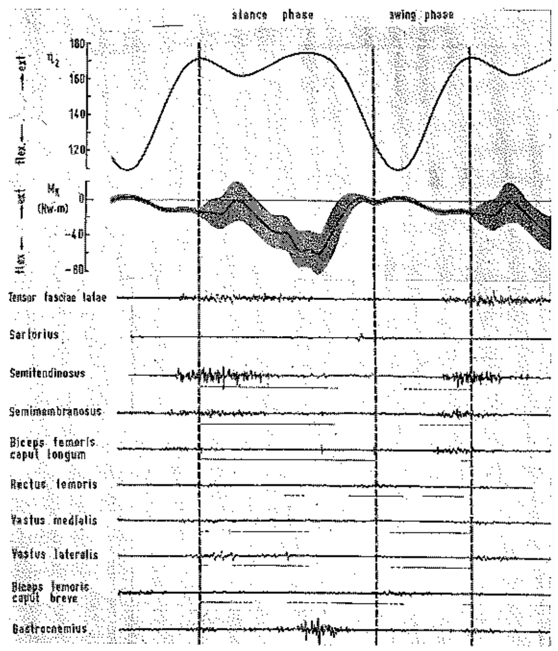

Fig. 3 - The same concerning the knee joint.

from iliacus is lacking because we found it difficult to obtain without causing pain or limiting the freedom of movement of the subject. The records obtained at the University of California (1953) strictly agree, however, with our findings because they show its activity to be right at

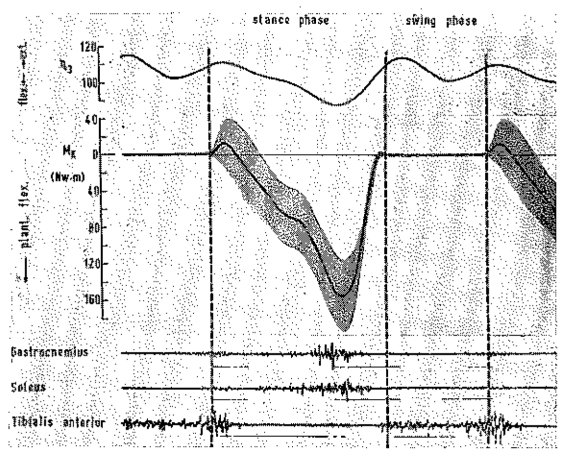

Fig. 4 - The same concerning the ankle joint.

the end of the stance phase, when the hip torque in every subject is flexor and no other muscle can support it.

Around the toe-off a slight activity of the rectus femoris muscle was evident also. It contributed to the flexor moment at the hip joint and, at the same time, it supported the little extensor torque required in that phase on the knee joint. The tibialis anterior muscle has presented a burst of activity during the heel-strike providing the torque required during the yielding phase at the ankle joint. Moreover it provides a slight torque in order to sustain the foot during the swing phase. The vasti muscle group showed a little activity around the heel-strike when an extensor torque, particularly slight in this subject, was required on the knee joint.

The importance of considering the dynamic aspects of movement becomes clearly evident in the correlative analisis of these data. In fact, and despite the similar kinematics, the muscular torque time courses of the other subjects have presented significant differences causing a different temporal sequence of muscle activation. Moreover, the rigorous analysis of the synchronization of muscle activation actually exerted by neural control, in relation to other dynamic and kinematic variables, provides interesting suggestions.

At the natural walking speed, the biarticular muscles were activated only when their action was required by both joints which they were acting on. A simultaneous action of antagonists was never present, apart from some slight activity during the heel-strike when the problem of placing the foot properly assumed particular importance. This suggests, according to Basmajian's observations (1969), that the control system realizes the movement in an optimal way, that is, besides maintaining stable equilibrium and synchronizing various body segments, it minimizes the energy expenditure related to the global muscle activation.
From these results and considerations it clearly appears that the muscular activity during locomotion cannot be reduced to an alternate activation of flexors and extensors at various joints during the different phases of the stride. In contrast, in every instant of time the motoneuron commands acting on muscles must be rigorously modulated in relation to the state of muscles themselves, to the complex interaction between them and to movement dynamics.

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CALCIUM DEPENDENT MYOTONIA: A NEW MATHEMATICAL MODEL

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The reduced C_{Cl} across muscle membrane was claimed to be responsible for human (Bryant and Morales-Aguilera, 1971), goat (Lipicky and Bryant, 1966) and experimental induced by carboxilic acids (Rüdel and Senges, 1972) myotonia. Starting from this hypothesis and using the Adrian's model of sarcolemma, Bretag (1973) developed a mathematical algorithm able to simulate myotonic bioelectric activity.
Certain recent data however support the postulate of Ca-transport and pCa homeostasis disturbance as possible factors inducing myotonia (Radu, Radu and Blücher, 1970; Gillis and Maes, 1973). In fact, even in the Bryant's model, the myotonic behaviour of the membrane could be controlled by increasing $Ca_{(o)}$.
A mathematical model was developed in order to test the hypothetic role in myotonia of the modified Ca transport parameters across sarcolemma. The program was based on the quantitative description of the membrane (Hodgkin and Huxley, 1952), more recently revised by Adrian and Peachey (1973). The operational procedure consist in resolving by normal numerical method on a 1800 IBM Computer, the equational system developed from the equation

$$I_K + I_{Na} + I_{Ca} + I_L + {}_{Cm} + I_S = 0 \qquad (1)$$

in which was innovated the introduction of I_{Ca}. The parameters I_K, I_{Na} and I_S, were the same as in the original work.
The variable parameters tested were:

1. G_{Na} and G_{Ca}, assuming that they follow the same laws and considering $G_{Ca}=16$ mS. cm^{-2} for a $Ca_{(o)}/Ca_{(i)}=10^4$ coresponding $E_{Ca}=+103$ mV (normal conditions).

2. The slope of the stright line assumed to pass through the point pCa=7 and defined in terms of time constant τ_u of the exponential time evolution of $Ca_{(i)}$ supposing I_{Ca}=constant.

3. G_L was carefully computed in the ranges of 0,126-0,278 mS. cm^{-2}, in order to maintain V_r at the physiological level for each set of other parameters.

In order to investigate the electrochemical coupling, the model monitored the pCa dynamics and some other quantitative parameters involved in G_{Ca} dynamics. For each case an initial depolarisation pulse was simulated causing a step of 20 mV in membrane potential.
The conditions requested to induce the myotonia, as revealed by this model, apper to be:

1. Prolonged τ_u of pCa over a critical time at depolarisation threshold with the possibility that such process become regenerative and subsidiary.

2. The additional effects of other time-dependent factors like the delayed rectification of sarcolemma.

The inactivation of repetitive activity seems to depend of the pCa decrement and G_{Ca} inactivation itself. The exchange againts an internal cation through an electrically neutral complex of sarcolemma or the existence of other large currents canceled Ca current, have been suggested (Reuter, 1975) as posible explanations for the failure of the voltage clamp experiments to detect a Ca inward current, despite the considerable Ca influx across sarcolemma during depolarisation. The present model predicts such possibility, and clearly demonstrates that not only the normal, but even the V_r stabilizing G_L is compatible with myotonia. Thereby there are no reasons to infer, as was claimed, G_L as the single factor inducing myotonia; it would be much more realistic to consider as such all potential factors interfering in certain wayes the ion transport across sarcolemma.

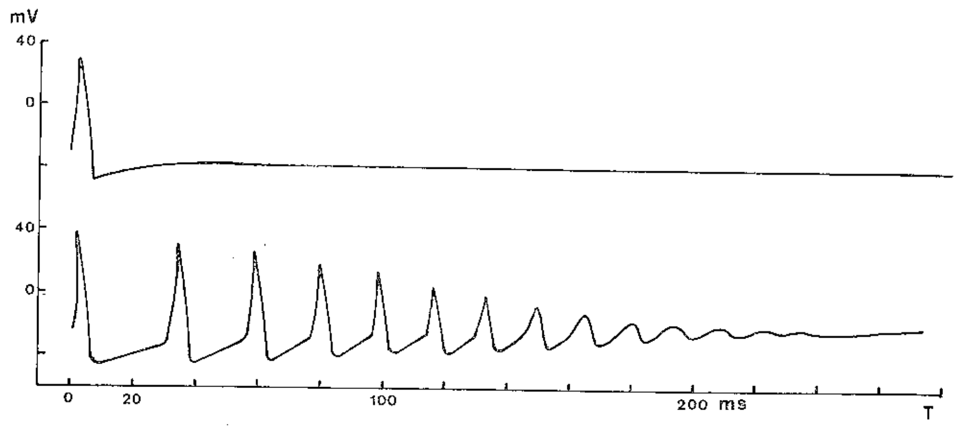

Fig. 1 - Upper trace: normal action potential
Parameters: $\tau_{Ca}/\tau_{Na}=128$; $\overline{G}_{Na}=120$ mS.cm^{-2};
$G_L=0{,}278$ mS.cm^{-2}; $\tau_u=4$ msec.
Lower trace: myotonic burst
Parameters: as above excepting $\overline{G}_{Na}=64$ ms.cm^{-2}.

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MEASUREMENTS AND MODELING ON THE NATURE OF PHYSIOLOGICAL TREMOR IN ISOMETRIC CONTRACTIONS

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The origin of physiological tremor is still a subject of controversy. Various authors reported differing result and gave different explanations for the involuntary rhythmic movements they observed. Four main hypotheses are in discussion. Together with some of their proponents, they can be characterized as follows: 1. Mechanical resonance oscillations (Hamoen, 1962; Foz and Randall, 1970). 2. Reflex loop instability (Lippold, 1970). 3. Oscillations as a consequence of a nervous sampled data control loop (Linn, 1975). 4. Increased synchronization in motor unit firing (Dietz et al., 1976). Unfortunately, some authors regarded one or the other hypothesis to be the «correct one» more or less exclusively. However, considering the different results it must be assumed that even in nonpathological situations various «oscillation modes» can exist which contribute to «physiological tremor». Dependent on the experimental situation as e.g. positioning of the limb under investigation, the kind and magnitude of load (Joyce and Rack, 1974), and the magnitude of the exerted force. Any or all of the oscillation modes may prevail or become modified according to the situation, but they can exist simultaneously, anyway. Which hypothesis or hypotheses are correct may depend on the specific experimental conditions.

