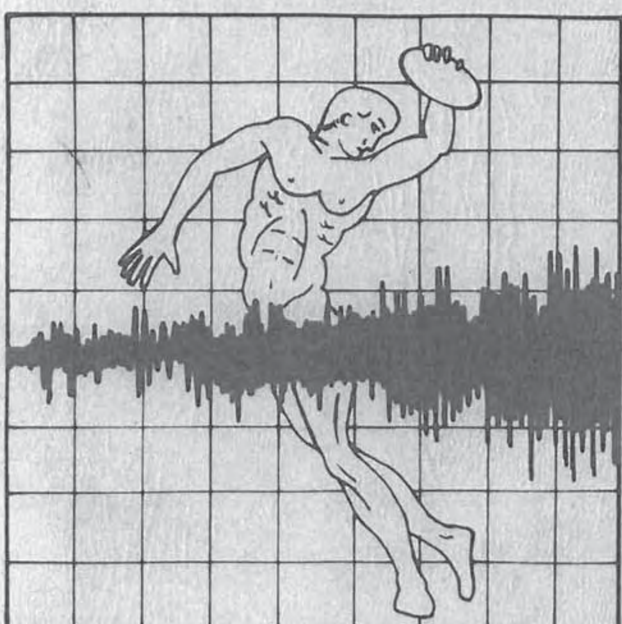
# PROCEEDINGS OF THE 4th CONGRESS OF THE INTERNATIONAL SOCIETY OF ELECTROPHYSIOLOGICAL KINESIOLOGY



5-10 AUGUST 1979

BOSTON, MA., U.S.A.

PROCEEDINGS

OF

THE 4TH CONGRESS

OF

THE INTERNATIONAL SOCIETY

OF ELECTROPHYSIOLOGICAL KINESIOLOGY

5 – 10 AUGUST 1979

BOSTON, MA, U.S.A.

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*Dear Colleagues and Friends:*

*It is my pleasant duty to welcome you to the Fourth International Congress of I.S.E.K.. This is the first time the Congress has been held in the United States and we are proud that Boston was chosen as the host city.*

*The Congress will truly be an international meeting, with at least 18 countries represented by participants and contributors. This varied international participation, along with the quantity and diversity of scientific papers, is a testimony to the coming-of-age of our Society. My efficient and enthusiastic colleagues of the Organizing Committee deserve mention. Their dedicated involvement certainly has enhanced the quality of the Congress. I would like to express my gratitude to all of them. A special acknowledgment is in order for Liberty Mutual Insurance Company, who generously supplied us with operating funds throughout the planning phase of the Congress.*

*We are particularly pleased that the distinguished scientist Dr. Elwood Henneman accepted our invitation to present the keynote address.*

*The scientific program, exhibits and social activities promise to provide a great opportunity to review the current interests of various laboratories throughout the world, to exchange ideas, to rekindle old friendships and cultivate new ones. The cultural ambience of Boston provides a superior setting in which to realize these goals. Welcome and best wishes for a fruitful Congress.*

*Carlo J. De Luca*
*Chairman*
*4th Congress of I.S.E.K.*

INTERNATIONAL SOCIETY OF ELECTROPHYSIOLOGICAL KINESIOLOGY

It is indeed my pleasure, on behalf of the Membership, to thank Dr.Carlo DeLuca for accepting the responsibility of Secretary-General for this Fourth International Congress. The planning and organization of any international congress requires an excessive amount of time and cooperation between various groups of individuals. In this respect, Dr.DeLuca and his Associates have been involved with this Congress for the past two years; and the collaboration between Dr.DeLuca as Secretary-General and his Associates in Boston with myself as President of ISEK, The Council of ISEK, and my Associates in Montreal has been outstanding in every respect. In reference to my Montreal Associates, it is my pleasure to single out Mrs. Patricia Larochelle, Administrative Assistant, Department of Research, Rehabilitation Institute of Montreàl. It has been mainly through her efforts, guidance and understanding that we were able from within the Montreal Group to achieve and maintain the effective and excellent collaboration with the Boston Group and the ISEK Council.

This Congress is the Fourth for ISEK. It will, with its published proceedings as the first three have, enhance the stature of ISEK as an International Organization, and, at the same time, allow for meaningful and friendly interaction between friends and colleagues from around the world. The First International Congress was held in Montreal, CANADA in 1968. Drs.John Basmajian (CANADA), Bengt Jonsson (SWEDEN), Sven Carlsoo (SWEDEN) and Thérèse Simard (CANADA) were responsible for the First Congress. The Second Congress took place in Barcelona, SPAIN, 1972; and Drs.Basmajian and Jonsson were the principal organizers. The Third Congress occurred in Pavia, ITALY in 1976. Dr.Paolo Pinelli (ITALY) was the Secretary-General for the Third Congress.

The past three years have been active ones for ISEK. There have been two Regional Meetings, one in North America and one in Southern Europe, plus the planning and organization for this Fourth Congress. Dr.Donald Hobart, (USA) was the Chairman of the North American Meeting, and Dr.Franjo Gracanin, (YUGOSLAVIA) was Chairman for the European Meeting. I wish to take this opportunity, to thank each member of the Council and the Membership for the help and assistance extended to me during these busy three years.

The following is a brief historical overview of ISEK. This has been included for the benefit of new members and non-members who may be interested in joining ISEK. The next two paragraphs have been excerpted from page 3, ISEK Newsletter #1, December, 1967 edited by Dr.Bengt Jonsson, SWEDEN.

"Electromyography has been used to study muscular function and human movement almost from the beginning of modern electrophysiology. However the research was scattered and no sense of community developed among electromyographic kinesiologists until the 1960's (although several large national organizations and conventions had been organized in the related field of clinical EMG). Thus the time was ripe for the formation of an International Society.

Wiesbaden, GERMANY, was the scene in August 8-13, 1965 for the International Congress for Anatomists. A number of electromyographers who were present found this an opportunity to strengthen their personal ties and soon became convinced that they must take some action. At a meeting on August 13 in the Rhein-Main-Halle, they formed the nucleus of ISEK. The organizers were John V. Basmajian (CANADA), Sven Carlsoo and Bengt Jonsson (SWEDEN) and John E. Pauly (USA). Also contributing to the founding meeting were M.A.MacConaill (IRELAND) and Laurence Scheving (USA)."

The response to these initial organizational meetings was excellent over the following two years, resulting in a membership of approximately two-hundred (200)individuals and the First International Congress being held in 1968. In the ensuing years ISEK has increased its membership, and broadened its scope to include areas of electrophysiology other than just electromyography.

The composition of the membership in 1979 includes the Basic Sciences, Health Services, Biomedical Engineering, Dentistry and others; representing thirty (30) countries. The primary objectives are to encourage research, both basic and applied, in electrophysiological kinesiology; to offer training possibilities to interested individuals; and to provide for the dissemination and sharing of research through regional and international meetings, publications in the ISEK Newsletter and appropriate journals. It is anticipated that a more definitive history of ISEK will be published in the Newsletter during the next six (6) months.

On behalf of the Council of ISEK I thank each member and guest for participating in this Fourth International Congress. In the final analysis, it is the individual commitment and participation that renders any congress successful.

HW Ladd

Herbert W. Ladd
President

# CONGRESS SCHEDULE

SUNDAY 5 AUGUST

09:00 Registration Desk Open
(Throughout Congress)

17:00 - Cocktail Reception
19:00

MONDAY 6 AUGUST

09:00 Plenary Session

Official Congress Opening

President's Address

Keynote Address

"The Size Principle: How the Dimension of Motoneurones Influence Their Properties and Those of the Muscle Fibers They Supply."