It turned out e.g. that especially in isometric contraction a type of tremor can be observed which was reproducible with respect to the main frequency components while the amplitude values were instationary. This type was previously referred to as force tremor in contrast to the mechanical resonance oscillations (Rau, 1974). It was suggested that the reflex loop hypothesis seem to provide a satisfying explanation in isometric contractions.

To test this suggestion further, tremor measurements were made by means of spectral analysis. In this further study, the right arm of the sitting subject was supported at the elbow and anchored just proximum to the wrist. The hand was enclosed around the metacarpal joints by a metal cuff which was connected to a force transducer. The torque at the wrist and its changes in time as well as the surface EMG signal obtained from the M. ext. carpi rad. longus involved were recorded. The subject was instructed to exert a constant torque by controlling the torque displayed on an oscilloscope screen.

Figure 1 shows the power spectrum density (PSD) of the torque and the PSD of the EMG signal in the range between 5 and 17 Hz for two subjects

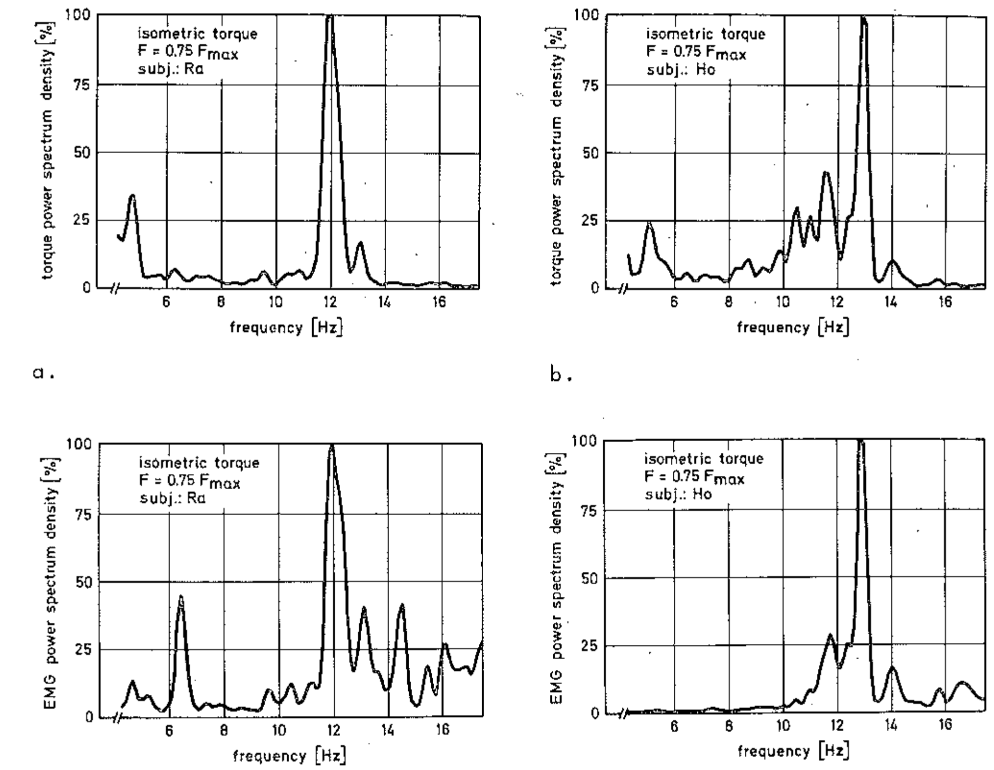

Fig. 1 - Power spectrum densities (PSD) of the torque (prefiltered: 2.5 Hz$<$f$<$100 Hz) and of corresponding EMG activities at the M. ext. carpi rad. longus (prefiltered: 3 Hz$<$f$<$100 Hz) during a constant isometric extension torque at the wrist (a and c: subj. Ra; b and d: subj. Ho).

towards the end of a 20 second contraction while 75% of maximum torque was sustained. A very clear tremor peak in the torque spectrum was found at 12 Hz for subj. Ra (figure 1 a, c) and at 12.8 Hz for subj. Ho (figure 1 b, d). At the same frequencies the main peak also occured in the EMG spectra. In addition, the EMG spectra sometimes showed a second peak at about 25-26 Hz which is approximately the first harmonic of the main peak. The second peak was barely to be seen in the torque spectrum; this could be due to the fact that the mechanical force development of the muscle fibres might be to slow to follow this high frequency of activation. The high concidence of the peaks in the torque and EMG spectra at 12 Hz and 12.8 Hz, respectively are evident as to be seen in figure 1. It indicates which part of the torque oscillations might be generated by the muscle activities. This was confirmed by calculating the cross power spectra of the torque and the EMG.

The above observed tremor mode can be explained by a reflex loop hypothesis and can be approximately described by a feed back model as reported by Zipp (1975). The frequency characteristics of an even more simplified model as shown in the inset of figure 2 demonstrates the effect

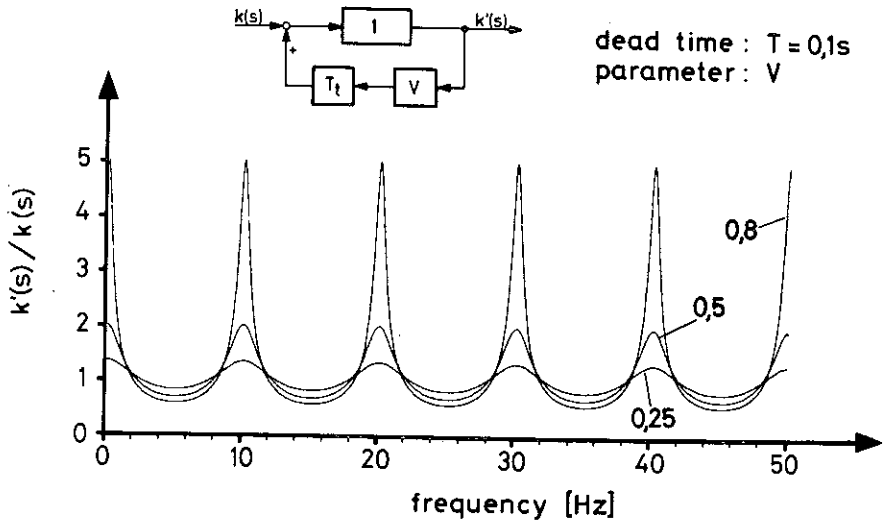

Fig. 2 - Transfer function of a feed back loop including a dead time as indicated by the inset.

of a dead time in a feed back control loop which is also included in Zipp's model. In this configuration the property of the dead time leads to a peak at a frequency corresponding to the reciprocal value of the dead time of the loop, and to additional peaks at harmonic frequencies as shown in figure 2; a change in the sign of the feed back would cause a shift of the peak frequencies of $f = -1/2\,T_t$. The amplitude values apparently depend on the loop gain.

At least in isometric conditions, this feed back loop could correspond to the reflex loop as mentioned. Sälzer (1975) reported also for a non-isometric situation in foream movements that the neuro-muscular reflex loop has a tendency to go into resonance in the 9-12 Hz frequency range which further supports the reflex loop hypothesis. In the reflex loop the higher

order resonance frequencies as expected according to figure 2 might be suppressed due to mechanical properties of the system as indicated by the more elaborate model given by Zipp (1975).

The reflex loop hypothesis is further supported by the results of preliminary measurements in finger and upper arm muscles. During isometric contractions the EMG spectrum of the first interosseus dorsalis showed the peak at about 8-9 Hz which is also reported by Dietz et al (1976); it corresponds to the 12 Hz-peak of the hand extensor muscle. Similarly the spectrum of the triceps muscle showed a peak at about 14 Hz. The peak differences of the different muscles may be explained by the different nerve conduction delays. The existence of this tremor mode as described does not necessarily exclude other types of tremor, because e.g. the pure mechanical resonance oscillations of the hand (Fox and Randall, 1970) could be only suppressed and/or modified under these isometric conditions. – Some consequences of the results for the fields of medicine and ergonomics are discussed.

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HUMAN INTRAFUSAL MUSCLE FIBERS (A HISTOCHEMICAL STUDY)

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The mechanical properties of the intrafusal fibers have been largely inferred from their structure and innervation (Boyd, 1962) as well as the frequency at which the primary and secondary afferents can be driven (Bessou and Laporte, 1962; Matthews, 1964). This data has led to conflicting interpretations such as nuclear bag fibers would structurally be fast type (Boyd, 1962) while Crowe and Matthews (1964) feel that their data could be explained if the bag fibers were of the slow type.
In order to further understand their function, in this communication we present the histochemical characteristics of the human intrafusal fibers.

MATERIALS AND METHODS

More than 50 muscle spindles from the various human muscles (axial and appendicular) formed the basis of our observations. Each spindle was studied with more than ten serial sections. Adjacent sections could not always be used due to technical difficulties. Sections were prepared according to the method previously described by Sahgal et al. (1973, 1975). The following reactions were utilized in this study:

1) Myofibrillary adenosine triphosphatase activity was determined by the method of Padykula and Herman (1955). The effect of pH was studied by the following preincubations:
   a) Veronal acetate buffer pH 3.8-5.5 for 5 minutes.
   b) 0.1 m glycine for 10-15 minutes pH 8.0-10.2.
   c) p-hydroxy mercurobenzoate (2.5 mM) (PHMB) at pH 7.0 in tris buffer for 30 minutes and post-incubation in cystein (2.5 mM) for 30 minutes.
2) *Oxidative enzymes:*
   a) Nicotinamide adenine dinucleotide dehydrogenase (NADH), Novikoff et al. (1970).
   b) Lactic dehydrogenase (LDH), Hess et al. (1958).
   c) Succinic dehydrogenase (SDH), Pearse (1957).
3) PAS.
4) Phosphorylase A and B.

RESULTS

Myofibrillary ATPase Reaction:
With this reaction darkreacting nuclear chain and light-reacting nuclear bag fibers were identified.

Preincubation in glycine at pH 8 consistently showed dark and light-reacting nuclear bag fibers, Fig. 1B. At pH 10.2 the nuclear bag fibers showed a dark reaction. When only one bag fiber was present per spindle, it showed reaction only at pH 10.2, Fig. 1A.
The nuclear chain fibers showed a uniformly dark reaction at pH 10.2 and intensity decreased at pH 8, Fig. 1A, B.
Preincubation in veronal acetate buffer at 3.8-4.2 abolished the enzyme activity in both nuclear bag and chain fibers, Fig. 1E. At pH 4.4 the nuclear bag fiber showed intermediate reaction. At pH 5.5 only the nuclear bag fiber showed activity, Fig. 1D. The nuclear chain fibers showed no activity after these pH treatments, Fig. 1C, D, E.
Preincubation in p-hydroxy mercurobenzoate inhibited the enzyme activity in the nuclear chain and bag fibers. Post-incubation in cystein restored the activity in the nuclear bag fibers while the chain fibers showed very faint reaction, Fig. 1F.