By E. Henneman

Closing Remarks

12:00 Lunch

## SESSION 1A – TREMOR
(Chair: R.N. Stiles)

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|---|---|---|
| 13:30 | Postural Tremors and Damped Oscillations of the Hand Correlated with Demodulated Extensor EMGs.<br>R. N. Stiles | 2 |
| 13:50 | Suppression of Abnormal Intention Tremor by Application of Viscous Damping.<br>M. J. Rosen, D. E. Dunfee and B. D. Adelstein | 4 |
| 14:10 | The Effects of Running on Physiological Action Tremor in Normal Subjects.<br>P. A. Iaizzo, R. S. Pozos and R. W. Petry | 6 |
| 14:30 | Physiological Action of the Elbow and the Wrist<br>R. S. Pozos, S. A. Burgstahler, P. A. Iaizzo and R. W. Petry | 8 |
| 14:50 | Action Tremor Due to Underdamped Movement with Slowly Decaying Oscillation.<br>M. Hallett | 10 |
| 15:10 | BREAK | |

## SESSION 1B – KINESIOLOGY I
(Chair: B. Jonsson)

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| 13:30 | Electromyographic Features of an Agonist Muscle.<br>N. H. Mayer, G. Wannstedt and H. Rosenholtz | 14 |
| 13:50 | Electromyographic Features of an Antagonist Muscle.<br>G. Wannstedt, N. H. Mayer and H. Rosenholtz | 16 |
| 14:10 | A Kinesiological Study of Shoulder Flexion.<br>P. A. Anderson, D. Hurbon and D. J. Hobart | 18 |
| 14:30 | A Kinesiological Study of Shoulder Extension<br>D. Hurbon, P. A. Anderson and D. J. Hobart | 20 |
| 14:50 | Electromyographic Study of Shoulder Joint<br>N. Ito, R. Suzuki, K. Ishimura amd H. Kuwahara | 22 |
| 15:10 | BREAK | |

## SESSION 2A – NEUROPHYSIOLOGY
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| 15:40 | Reinnervation of the Tibialis Anterior Tibial Muscle of the Rabbit After Grafting of the Sciatic Nerve.<br>P. Pinelli, A. Moglia, M. Zanlungo, G. Sandrini, I. Pagani, and A. Arrigo | 26 |
| 16:00 | Compression Effects on the Compound Action Potential of IN SITU Severed Nerves.<br>L. D. Gilmore, L. J. Bloom and C. J. De Luca | 28 |
| 16:20 | Consequences of Removal of the Inhibitory Effects of Renshaw Cells and Ia Inhibitory Interneurons on the Locomotor Rhythm of Alpha Motoneurons.<br>C. Pratt, L. Jordan and J. Menzies | 30 |
| 16:40 | One Type of Specific Effect of Reticular Activity on Cortical Evoked Potentials.<br>C. Torda | 32 |
| 17:00 | Convergent Visceral (Aortic Occlusion) and Somatic Influences on Thoracic Spinal Neurons.<br>C. A. Ohata and R. D. Foreman | 34 |

## SESSION 2B – MOTOR CONTROL
(Chair: W. R. Powers)

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| 15:40 | Single Motor Unit Control and the Number of Muscle Spindles.<br>H. E. Groendijk, D. Kleppe, P. A. Huijing and P. C. van Wieringen | 38 |
| 16:00 | Postural Control Mechanisms.<br>P. L. Weiss and K. C. Hayes | 40 |
| 16:20 | The Effect of Expected and Unexpected Movement Perturbation on Event-Related Brain Potentials and Muscular Responses.<br>R. J. Lockwood, G. A. Wood and J. G. Jones | 42 |
| 16:40 | Variation of Second-Order Parameters in Human Hand Target Tracking.<br>J. L. Young and R. F. Mayer | 44 |
| 17:00 | The Effects of the TM-Sidhi Program on the Paired Hoffman Reflex Response.<br>R. K. Wallace, D. W. Orme-Johnson, D. E. Jacobs and P. J. Mills | 46 |