Oxidative enzymes: The nuclear bag and chain fibers demonstrated uniformly high oxidative enzyme activity, Fig. 1G.

Phosphorylase reaction: Demonstrated high activity in the nuclear chain and intermediate reaction in the nuclear bag fibers.

PAS Stain: Showed dark-reacting nuclear chain and light-reacting nuclear bag fibers.

DISCUSSION

The normal human spindle consists of the nuclear bag and chain fibers. The nuclear bag fibers show acid and alkali stable and only alkali stable adenosine triphospatase activity. This activity is inhibited by PHMB treatment and restored by cystein post-treatment. This indicates the SH group dependence of this reaction. The nuclear chain fibers show only acid labile and alkali stable ATPase reaction which is inhibited by PHMB treatment and only partially restored by cystein preincubation indicating the SH group dependence of this reaction. The adenosine triphosphatase reaction in the intrafusal fibers was similar to that of the red and white extrafusal fibers (Padykula and Herman, 1955; Guth and Samaha, 1969; Brooke

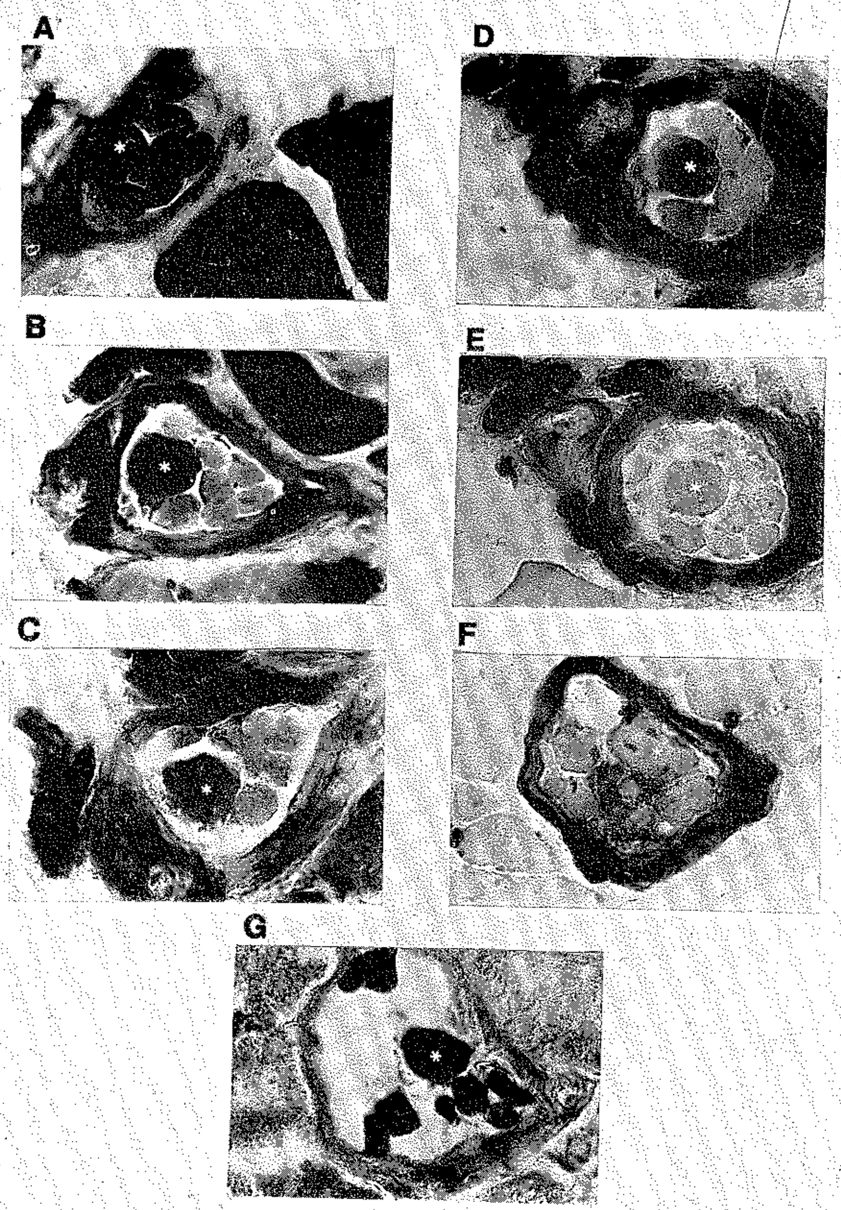

Fig. 1 - Histochemistry of the Intrafusal Fibers: Human Opponens Pollicis Muscle.
a) Myofibrillary adenosine triphosphatase reaction at pH 10.2. Note the dark nuclear bag (*) and chain fibers. b) ATPase reaction at pH 8. Note the dark (*) and intermediate nuclear bag fibers and light nuclear chain fibers. c) pH 7 as in b. d) pH 5.5 Note the dark (*) and intermediate nuclear bag and non-reacting nuclear chain fibers. e) pH 4.2 - no reaction. f) Perchlor mercurobenzoate pre-incubation. Note no reaction in the intrafusal fibers. g) NADH reaction. Note the dark-reacting nuclear bag and chain fibers.

and Kaiser, 1969). Recent work of Tsaris et al. (1971) in cat has shown that acid stable ATPase is associated with slow fibers and alkali stable ATPase with fast fibers. Inferentially, the nuclear bag fiber ATPase is like that of the fast and slow fibers, while the nuclear chain ATPase is like the fast fibers. The PAS and phosphorylase reaction in the extrafusal fibers show increased activity in the fast fibers and light reaction in the slow fibers. The PAS and phosphorylase reaction of the nuclear bag fibers, thus, is similar to the slow and that of the chain fibers to the fast extrafusal fibers. The nuclear bag and chain fibers show very high oxidative enzyme activity. The high oxidative enzyme activity has been described in the red fibers. Correlating the above data, we can conclude that the nuclear bag fibers show the histochemical characteristics of the slow and fast speed muscle fibers, and the nuclear chain fibers of the fast-twitch fibers. The recent cinematographic observation of Boyd et al. (1974) showed that the nuclear bag fibers contract alone for the dynamic and along with the chain fibers for the static function of the spindle. This observation seems to be corroborated by their histochemical properties.

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SIGNAL PROCESSING TECHNIQUE FOR ESTIMATING NORMALIZED GAIT PATTERNS

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INTRODUCTION

Kinesiologic studies have provided valuable information about the muscle synergies and kinematics of normal human locomotin. Recently there has been an increase in the use of gait analysis for clinical testing of the efficacy of operative procedures, physical therapy, and other treatment programs.
One of the main difficulties with locomotion studies is analyzing the large quantities of data measured and finding the significant relationships. Most present analytic techniques rely on some type of signal averaging. Eberhard and Inman (1954) and Milner et al. (1971) found average individual and populations gait patterns by superposition and averaging. This method results in inaccurate estimates of the activation and relaxation times of muscles because successive gait cycles have different durations. Also each person walks with different average cycle durations. Rozin et al. (1971) found the population gait patterns so different that they did not attempt to calculate average patterns. Sherwood (1975) has proposed a method that involves normalizing the data each stride before averaging. These estimates are more accurate than those provided by the other techniques.
The main purpose of the gait laboratory in the Veterans Hospital is to use gait analysis clinically. The data analysis techniques mentioned have been found unsuitable. A new technique has been devised which provides accurate estimates of the normalized gait pattern, the standard deviation of electromyographic and foot-fall pattern events, and the correlation between these events.

METHOD

The method for providing this information involves a different analytic approach. The measured electromyograms and foot-fall patterns are not to be averaged in the same manner but only certain events in each measurement are considered. The electromyographic events of import are the times when muscles are activated and relaxed. The important foot-

contact events are the times when the heel, fifth and first metatarsal heads, and toes of both feet strike and recede from the walking surface. The reference point for a gait cycle is heel strike of the concerned lower extremity.

The average gait pattern is calculated in several steps:

1. All the event times are measured with respect to the previous heel strike.

2. The average relative time of occurrence for each event is calculated (i.e. all gastrocnemius activation times are averaged).

3. These event times are normalized by dividing by the average stride duration.

4. The average normalized gait pattern is plotted.

A population average is then simply achieved by calculating the average event times for all the subjects. Figure 1 shows an average post-treatment pattern for five children.
Having the event times provides two computational advantages. Firstly, the standard deviation of specific events can be calculated and the variability of events can be evaluated. Secondly, the correlation coefficient between events can be calculated. Thus the strength of the relatinship between muscular events or between a muscular event and a foot event can be evaluated. Some preliminary hand calculations demonstrated the utility of these kinds of measures. Consider two examples.

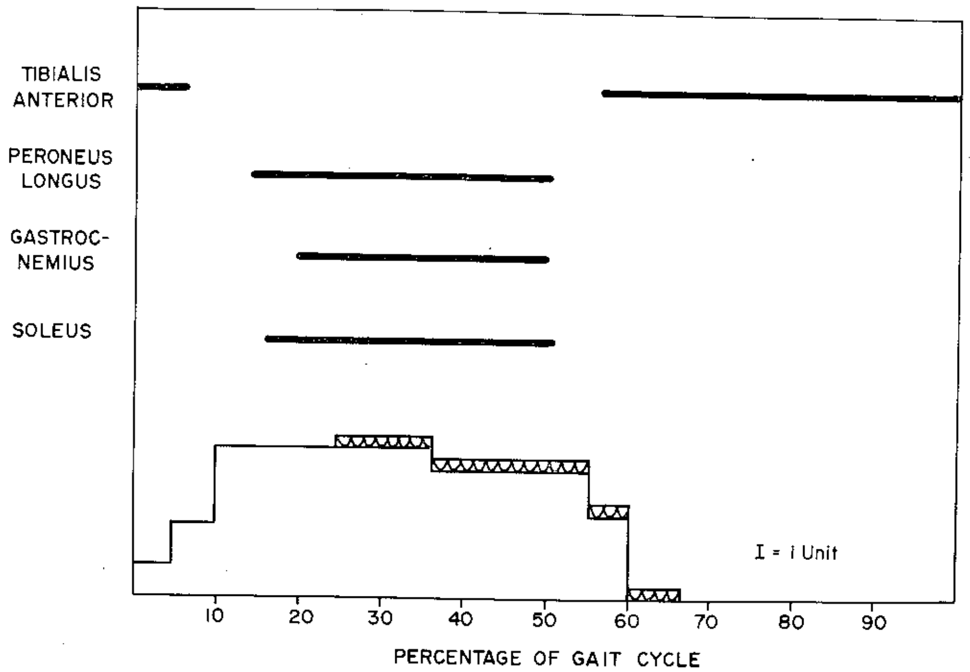
Example 1. The peroneus longus muscle has a fairly variable range of activity. However, in many individuals it is active very regularly. This is shown by the small variance of its activation and relaxation times. Also in these same individuals it is synergic with the gastrocnemius muscle. This is demonstrated by a high correlation between the activation and relaxation times of these muscles.