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| 09:20 | Electromyographic and Anatomical Study of the Human Tensor Fasciae Latae Muscle. | 52 |
| | E. B. Pare, J. M. Schwartz and J. T. Stern, Jr. | |
| 09:40 | Kinematic and Electromyographic Analyses of the Strength of the Abdominal and Back Muscles. | 54 |
| | M. Fujiwara, H. Miura, K. Namasaki and M. Hasue | |
| 10:00 | EMG Versus Isometric Force and Muscle Length. | 56 |
| | B. S. Hudgins, P. A. Parker and R. N. Scott | |
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## SESSION 3B – PROSTHESES
(Chair: J. Van der Straaten)

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| 09:00 | The Interference Between Electrical Stimulation Applied for Prostheses' Sensory Feedback and Myoelectrical Control Systems. | 60 |
| | C. Almstrom, A. Anani, P. Herberts and L. Korner | |
| 09:20 | A New Concept for Knee Endoprosthesis. | 62 |
| | R. Seliktar, D. Elad and D. G. Mendes | |
| 09:40 | Drive Mechanisms for Limb Orthoses. | 64 |
| | Y. Lozac'h, H. W. Ladd and E. D. Sherman | |
| 10:00 | The Boston Elbow: A Case for Proportional Myoelectric Control. | 66 |
| | T. W. Williams, III | |
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(Chair: H. Broman)

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| | | M. Hagberg | |
| M | 11:00 | Localized Muscle Fatigue in Shoulder Muscles: A Preliminary Study Employing the Spectral Moment Analyzer. | 72 |
| | | P. Herberts, R. Kadefors and H. Broman | |
| R | 11:20 | The Fatigue and Recovery of the Human Diaphragm. | 74 |
| | | F. Bellemare and A. Grassino | |
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| | | L. Nistor, G. Andersson and R. Ortengren | |
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| 11:00 | Studies of the Postural Muscle Activities and the Oxygen Consumption in Juvenile and Adult Patients with Cerebral Palsy. | 82 |
| | Y. Shirai, T. Ito, N. Sano, Y. Nakayama, S. Kota, K. Ii, K. Ishizaka and J. Yamaguchi | |
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| | M. Crivellini, L. Divieti, A. Giannangeli and M. Pissarello | |
| 11:40 | Scoliosis Prognosis by Means of Thermographic Analysis. | 86 |
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| 12:00 | LUNCH | |

### SESSION 5A – FATIGUE II



SUPPLEMENT



### SESSION 5B – CLINICAL II
(Chair: G. Andersson)



17:00 CRUISE OF BOSTON HARBOR

## WEDNESDAY 8 AUGUST

### SESSION 6A – LOCOMOTION I
(Chair: D. A. Winter)



### SESSION 6B – CLINICAL III
(Chair: C. F. McCarthy)

## SESSION 7A – LOCOMOTION II
(Chair: J. Joseph)

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| 10:40 | A Study of the Quadriceps Activity Under Isometric Conditions in the Presence of Pathological Knee Joint Effusion.<br>P. W. Grantham and M. Peat | 138 |
| 11:00 | Technical Provisions in Dysfunction of the Lower Leg Musculature.<br>G. Zilvold, K. Postema and J. L. A. Toornend | 140 |
| 11:20 | Muscle Activation During Gait and Iso Kinetic Knee Movements in Patients with Rheumatoid Arthritis.<br>C. Richards, D. Burke and K. Mullarky | 142 |
| 11:40 | Locomotion Data Extraction Using a Micro-Programmed Processor.<br>M. Slonosky and E. Shwedyk | 144 |
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(Chair: K. C. Hayes)

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| 11:20 | Resonant Characteristics of the Human Body During Forced Oscillations About the Ankle Joint.<br>T. M. Bach, A. E. Chapman and T. W. Calvert | 152 |
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(Chair: G. Rau)

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| 14:10 | Quantitative Electromyography: A Computerized Method for Kinesiological Studies.<br>J. P. Boucher and P. P. Lagasse | 174 |
| 14:30 | A Factor-of-Five Improvement in Myoelectric Signal Processing.<br>N. Hogan and R. W. Mann | 176 |
| 14:50 | The Effect of Electronic Data Processing on the Quantitative EMG Signal.<br>J. V. Danoff and G. Alon | 178 |

15:10 GENERAL BUSINESS MEETING

20:00 MUSCLE FATIGUE SYMPOSIUM

## THURSDAY 9 AUGUST

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(Chair: H. P. Clamann)

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| 09:20 | Effect of Motor Cortex Stimulation on Recruitment Order of Fast and Slow Ankle Extensor Motor Units in the Cat.<br><br>T. Minagawa, S. Morita, S. Kobayashi and H. Niu | 184 |
| 09:40 | Indirect Estimation of Innervation Ratio and Specific Tension of Muscle Units of Cat Tibialis.<br><br>M. D. Binder, J. Crockett-McDonagh, R. M. Reinking and D. G. Stuart | 186 |
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(Chair: G. C. Agarwal)

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| 09:20 | Modelling of the EMG Response to Applied Torque Perturbations of Human Forearm Position.<br><br>J. R. DuFresne and J. F. Soechting | 194 |
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(Chair: F. Gracanin)

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| 11:00 | Patterns of MU Activity in the Course of a Movement.<br><br>B. Maton and S. Bouisset | 204 |
| 11:20 | Dynamic and Static Components of Motoneurone Activation in Man.<br><br>M. Mirsky | 206 |
| 11:40 | Single-Motor-Unit Electromyographic Analysis of Fundamental Frequency Control During Phonation.<br><br>T. Baer | 208 |
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(Chair: J. V. Basmajian)

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| 11:00 | Implications of Variability in Spasticity Observed During EMG Feedback Training.<br><br>G. A. DeBacher and J. V. Basmajian | 214 |
| 11:20 | EMG Biofeedback Instrumentation for Clinical Use.<br><br>M. D. O'Riain, H. W. Ladd and E. D. Sherman | 216 |
| 11:40 | Design and Evaluation of a Clinically-Oriented Myofeedback Device.<br><br>C. F. McCarthy, R. S. Le Fever, A. L. Oslan and C. J. De Luca | 218 |
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SESSION 11A – LOCOMOTION IV
(Chair: M. Kumamoto)

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| 13:50 | The Use of Impulse Momentum for Quantification of Gait Disorders with the Aid of Force Plates.<br>R. Seliktar | 224 |
| 14:10 | Motor Evaluation Procedure for Testing Hemiplegic Locomotion.<br>C. Boccardi, C. Frigo and A. Pedotti | 226 |
| 14:30 | Gait Timing in Normal and Hemiparetic Children and Adults.<br>V. Alfieri, L. Grillo and R. Merletti | 228 |
| 14:50 | Gait Analysis of the Leg After 'Charnley' Knee Anthroplasty.<br>R. J. Minns and K. Hardinge | 230 |
| 15:10 | Fractionated Knee Extension Response Time, Reflex Time, and Fiber Type Composition in Relation to Age and Activity Level.<br>P. M. Clarkson, W. Kroll and A. Melchionda | 232 |

SESSION 11B – KINESIOLOGY IV
(Chair: J. Vorro)

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| 13:50 | Frequency Spectrum of the Human Myoelectric Signal as Related to Handedness.<br>R. P. Lehr, Jr. and C. M. Hoskiko | 238 |
| 14:10 | Effects of Forced Exercise on Normal and Myopathic Hamsters: An Electromyographic Study.<br>J. Vorro, S. Herman, R. Hallgren and R. Carrow | 240 |
| 14:30 | A Model for the Analysis of Skeletal Muscle Functions.<br>P. Scholten and W. Vuisting | 242 |
| 14:50 | Multidimensional Correlation of Muscle Properties.<br>E. F. Murphy | 244 |
| 15:10 | Electromyographic Evaluation of Mopping and Swabbing in Cleaning Work.<br>J. Winkel, B. Ekblom, M. Hagberg and B. Jonsson | 246 |

17:15 GALA BANQUET

## FRIDAY 10 AUGUST

10:00 DISCUSSION OF PROPOSED STANDARDS

# TREMOR

## Session 1A

POSTURAL TREMORS AND DAMPED OSCILLATIONS OF THE HAND CORRELATED WITH DEMODULATED EXTENSOR EMGs.