Example 2. The gastrocnemius muscle imparts forward momentum to the body during the single leg support portion of stance phase. It is usually thought to become quiescent at the time of toe off of the ipsilateral foot. However, our calculations show that it is more dependent upon the heel strike of the contralateral foot. This was demonstrated by a higher correlation coefficient between contralateral heel strike and gastrocnemius relaxation.

IMPLEMENTATION

All the data collected during a gait measurement session are recorded on separate channels of an analog magnetic tape recorder. A 1000 Hz sinusoidal timing signal is recorded simultaneously to demark sections of data that are to be analyzed. A PDP 8/E computer with an eight channel analog to digital (A/D) converter, real time clock, and digital magnetic tape is used to perform the analyses. The analyses are performed off-line and in two stages. In the first stage the data are played from the tape recorder into the computer. The A/D converter samples all data channels at instants when the timing signal causes interrupts via the clock. The core program then decides whether a muscle or foot event exists and measures the absolute time of occurrence. The time of occurrence and the event are stored in memory. After all the data for one run has been sampled the reduced data is stored on digital magnetic tape under a proper file name.

parameters of patient gait patterns with those of normal gait patterns.



HABITUAL TOE WALKERS, POST-TREATMENT

Fig. 1 - Post-Treatment Gait Pattern. The lines in the first four traces represent the times when the respective muscles are active. The bottom trace is the foot-fall pattern. One unit corresponds to contact of heel; two units, fifth metatarsal head; four units, first metatarsal head; one unit oscillation, toe. Any combination is a simple sum.

At a later time, the second stage of analysis is performed. A program has been developed which reads the reduced data from the designated tape file, calculates the relative event times, the averages, standard deviations, and correlation coefficients. All but the relative event times are then printed on the teletype and appended to the proper data file.

SIGNIFICANCE

This computerized data analysis scheme was implemented based upon useful information obtained from hand calculations. The analysis results from data on normal individuals will provide a list of significant data parameters. Thus the deficiencies of abnormal patterns and the efficacy of treatment programs can be judged by comparison of significant parameters of patient gait patterns with those of normal gait patterns.

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FINE VOLUNTARY CONTROL OF THE DELTOID MUSCLE WHILE FINGER MOVEMENT IS RESISTED

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It would appear that in the functional evaluation and treatment of patients with neuromuscular lesions there is a need to refine the methodology in terms of functional relationship between body segments. The present investigation has considered this relationship in terms of the effect of finger extension on neuromuscular activity within the deltoid muscle. Lundervold (1951) has observed that shoulder muscles were activated in relathionship to the different finger movements required for typewriting, and Paillard (1960) has pointed out that delicate hand movements depend on healthy function of the scapular girdle. Pauly (1976) has demonstrated that the level of activity in the elbow flexors increases as finger movement was resisted, while Paquette and Simard (1975) have observed that shoulder activity varied according to elbow position and to resistance applied to hand movement. It has been deemed necessary, based on this later study, to evaluate the effect of varying degrees of resistance applied to a given body segment on the ability of subjects to voluntarily control the neuromuscular activity of a different body segment. The segment where resistance was applied was the more distal segment.

The effect of finger extension against resistance on the voluntarily controlled neuromuscular activity from within the posterior portion of the deltoid muscle was considered with female subjects. The detection of the neuromuscular activity was done with a specially developed quadripolar wire electrode. The electrode was made with four fine insulated wires which were bared one mm at the end to be inserted into the muscle. The wires were placed in a hypodermic needle in such a way that each end was isolated from the next one by one mm. This electrode has allowed, in turn, for the detection of four bipolar signals. Two of these electrodes were inserted, one at one cm above and one at one cm below a central point of the upper one-third of the posterior portion of the right deltoid muscle. The four bipolar signals from each of the two electrodes were recorded simultaneously with an electromyographic system, and the system allowed for the feedback of the auditory and visual signs of the action potentials. These sensory signs were used to facilitate voluntary control over the action potentials. The utilization of electromyographic feedback in motor learning has been reviewed by Basmajian (1974).

Each subject was seated comfortably, hand his right upper extremity was stabilized with an experimental splint which held the arm in slight abduction and flexion, the elbow in extension, the forearm in mid-supination, the wrist in a slightly extended position, and the fingers were left free. The training procedure for voluntary control of the neuromuscular activity has included the isolation and maintenance of the activity of a single motor unit (SMU) for two minutes while all other activity was suppressed. The subject was then required to: 1) maintain the activity of the SMU during free extension of the fingers, and 2) to inhibit the activity of the SMU during free extension. A bar of an electrodynamometer was positioned then at the level of the first phalanges of the four fingers. The maximal force generated through finger extension was transduced to a linear frequency signal, and recorded on a digital voltmeter and simultaneously with the electromyographic data on a linograph paper system. The subject was required now to maintain the controlled activity of the SMU while the fingers were extended against the bar with maximal force, and subsequently with 25 per cent of the maximal force, then 50 per cent and them 75 per cent.

The preliminary analyses of the results have shown that the isolation of the activity of a SMU was achieved, and the ability to both activate and inhibit the SMU was successful, in general, during free extension of the fingers. The application of 25 and 50 per cent of maximal extension force has increased activity in the deltoid. This increased motor unit activity was overcome with practice, however, and control of the SMU was reestablished, and only some of the subjects were able to overcome the influence of 75 per cent and maximal extension force. The use of the quadripolar electrode has permitted a greater sampling of different motor units within a given muscle when the isolation and control of a SMU is required.

The results are in agreement with the previous studies cited, which have shown that the shoulder muscles are susceptible to activation while the hand is in a functional state. The ability of subjects to overcome the influence of the application of 25 and 50 per cent of maximal finger extension force on activity in the deltoid muscle demonstrates a method to evaluate motor function to its highest degree.

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CONTROLLED INTRAMUSCULAR MICROSTIMULATION: A NEW METHOD FOR THE STUDY OF MOTOR UNIT MECHANICAL PROPERTIES IN MAN

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The understanding of the physiology of muscle has been geatly advanced in recent years by studies in animals in which correlations have been found between the histochemistry and morphology of motor units and their contraction characteristics. The application of this in man has been impeded by the lack of a suitable method for recording single motor unit contractions. While the spike triggered averaging technique ol Milner-Brown, Stein and Yemm (1973a) has proved valuable in relating recruitment order to motor unit twitch amplitude, contraction time and fatigability (Stephens and Usherwood, 1975) it suffers from the disodvantage of not allowing the study of ûnit tetani or selective glycogen depletion. Moreover, study of high threshold units is very difficult because of problems in isolation of single motor unit spikes in strongly active muscle. The alternatively available technique described by Sica and McComas (1971) in which carefully graded stimulation is applied to the nerve trunk, is also unsuitable because the selection of units in any one muscle is inevitably restricted and problems of stability of stimulation make study of tetani very difficult. With these limitations in mind we have recently developed a technique we call controlled intramuscular microstimulation which suffers none of these drawbacks and allows the mechanical properties of human motor units to be characterised as fully as in animals (Stephens and Taylor, 1975). We present it here in the expectation that it may form a valuable supplement to clinical electromyography in normal and abnormal muscle.

The method depends on the excitation of motor axons within a muscle where they are spatially separated (c.f. Stålberg, 1968; Stålberg and Trontelj, 1970). as found by Milner-Brown, Stein ad Yemm (1973b), finely graded stimulation of these axons can result in unitary EMG and force responses. The present report establishes that this technique is suitable for general use in the study of human motor unit contractions. Electrical responses are monitored by a conventional monopolar EMG needle inserted into the body of the muscle. A bipolar concentric EMG needle is used for stimulation and inserted into the muscle near the motor point 1 or 2 cms from the recording electrode. The position of the stimulating electrode and stimulus strength (100 µsec, 0-30v) are adyusted till an all or nothing EMG response is detectable (Fig. 1 A). Provided reliable stimulation can be achieved without stimulating another unit, the mechanical response to

repeated stimuli can be averaged to give twitch and tetanic responses (Fig. 1 B and C). Recording of the mechanical responses depends of course on the existence of a suitable anatomical situation. We have so far used the method successfully for the first dorsal interosseous muscle in the hand (Stephens and Taylor, 1975) and gastrocnemius (O'Donovan and Stephens, 1976).

Two problems have to be faced in deciding the validity of the method. First it must be established that not more than one unit is being excited

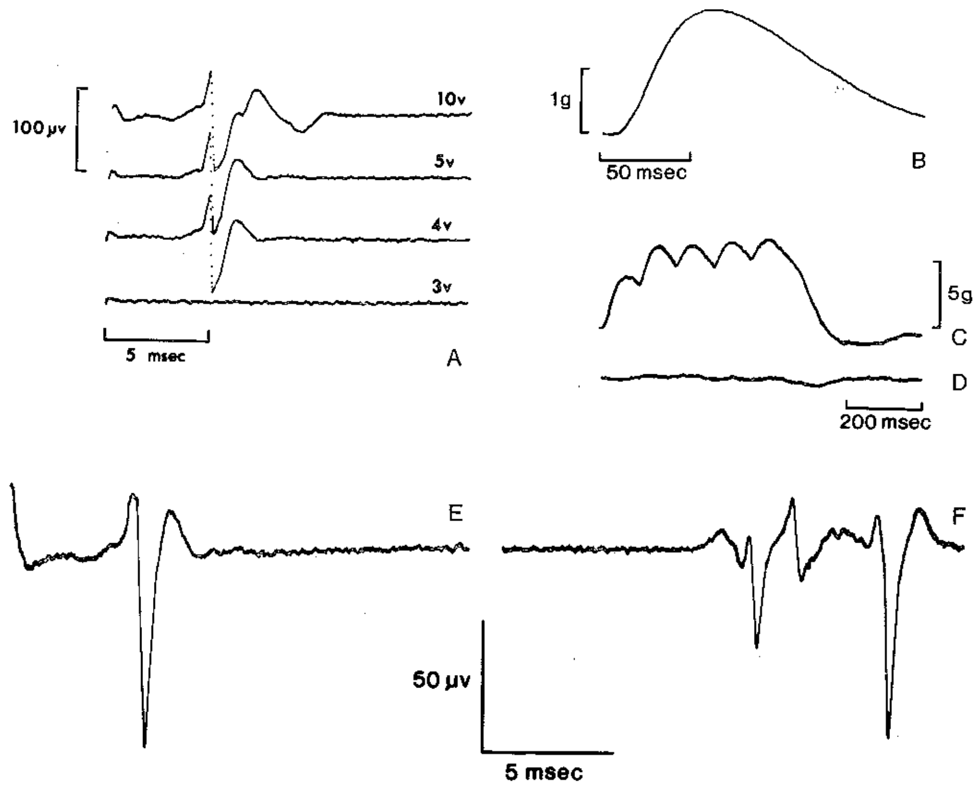

Fig. 1 - Electrical and mechanical recordings of single motor units in human first dorsal interosseous muscle. A: All or nothing EMG response following graded intramuscular microstimulation. Note the added stimulation of a second unit at stimulus strength 10V. B & C: The twitch and tetanic contractions of a single unit. 10pps stimulus set just below threshold for the EMG of the same unit in D, B, C & D averages of 128, 17 & 17 sweeps respectively. E: Unitary EMG response elicited by intramuscular microstimulation. F: Electrical activity recorded during weak voluntary muscle contraction. Same recording electrode position as in E. Note the same unitary EMG potential in E & F.

and second that the stimulation results in full activation of the unit. It is known that the muscle fibers of a unit are widely scattered within a muscle (Brandstater and Lambert, 1969; Burke and Tsairis, 1973) which makes it likely that some electrical activity would be seen of a second unit firing during carefully graded stimulation (c.f. Fig. 1 A). Furthemore, control records with stimulus reduced to just subthreshold for the unit concerned show no force development indicating that either no other unit is being excited or it has such a weak contraction strength that it does not contribute significantly to the force recorded. Activation of the whole unit is assured because on some occasion it is possible to record the recruitment of the particular unit by voluntary contraction and show that the EMG unit waveform is identical to that produced by stimulation (Fig. 1 E and F).

To date, risults have been obtained in normal muscles and in muscles affected by prolonged ischaemia following chronic arterial disease (O'Donovan, Rowlerson and Taylor, 1976). In the latter case, evidence has been obtained that the fast contracting units in these muscles have abnormally low contraction strengths. Selective glycogen depletion of these units has shown them to be composed of abnormally small Type II fibres.

The widespread use of EMG in the clinical investigation of muscle diseases has been dictated more by its convenience than an understanding of the underlying processes involved in its interpretation. The technique of microstimulation, on the other hand, offers a direct means of examining the proper functioning of normal and abnormal muscle in terms of the mechanical properties of motor units. As such it may be particularly useful in detecting the consequences of abnormal muscle morphology, histochemistry and tissue chemistry in terms of altered motor unit force development, and in so doing, reveal changes in muscle function previously undetectable by routine EMG evaluation (Tobin and Porro, 1973).

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THE RECRUITMENT ORDER AND FATIGABILITY OF HUMAN MOTOR UNITS

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To date, studies of the mechanical properties of human motor units have been limited to measurements of twitch tension, twitch contraction time and recruitment threshold (Sica & McComas, 1971; Milner-Brown, Stein & Yemm, 1973a). Using the technique of spike triggered averaging (Buchtal & Schmalbruch, 1970; Milner-Brown, Stein & Yemm, 1973b) we have now been able to examine, for the first time in normal subjects, the relationships between these properties and motor unit fatigability.

The hand is immobilised in Plasticine ad the force of abduction of the index finger produced by contraction of the first dorsal interosseous muscle recorded (Stephens & Usherwood, 1975). Using the action potential of a single motor unit from this muscle as trigger, the twitch response of the unit was extracted from the total force of abduction by averaging over 64 sweeps, the subject maintaining a steady contraction such that the unit fired at a slow rate ($\leq$ 3pps). After at least two control readings, the subject maintained a contraction such that the unit fired steadily at 10 $\pm$ 3pps for 5 minutes. At the end of this fatigue period, circulation to the muscle was occluded to prevent recovery and the force of abduction immediately averaged as before. Circulation was then restored and the force averaged again after 10 minutes recovery. Only one unit had not recovered completely by this time. An index of motor unit fatigability was defined as the ratio of the twitch amplitude at the end of the fatigue test to its mean control amplitude.

Figure 1 shows that motor units recruited at low contractions strengths were generally unfatigued or even potentiated after the fatigue test. High threshold units, on the other hand, were uniformly higly fatigued. Units recruited at contraction strengths $<$ 5% of maximum voluntary contraction strength (MVC) had relatively low twitch tension (0.18-15g) were relatively slowly contracting (twitch contraction times 39-146msec) hand had fatigue indices in the range 0.73-1.83 (I unit 0.44). Units recruited at contraction strengths $>$ 10%MVC were larger (twitch tension 15-26g), faster (twitch contraction times 33-57msec) and had fatigue indices in the range 0.09-0.34. Significant positive correlations were found between fatigability and twitch tension and twitch amplitude and recruitment threshold.

The interrelationships round in this study between contraction strength, twitch contraction time and fatigability are broadly similar to those found previously in animal muscles (see for example Burke, Levine, Tsairis &

Zajac, 1973; Proske & Waite, 1974; Reinking, Stephens & Stuart, 1975). In contrast to animal experiments, however, studies in the human allow the association of motor unit mechanical properties to be given direct functional significance in terms of the order of motor unit recruitment (cf. Stephens & Stuart, 1975). It seems clear, for example, that the present results provide at least part of the explanation for the common experience that weak voluntary muscle contractions can be maintained without fatigue for much longer periods than strong contractions. Indeed this may

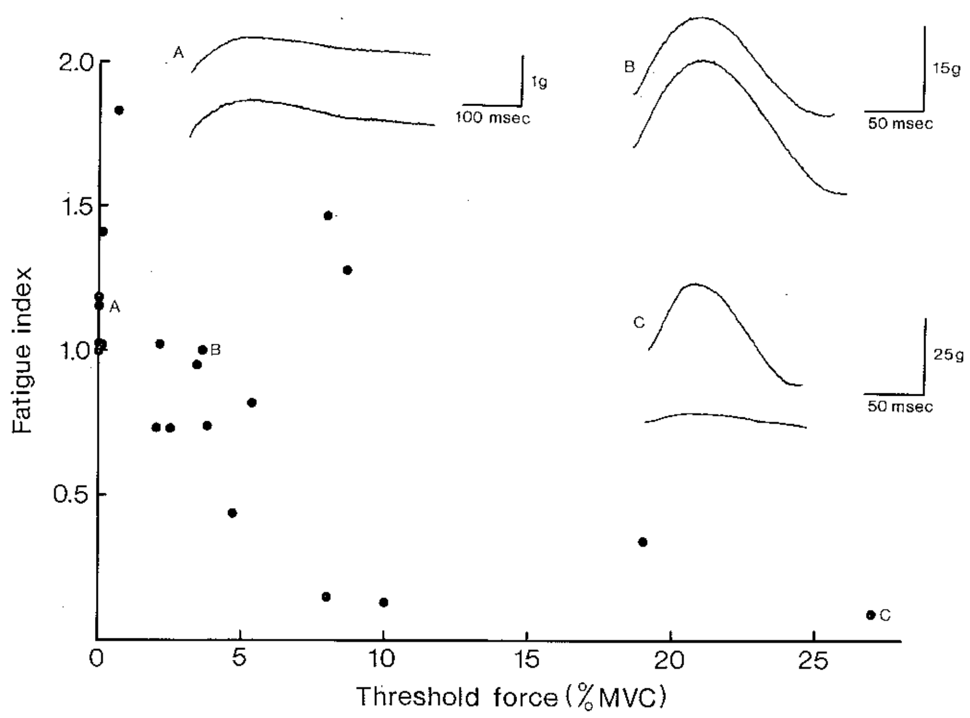

Fig. 1 - Relathionship between the threshold of motor unit recruitment and fatigue index. A, B & C show results from three different motor units. In each case the upper trace is a control and the lower the twitch recorded at the end of the fatigue test. The points corresponding to these units are labelled appropriately on the graph.

be a common design feature of the organisation of motor units within muscles, representing perhaps an adaptation to usage which enables them to participate in a wide range of activities from those demanding maintenance of small forces for long periods of time to those requiring the development of large force for a short period of time.

In view of the known correlations between muscle fibre histochemistry and mechanical properties (Burke et al. 1973) it would be of interest to include an examination of motor unit fatigability as well as motor unit contraction strength in patients suffering from various neuromuscular disorders. Such investigations may be of particular significance in cases where the recruitment order is abnormal following for example, reinervation (cf. Stein & Brown, 1973) and where comparison can be made with the morphology and histochemistry of muscle fibres revealed at muscle biopsy.

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EXCITATION OVERFLOW IN THE MUSCLES OF THE UPPER LEG: A POLYELECTROMYOGRAPHIC STUDY

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The purpose of this investigation was to evaluate:

1. the excitation overflow at different loads, including 25%, 50%, 75% and 100% of the maximal isometric force of the opposite limb.
2. the comportament of the overflow activity as a function of time.
3. the individual factors.