Robert N. Stiles, Dept. of Physiology and Biophysics, University of Tennessee Center for the Health Sciences, Memphis, Tennessee.

## INTRODUCTION

Studies of postural tremors in man and experimental animals have led to at least five major hypotheses for the tremor mechanism. These are the: 1) Central Oscillator; 2) Reflex Oscillator; 3) Mechanical Resonance Oscillator; 4) Mechanical Filter (low-pass of high-frequency input); and 5) the Mechanical-Reflex Oscillator hypothesis. The reduction of tremor frequency with added mass, and the presence of damped (die-away) oscillations of the limb in response to external disturbances, provide major support for hypothesis No. 3. Amplitude modulation at the tremor frequency of the surface EMGs of muscles controlling the limb, and the absence of effect of added mass upon the tremor frequency, provide major support for hypotheses No. 1 and 2 (and, perhaps, No. 4). However, recent evidence (Stiles, 1976) suggests that combined mechanical resonance and neural feedback factors (supporting No. 5) may largely determine the frequency and amplitude of postural hand tremor.

In testing hypothesis No. 5, the following two relations are expected for the neuromuscular-load system involved in hand tremor: 1) Under similar conditions of added mass and displacement amplitude, the damped oscillations resulting from external disturbances should occur at approximately the same frequency as that of the hand tremor; and 2) If tremors of a certain displacement amplitude are accompanied by amplitude modulation of wrist extensor EMGs, then damped oscillations obtained under similar conditions should also be accompanied by similar EMG modulation.

The major purposes of this study are: 1) Determine the frequency and displacement amplitude (root-mean-square, or rms, amplitude) of postural tremors and damped oscillations of the hand under conditions of equal added mass and of rest between recordings of the muscles controlling the hand; and 2) Determine the presence or absence of correlated EMG modulation for each of these two kinds of oscillations.

## EXPERIMENT

Postural tremors and damped oscillations of the hand were detected from each of seven normal human subjects with a small accelerometer. Each subject was seated with the forearm pronated and supported to the wrist in a horizontal position by a solid cradle mounted upon a short, heavy table. Each subject extended his fingers, and the thumb and fingers (including the metacarpal region of the hand) were held tightly between two sheets (forming a sandwich) of plastic material (100 g). Each subject was asked to keep the mean position of the hand approximately horizontal throughout each 16-s digitization period. Damped oscillations of the subject's hand were produced by applying a vertical pulse of force (tapping) repetitively (about one/s over 16 s) to the dorsal aspect of the outstretched hand. Hand oscillations in the vertical plane were detected using a Gulton AVR-250 accelerometer (2.5 g) taped to the upper plastic sheet at 16 cm from the subject's wrist. In addition to hand acceleration records, two bipolar surface EMG records (from the extensor digitorum communis, EDC, and the extensor carpi radialis longus, ECR, muscles) were obtained simultaneously. Amplification was obtained using Tektronix (model 122) preamplifiers and locally constructed DC amplifiers. Any motion artifact in the EMG signals was eliminated by the use of the 8-Hz high-pass filter setting on the preamplifiers and an additional high-pass digital filter. The low-frequency cut-off of the high-pass digital filter was approximately 50 Hz. The acceleration and EMG records were obtained under conditions of approximately 90-s of rest of the muscles controlling the hand before each 16-s run. From 4 to 20 records (generally 5 records) each of tremor and of damped oscillations of the hand (and associated EDC and ECR EMG records) were obtained for analysis from each subject.