The following muscles of 23 young college students were investigated by means of an 8 channel polymyograph with mirror galvanometer inscription: m. vastus mediali, m. vastus lateralis, m. rectus femoris and m. biceps femoris of the right leg for evaluation of the excitation overflow and the same muscles on the left for evaluation of the activity during contraction. Surface electrodes with a markedly constant bipolar distance were used for detection. The activity was integrated by full waves integrators, constructed by the service of bioengineering of the university hospital (Ir. Fauconnier).
The subjects were lying on an inclined table, adjusted to an angle of 30° with the horizontal. The right knee was extended in the neutral or zero starting position, while the left was flexed at 60° starting from the zero point.
The subjects were well secured by belts. They were then instructed to execute a maximal isometric contraction, in extension, for 3 seconds with the left leg.
Subsequently, they executed isometric contractions at 25%, 50% and 75% of the maximal force, with 3 minute intervals between each of these loads.
Two measurements of the integrated activity during 1 sec. were taken and expressed in mV×sec by means of the reset method. The first measurement was taken at the beginning of contraction against the specified loads, and the second about 1 second later.
The movements, skewness and kurtesis were calculated from the data collected on the different muscles.
Barlett's chi-square statistic indicated that the variances for a specified muscle at the same % of loading in the different subjects were not all extimates of the same population variance.

Consequently non-parameter statistics were used to test the nulla hypothesis H_0 that the samples extracted from populations with the same distribution.

A. This hypothesis is rejected because of the differences in degree of distribution between 100%, 75%, 50% and 25% (cfr. Test of Friedman). This is expressed by a X^2 distribution with 3 degrees of freedom, as is shown in the following table.

	First measurement				Second measurement			
=3	Vast. M.	Rect. F.	Vast. L.	Bic. F.	Vast. M.	Rect. F.	Vast. L.	Bic. F.
X^2 =	54,66	53,34	51,99	50,21	61,60	61,60	52,40	57,5

All these values are significant on 0,5‰ scale.
There is also a significant difference between the integrated overflow activity, detected at 25% of the maximal load, and the integrations made at complete rest (base line), which equals 1,2 mV×sec. This is illustrated by the t distribution for 22 degrees of freedom.

	First measurement				Second measurement			
	Vast. M.	Rect. F.	Vast. L.	Bic. F.	Vast. M.	Rect. F.	Vast. L.	Bic. F.
t=	3,97	6,23	4,19	5,8	4,11	5,83	4,75	6,04

This table is significant on 1‰ scale.

B. There is also a difference in the degree of distribution between the subjects for the different muscles and the loads (cfr. Test of Kruskal and Wallis).
This is demonstrated in the following table of X^2 values for 22 degrees of freedom.

	Vast. M.	Rect. F.	Vast. L.	Bic. F.	Significance
25%	41,61	163,87	42,33	44,30	$X^2_{0,95} = 33,92$
50%	41,74	41,92	40,03	43,01	at least significant
75%	42,07	40,06	42,03	42,39	on 2,4% scale
100%		34,42			on 5% scale

However the nulla hypothesis (H_0) is not rejected for m. vastus med., m. vastus lat., m. biceps fem., at maximal loading (X^2 : vast. m. 32,71; vast l. : 33,13; bic. : 31,36).

C. Spearman's rank correlation coefficient between the integrated excitation overflow in the right leg and the exterted force on the left is slightly significant for m. rectus femoris, and very partially so for m. vastus lat.

m. Rect. F. $r_s = 0{,}46$; test for H_0; $2 = 2{,}17$ (1,5% s.): First measurement.

m. Rect. F. $r_s = 0{,}49$; test for H_0; $Z : 2{,}28$ (1,1% s.): Second measurement.

m. Vast. L. $r_s = 0{,}37$; test for H_0; $Z : 1{,}75$ (4% s.): Second measurement.

The null hypothesis of independance is maintained for the other muscles and test situation.

D. Evaluating the comportment of the excitation overflow as a function of time, the second measurement showed an increase in the integrated activity in the muscles investigated and at different loads, as shown by the sign test:

100% loading: T significant on 0,5 scale unilateral for all the muscles.

75% loading: vast. med. and lat. on 0,5 s.; rect. fem.: 2,5%; bic. fem.: 5% s.

50% loading: T significant on 2,5% s. for all the muscles.

25% loading: vast. med. and lat. on 5,0% s.

E. Kendall's rank correlation test for 4 paired observations (25% up to 100%) is positive in 59,23% (range 51,78-66,39) of the 184 sets in this study. A power curve with a significant coefficient of determination can be drawn in 36,54% of the position rank correlations (range 32,62-51,04).
The mean value of the power equals 2,10. the standard deviation 0,72 and the standard error 0,17.
An exponential function of the form $Y = ae^{bx}$ describes significantly the curves in the other 63,64% (48,96-76,38).
It is equally possible to realize a linear equation of the form $Z = a_0 + a_1x + a_2y$ between the integrated activity of excitation overflow Z, the integrated value X of the activity during contraction in the investigated muscle and the developed force Y for example: a_0: —20,25; a_1: 1,67; a_2: —3,77 in one set of data about m. rectus femoris.

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EMG AUTOMATIC ANALYSIS OF THE TIBIALIS ANTERIOR MUSCLE IN NORMAL SUBJECTS

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The aim of this study is to set up a program using a digital minicomputer, for an automatic analysis of the EMG tracings.

The main steps of our research are: 1) isolation of an EMG signal (MUAP) with its own real onset and offset points, 2) identification of the MUAPs among the sampled signals. In other words identification of each discharging signal that is found to be repeated at least 3 times; 3) measurements of the MUAPs parameters (duration, amplitude, number of phases, number of peaks etc.): mean value ± S.E. as well as statistical evaluation of the parameters under examination.

During the first step of this program, the computer constitues a «ciclic buffer» representing at least 17 msec of EMG tracing; when a fixed digital threshold is exceeded, a sufficient number of points are accumulated in such a way that each signal is preceded by an initial tract of at least 5 msec.

In a subsequent phase a cleaning operation is effected, eliminating the signals having insufficient amplitude in comparison with a fixed value (as they come from a distant M.U.) or displaying a remarkable difference between the initial and the final tract.

The accepted signals must be recognized to be «real MUAPs» as described above, before being analyzed; the other signals (artifacts or superimpositions of different MUAPs) are eliminated. This cleaning operation has been carried out using the cross correlation coefficient between each considered MUAP and a chosen series of following MUAPs, accepting as «real» those MUAPs that repeat themselves 3 times.

The signals so recognized are then submitted to an averaging operation to improve the signal-noise ratio.

Afterwards a comparison is made between the last identified MUAP and a set of 3 (or 5) previously identified MUAPs; the last MUAP will be therefore accepted only if it appears to be different from the others. At the end of this stage a series of different «real MUAPs» will be obtained, each different from the others, which could be stored on a mass storage device.

During the third step the parameters of a single MUAP are calculated, using suitable threshold and comparison among the points of each MUAP. The reasults will be output on a peripheral device.

For this research we have employed a minicomputer Digital PDP/8L, provided with 8 K memory, connected to an A/D converter (TMC CAT 400/C) and supplied with a dual unit of digital tape devices.

The MUAPs of the tibialis anterior muscle of normal subjects were picked up by a concentric needle using a Medelek apparatus for amplification, connected with a J. Packard F.M. tape recorder.

The subjects were voluntary with ages ranging from 27 to 75, as shown by Arrigo e Coll. (1975).

The clinical examination did not show any evidence of peripheral nerve or muscular involvement.

The values of MUAPs obtained in cases 1 and 2, who were certainly normal subjects, as in previous studies, were averaged and compared with all the rest of the cases.

Fig. 1 and 2 show the results and statistical evaluation.

Significant differences were found when taking into consideration the age range, subclinical diabetes (2 subjects) and alchool addiction (2 subjects). The data obtained confirming this analysis, seem to be useful for a better identification of subclinical or bordeline cases.

Case 1+2 → Case 3	Amplitude	t = n.s.	N° Phases	t = *
	Duration	t = ***	N° Peaks	t = ***
Case 1+2 → Case 4	Amplitude	t = *	N° Phases	t = n.s.
	Duration	t = n.s.	N° Peaks	t = *
Case 1+2 → Case 5	Amplitude	t = n.s.	N° Phases	t = n.s.
	Duration	t = n.s.	N° Peaks	t = n.s.
Case 1+2 → Case 6	Amplitude	t = n.s.	N° Phases	t = n.s.
	Duration	t = *	N° Peaks	t = n.s.

$p > 0.05$ n.s. $p < 0.05$ * $p < 0.01$ ***

Fig. 1

Cases 1+2 → Case 7	Amplitude	t = *	N° Phases	t = *
	Duration	t = ***	N° Peaks	t = ***
Cases 1+2 → Case 8	Amplitude	t = n.s.	N° Phases	t = *
	Duration	t = *	N° Peaks	t = n.s.
Cases 1+2 → Case 9	Amplitude	t = *	N° Phases	t = n.s.
	Duration	t = *	N° Peaks	t = n.s.
Cases 1+2 → Case 10	Amplitude	t = *	N° Phases	t = n.s.
	Duration	t = *	N° Peaks	t = n.s.
Cases 1+2 → Case 11	Amplitude	t = *	N° Phases	t = n.s.
	Duration	t = *	N° Peaks	t = n.s.

$p > 0.05$ n.s. $p < 0.05$ * $p < 0.01$ ***

Fig. 2

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ELECTROMYOGRAPHIC AND KINEMATIC STUDIES ON WALKING PATTERNS WITH VARIOUS FOOTWEAR

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Various kinds of footwear have been used in daily life, and especially in Japan we have had many traditional and foreign types. It is postulated that the type of footwear should have some influence on gait.
The purpose of this study is to analyse walking patterns and activities of leg and foot muscles when various footwear are worn.

MATERIALS AND METHODS

Twenty-five young adults without any diseases were selected, and each subject was instructed to walk at his natural speed ten times with different footwear. Three kinds of footwear (Fig. 1), the first with fully-flexible soles (rubber sandal, setta, zori, and house slipper), the second with semi-flexible soles (shoes with various heel heighst), and the third with inflexible soles (geta, takageta, and wooden sandal) were used for the study.
The electrical activities of leg and foot muscles (tibialis anterior and posterior, gastrocnemius, peroneus longus, extensor hallucis and digitorum, flexor hallucis and digitorum longus, and abductor and adductor hallucis) were relayed from dual-wire internal electrodes into the amplifier. The signal was recorded on an eight-channel oscillograph. Electrical stimulation was performed to ascertain the position of the electrodes.
Walking patterns were observed with a stroboscopic camera, and temporal change of angles of the hip, knee, ankle, and first metatarso-phalangeal joints were measured and correlated with muscle activities.

RESULTS

Walking patterns and activities of leg and foot muscles when a given footwear was worn was compared with those in barefoot walking. The results were as follows:

1. Electromyographic study
   Group A muscles which are working as the movers of the ankle (tibialis anterior and posterior, gastrocnemius, and peroneus longus) chan-

ged their phasic and quantitative activities when the footwear with semi-flexible soles were worn. The active phase of the tibialis anterior shifted gradually to the stance phase as the heel heigh was raised. Group B muscles ranging from the leg to the toes (extensor hallucis and digitorum longus and flexor hallucis and digitorum longus) showed quantitative changes when the footwear with inflexible soles were worn. Furthermore, activity of the flexor hallucis longus was revealed both in the stance and in the swing phases.
Group C muscles which are intrinsic muscles of the foot changed their phasic quantitative activities as did group B when the footwear with inflexible soles were worn. Those muscles were active even in a part of the swing phase.
On the other hand, the footwear with fully-flexible soles had little influence on activities of all the three groups of muscles.

2. Kinematic study
Temporal change of angles of the hip during walking was hardly influenced by the degree of flexibility of footwear soles. The range of motion of the knee joint decreased when the footwear with inflexible soles were worn, but was identical to that in barefoot walking when the footwear with fully-flexible soles were worn.
Temporal change of angles of the ankle joint was influenced by the footwear with inflexible and semi-flexible soles, and its range of motion gradually decreased as the heel height of those with semi-flexible soles was raised. The range of motion of the first metatarso-phalangeal joint decreased markedly when the footwear wtih inflexible and semi-flexible soles were worn.
On the other hand, the footwear with fully-flexible soles had little influence on angular changes of all the joints as in the case of the electromyiographic study.

The results obtained in these studies are briefly summarized in Table 1.

CONCLUSIONS

1. Footwear with fully-flexible soles hardly exert any influence on angular changes of the joints and muscle activities of the lower limb. Therefore, walking with such footwear in almost identical to walking barefoot.
2. Footwear with semi-flexible soles decrease the motion of the first metatarso-phalangeal joint alone, and have some influence on activities of the ankle muscles only when the heel height is raised. Those footwear with a heel height over seven centimeters exert an especially great influence on muscle activities.

Fig. 1

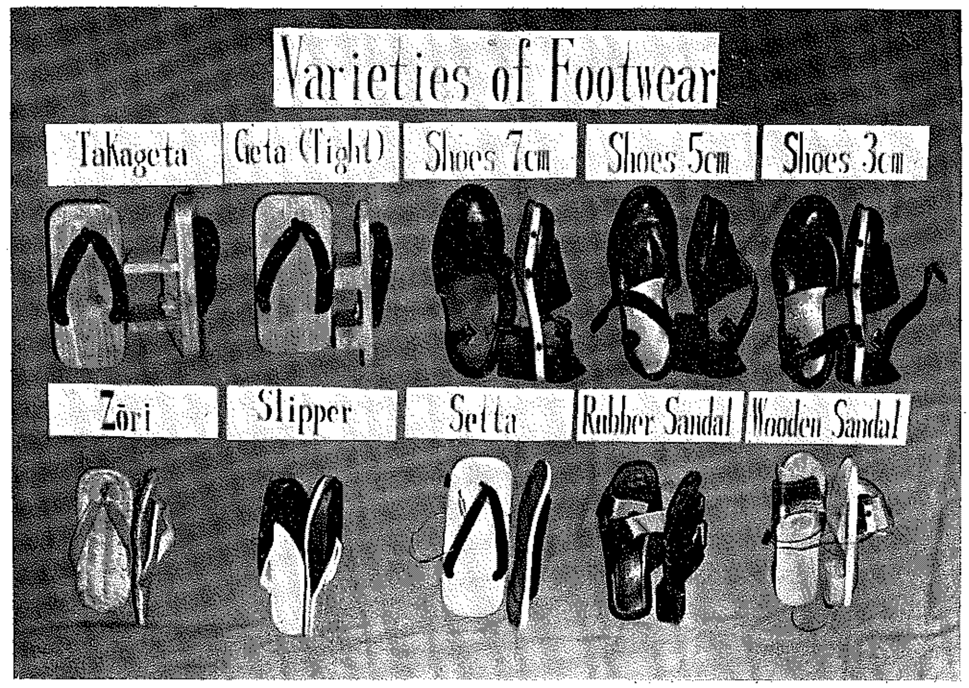

Table 1

Change of muscle activities in various footwear

	Shoes semi-flexible sole	Geta, Takageta, Wooden sandal inflexible sole	Rubber sandal, Setta, Zōri, Slipper fully flexible sole
A - groupe Tib. ant. Gastro. Peroneus. Tib. post.	+	−	
B - groups E.H.L. F.H.L. E.D.L. F.D.L.	±	+	−
C - groupe Abd. hall. Add. hall.	−	+	±

3. Footwear with inflexible soles decrease angular changes of the knee, ankle, and first metatarso-phalangeal joints, and have certain influence on activities of the extrinsic and intrinsic muscles of the toes especially when those with interdigital thongs were worn. Consequently, walking with such footear is clearly different from that with other kinds of foowear.
4. According to the above-mentioned results, it should be presumed that rigidity of the soles of footwear and their heel heights have an influence on various aspects of level walking.

EMG ANALYSIS IN HEMIPLEGIC PATIENTS TREATED BY MEANS OF FUNCTIONAL ELECTRICAL STIMULATION (FES) OF THE PERONEAL NERVE

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METHOD

Thirteen patients with a chronic residual hemiplegic state were examined before treatment, after a conventional physiotherapeutic treatment for three months, and after a treatment with Functional Electrical Stimulation (FES) of the peroneal nerve for three months. Standardized constant isometric contractions of half maximal strength were executed during 60 seconds, first dorsal flexion of the foot and then plantar flexion of the foot. The EMGs of dorsal flexion muscles (tibialis anterior muscle) and plantar flexion muscles (soleus muscle) were recorded with surface electrodes and analyzed by a PDP 8/I computer. Executed tension, integrated amplitude, peak-to-peak amplitude and number of peaks were measured per second.
In the 60 second contraction period, trend, mean and standard deviation were calculated. The more detailed description of the method and normal values have been reported by Visser and de Rijke (1975).

RESULTS

1. *Before treatment*

In dorsal flexion the tension executed was increased, nevertheless the amplitudes and the number of peaks were decreased. In plantar flexion the tension as well as the amplitudes and number of peaks were decreased. This can be interpreted as a paresis with a dominance of dorsal flexor muscles. The dorsal flexor muscles were in a tonic condition (high tension, low level of electrical activity).

2. *Conventional physiotherapeutic treatment*

In dorsal flexion the tension executed decreased to value below normal. On the other hand, the amplitude and number of peaks increased (however, still below normal values). In plantar flexion the tension executed increased but was still low, and the amplitudes and number of peaks did not show any change.

The accompanying activity in the antagonistic muscles was found to increase. This can be interpreted as a paresis with decreasing dominance of dorsal flexor muscle and increasing dominance of plantar flexor muscles and too high a level of activity in antagonistic muscles which should be relaxed.

3. *After FES treatment*

In dorsal flexion a further decrease occurred in the tension executed. In plantar flexion the tension executed has decreased again. The increased activity of the antagonist has disappeared.
This can be interpreted as an improved balance between dorsal and plantar flexor muscles and an improved relaxation of antagonists.

DISCUSSION

Before treatment a paresis existed with dominance of dorsal flexor muscles which were in a tonic condition. After conventional treatment the paresis had shown a decreasing action of dorsal flexor muscles, an increasing action of plantar flexion muscles, and in antagonistic muscles too high a level of activity was observed. After FES treatment the paresis still existed, but the functional balance between dorsal and plantar flexor muscles had improved. The contractions were less tonic and there was less activity in the antagonistic muscles. This may very well improve walking capacity.

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INTEGRATED ACTIONS OF MASTICATORY MUSCLES: SIMULTANEOUS EMG FROM EIGHT INTRAMUSCULAR ELECTRODES

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Shortly after World War II, Moyers (1949) introduced electromyography for investigating masticatory dynamics. Since then many papers have been published, using different kinds of apparatus and instruments according to a particular study or investigation to be performed. This heterogeneity has been against potential uniformity in the interpretation of the electromyograms. Diverse opinions are numerous among authors who have studied the muscles of mastication electromyographically in normal subjects. It is obvious also that mastication is one of the most important functions of the mouth and dental medicine, and considerable data about these functions are still missing for the human being. We have undertaken, therefore, to analyze through electromyography the temporalis, masseter, medial pterygoid, anterior belly of the digastric, mylohyoid and geniohoid muscles in individuals having «normal» or «ideal» dentofacial growth, good dentition with only minor deviations from «normal» anatomy, and freedom from temporomandibular joint disturbances.
Indwelling bipolar fine-wire electrodes (Basmajian and Stecko, 1962) were used exclusively. The electrodes were inserted into each muscle by means of 27-gauge hyopdermic needles. These electrodes consisted of two nylon-insulated Karma alloy wires (diameter 25 μ) with their tips bared and bent back to make one bipolar electrode. The complete assembly was sterilized for one hour at a temperature of 130°C.
The proximal bared ends of the electrode were connected to spring tempered brass-ware connectors fashioned according to the procedure of Basmajian et al. (1966). This connected the subject to the input cables of 8 low-level preamplifiers (modified LRA 042 Differential Amplifiers from Argonaut Associate Incorporated, Blaverton, Oregon). All electrical potential signals were monitored either by an audio amplifier connected to an eight-inch loud-speaker and/or by the four-beam Type 564 Tektronix Storage Oscilloscope. A Hewlett-Packard (Model 3955 C.14 Channel) magnetic tape recording system was used to record and store the electromyographic signal. The data stored on the magnetic tapes were transferred to Kodak Linograph Direct Print bromide paper by means of an ultra-violet recorder (Honeywell 1508 Visicorder). The whole experiment was also recorded by a TV camera together with the voice of the experimentor on tape. This

permitted correlation of the different phases of movement with electromyograph on paper. The routine calibration was 500 μV and 1000 ms per division.

The muscles were studied during free movements and against resistance. The records showed that there was activity in the temporalis muscle during: closing the jaw with contact of the teeth, and against resistance; free lateral movement to ipsilateral side, against resistance, and with occlusal contact; retraction of the jaw from protroction without contact, and with occlusal contact; incisor chewing gum, chewing gum on the right and left molar side; normal mastication and in forceful centric occlusion. There was activity in the masseter and medial pterygoid muscles during the following movements: closing the jaw slowly without occlusal contact, with occlusal contact and against resistance; free lateral movement to contralateral side, against resistance, and with occlusal contact; protraction of the jaw without occlusal contact, and with occlusal contact; swallowing saliva and water; incisor gum chewing, chewing gum on the right and left molar side, normal mastication and in forceful centric occlusion.

There was acticity in the diagastric, mylohyoid and geniohyoid muscles during the following movements: opening of the jaw slowly and maximally, against resistance; closing the jaw against resistance; free lateral movement to ipsilateral and contralateral side, against resistance and with occlusal contact; protraction of the jaw without and with occlusal contact, and against resistance, swallowing saliva and water, protraction of the tongue. They were working in antagonism during chewing gum and normal mastication.

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KINEMATIC AND MECHANICAL CORRELATES OF SKILL ACQUISITION: 150 CM. SUBJECT-TO-TARGET DISTANCE

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The purpose of this study was to define quantitatively, the modifications in the dynamic characteristics of myoelectric activity and parameters of movement occurring during the acquisition of a motor skill. More specifically, this study utilized a combination of synchronized integrated electromyography and cinematography to analyze the effects of a training program on the discrete voluntary movement changes of an upper limb while involved in a novel motor task of throwing a ball for accuracy.

SELECTED LITERATURE REVIEW

The effects of practice on myoelectric activity have been investigated and factors allowing for the various musculatures order their temporal responses identified. Person (1958) and Okamoto and Kumamoto (1971) have noted a high level of activity in both agonists and antagonists in early attempts at novel movements, with practice establishing reciprocal relationships between opposing muscle groups. Fischer and Merhautova (1961) and Kamon and Gormley (1968) agree that trained individuals are able to increase the amplitude of their action potentials at decisive moments, while early unsuccessful attempts were characterized by slow and inappropiate responses. Hobart and Vorro (1974), Vorro and Hobart (1974) and Hobart et al. (1975) have been able to further break down the temporal characteristics of muscle responses. They agree that muscles will exhibit an earlier premovement contraction and a reduction of time needed for a muscle to reach peak electrical activity as skill is acquired. Investigations into the changes in total myoelectric activity resulting from increased proficiency have produced conflicting results, some authors finding increases (Hobart et al.), others finding decreases (Kamon and Gormely).

METHODS AND MATERIALS

Twenty-three adult male subjects participated in a throwing task, the goal of which was the development of accuracy. The subjects were required to propel a ball, with the right (dominant) limb, onto a target with a shoulder flexion movement. The subject-to-target distance was 150 cm. A detailed description of the task and experimental setting may be found in Hobart and Vorro (1974) and Vorro and Hobart (1974). Briefly, synchronized cinematography and electromyography were used to gather data intermittently (trials 1, 2-3, 51-53 and 101-103) throughout the learning sequence. A 16 mm. Model H 16 Bolex camera operating at a film transport speed of 64 frames per second was used for the photography. Surface electrodes, placed 3 cm. apart from each electrode center and along the muscle fiber bundles, monitored the electrical activity. They were placed at the mid-longitudinal and mid-cross sectional point of each muscle. Four pairs of Beckman standard bi-polar electrodes (disk diameter 1 cm.) were used for placement on the posterior portion of the deltoid, teres major, latissimus dorsi, and pectoralis major (pars clavicularis). Four additional pairs of Beckman miniature bi-polar electrodes (disk diameter 2.5 mm.) were used for the anterior and middle portions of the deltoid, coracobrachialis, and biceps brachii (caput breve). Skin resistance measures of less than 3000 ohms were achieved with all subjects. The myopotentials were processed and recorded by an eight channel Beckman Type RM Dynograph. The response characteristics of the integrator couplers were: 3DB cut-off points of 75 and 1500 Hz, with a 6DB per octave roll off at the high and low frequencies. These couplers took the high frequency myoelectric responses and routed them to the input transformer of the preamplifier, bypassing the preamplifiers input chopper. A frequency response extending to 500 Hz was obtained, exceeding the highest muscular potential frequencies. After amplification in the preamplifier, the signal was passed back into the integrator coupler by a single stage transitor circuit in order to produce sufficient signal amplification for linear rectification. The signal was then coupled to a full wave rectifier and «integrated» by a low pass filter. The signal was thus turned into DC and the ripple on the DC filtered out for frequencies above 40 Hz. The signal was then coupled to the amplifier for recording. The preamplifier gain was set at 1 mv/cm. and the amplifier gain at X1 mv/cm. Data were quantified numerically with the use of an X/Y coordinate digitizer and reduced to variables of interest by an BM 370-145 computer. A stepwise linear regression technique was utilized for the statistical analysis. This procedure recursively constructed descriptive equations, taking into account the interrelations among the variables, which best described the observed phenomenon, skill acquisition. Mean values were calculated for the dependent variable, performance score (the deviation of the ball from the target center). The kinematic variables: movement time, release angle, and release velocity were computed for each subject for each trial block. The myotemporal variables: premovement contraction time, time to peak electrical activity, and coordi-

nation time (elapsed time between the beginning of contraction for each muscle and the beginning of contraction for the first muscle to contract, the anterior deltoid) were also computed for each subject and for each trial block. A similar calculation was made for the remaining myoelectric variable, total electrical activity.

RESULTS AND CONCLUSIONS

The criterion measure or performance score was significantly altered during the learning sequence (166 cm. average deviation from the target center at trial 1, to 53 cm. during the final trial block), with the majority of learning occurring very early in the practice period (38 cm. between trial 1 and mean of trials 2 and 3). In order to produce this rapid and considerable performance improvement, the subjects found it necessary to make specific kinematic and mechanical changes in the throwing motion. The overall kinematic alterations occurring over the learning sequence were as follows: the angle of the limb at ball release decreased .914, .879, .793, and .741 radians; total movement time of the limb excursions decreased .243, .194, .191, and .189 seconds; and the velocity of the limb at ball release increased 5.53, 7.71, 8.09, and 8.27 radians/second. Apparently, the subjects found that success could best be mediated through a combination of these changes. Accuracy, then, was improved by producing a more rapid overall limb excursion, combined with a lower (more direct) limb angle at ball release, and an increase in velocity at that point.
Generally, the mechanical variables followed a consistent pattern. As practice progressed, all eight muscles began their premovement contraction times considerably earlier then in the initial trial, with the largest increases occurring between trial blocks 1 and 2-3. All coordination times decreased, indicating that the subjects found it necessary to shift the initiation of contraction for the various musculatures so as to more closely approximate that of the anterior deltoid. The largest portion of the change in coordination time occurred between trial blocks 1 and 2-3. The times to peak electrical activity were very individualistic. From the initial trial onward, the subjects either increased or decreased these times to affect whatever alterations were needed for success. Total electrical activity generally increased throughout the learning sequence, however, with the greatest increases occurring between trial blocks 1 and 2-3.
The initial equation, descriptive of the improvement of performance between trial blocks 1 and 2-3 indicated that 4 myotemporal, 2 kynematic, and 4 total activity variables were primarily responsible for early skill acquisition. These included: decreased release angle (0.35 rad.), increased re-

lease velocity (2.18 rad/sec.), increased premovement time for the middle deltoid, teres major, and pectoralis major (.039, .043, and .034 sec. respectively), increased time to peak electrical activity for the teres major (.002 sec.), and increased total electrical activity for the coracobrachialis, teres major, and anterior and middle portions of the deltoid (.390, .937, .708, and .241 sq.cm. respectively).

The second equation, descriptive of improvement later in the learning sequence, identified 7 myotemporal, 1 kinematic, and 2 total activity variables, while the final equation indicated 6 myotemporal, 1 kinematic, and 3 total activity variables as being of primary importance. The muscles making up these equations varied and their index of change was greatly reduced.

The subjects found it necessary to engage in considerable experimentation in order to achieve success and to be consistent with that success. The early alterations, as indicated previously, accounted for the sizable improvement in performance. These variables, indicated by the regression technique, were the most salient in the experiment. They represented the largest change in the actual values between the early trial blocks and that change accounted for the largest adjustment in skill level. Additional improvement at the task utilized the newly established baselines (post third trial) and only slight modifications were made from that point onward. As practice progressed, the subjects found it necessary to substitute variables dealing with temporal aspects into the equations. Seemingly, a temporal sequencing of the musculature became more important in the later stages of the practice sessions, accounting for any additional improvement in the skill level.

It was concluded that: 1. the largest mayority of skill improvement occurred within the first three trials, 2. the kinematic variables demonstrated a large initial adjustment within the first three trials, 3. total electrical activity increased considerably within the first three trials, as did the myotemporal variables, 4. myotemporal variables tended to replace the more quantitative changes in total electrical activity and kinematics as the subjects became more proficient at the task. Improvement subsequent to the third trial was more a function of subtle adjustments in temporal sequencing than further gross modifications in total electrical output or limb position and rate.

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GAIT ANALYSIS IN HEMIPLEGIC PATIENTS TREATED BY MEANS OF FUNCTIONAL ELECTRICAL STIMULATION (F.E.S.) OF THE PERONEAL NERVE

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METHOD

Thirteen patients with a chronic residual state (no spontaneous improvement) were treated 3 months with conventional physiotheraphy treatment (gait-training), active/passive movements, and treatment according to Kabat (proprioceptive neuromuscular facilitation). They were treated, thereafter, for another three months by means of Functional Electrical Stimulation (F.E.S.). In F.E.S. treatment the peroneal nerve was stimulated on the affected side during the swing phase of the leg. Dorsoflexion and eversion of the foot was obtained with this procedure.
The gait pattern was analyzed before treatment (BT), after 3 months conventional physiotherapy treatment (AC), and after 3 months treatment with F.E.S. (A.S.). This was done by the measurement of such phenomena as the stance-phase, swing-phase and the bipedal-time by means of a registration system developed specially for this goal (F. van Faassen, 1974). During the last examination the parameters were registered not only with the stimulator functioning (AS-SF), but also without the stimulator (AS-SO). In the healthy individual, if one divides the steep-time of the left leg by the right, one gets the symmetry-factor φ which is normally 1 when the condition is ideal.

RESULTS

In the statistical analysis of the results there was no significant improvement in the symmetry-factor from situation BT to AC. In the condition AC to AS-SF there was a significant improvement, which did not exist from situation AC to AS-SO.

CONCLUSION

The conclusion is that the gait-pattern of a hemiplegic patient becomes significantly more symmetric by the application of F.E.S.

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