## DATA ANALYSIS

Each 16-s record of the voltage analog of hand acceleration was digitized by a PDP-12 computer at a rate of 64/s, and the mean was computed and subtracted from these 1,024 digital values. Calculation of power spectra (auto-spectra) resulted in 65 spectral values between 0 and 32 Hz. In general, each spectrum exhibited one major frequency band, and a single number representing the frequency of the peak of this band was computed. The EDC and ECR EMGs (16-s records) were digitized at 1,024/s, estimates of the means were computed and subtracted from the data, and then these deviations were full-wave rectified and smoothed (16 nonoverlapping values were averaged). This process results in amplitude demodulation of the EMG signals and gives an effective sampling rate of 64/s. Power spectra (0 to 32 Hz) were then computed from these 1,024 values. For the analysis of these acceleration and EMG data, cross-covariance as well as cross-spectral functions were calculated. Finally, coherence values were computed from the combined auto- and cross-spectral values.

## RESULTS

For the seven subjects, the average rms displacement amplitude for the postural tremors and damped oscillations ranged from 25 to 91 micra and from 143 to 434 micra, respectively. In general, the rms displacement of the damped oscillations was larger than that for the tremors, depending in part upon the magnitude of the pulse forcing of the subject's hand. For six of the seven subjects, the average frequency of the damped oscillations was slightly lower (a maximum difference of about 1.5 Hz) than the average frequency of the tremor for the same subject. For both the damped oscillations and tremors, a band in the EMG spectra at the frequency of the hand oscillations was consistently found for both the EDC and ECR muscles. In addition, the variance of the EMG band correlated with the damped oscillations was consistently greater (about 1.5 to 6.0 times greater) than the variance of the EMG band correlated with the tremor. Average coherence values between the demodulated EMG's and damped oscillations (calculated at the frequency of the hand motion) ranged between 0.50 and 0.70, as compared to a range of 0.25 and 0.70 for the demodulated EMGs and tremor. The EMG spectra associated with both the tremor and damped oscillations generally contained another major band in addition to the one at the frequency of the hand oscillations. This band usually had its peak frequency between about 10 and 20 Hz, and appeared not to be a harmonic of the spectral band that occurred at the frequency of the hand motion.

## DISCUSSION

The results of this study show two important similarities between the tremors and the damped oscillations of the hand: 1) Both kinds of oscillations occurred at approximately the same frequency. While there was a small difference in the frequencies of these two kinds of oscillations, this may have been largely due to the difference in their rms displacement amplitudes. 2) Each kind of hand oscillation was accompanied by correlated amplitude modulation of the EMGs of both EDC and ECR muscles at the frequency of the hand motion. This amplitude modulation of the EDC and ECR muscle activity can be interpreted as having resulted from neural feedback within the particular skeletal motor servosystem. This interpretation is supported by the finding (Stiles, 1976) that the frequency of this modulation decreases with the frequency of the tremor for different amounts of mass added to the hand. Additional support is given by the presence of the prominent band in the EMG spectrum at the frequency of the damped oscillations of the hand.

The relatively high coherence between the extensor EMG modulation and the damped oscillations of the hand is of particular interest relative to the role of neural feedback in the tremor mechanism. This finding suggests that the presence of muscle activity which is correlated with small displacement oscillations of a limb does not necessarily indicate that a reflex mechanism (positive feedback mechanism) is responsible for the generation of these oscillations. Therefore, the exact role that neural feedback plays in the tremor mechanism appears uncertain. It may be that the effects of neural feedback on hand tremor are combined with the effects of muscle-load mechanics (a mechanical-reflex oscillator mechanism) in determining the frequency and amplitude of these oscillations.

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## ACKNOWLEDGMENT

This study was supported in part by Research Grants NS-08692 and NS-14730 from the National Institutes of Health.

# SUPPRESSION OF ABNORMAL INTENTION TREMOR BY APPLICATION OF VISCOUS DAMPING

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## INTRODUCTION

Many neuromuscular conditions produce functional disability despite useable levels of residual strength and voluntary control because of "intention tremor" of such large amplitude as to mask intended movement and constitute a hazard to the patient. The ineffectiveness of current therapeutic modes for suppressing abnormal tremor without attenuating willed motion led to the present work.

We hypothesize that abnormal tremor may be reduced to a clinically significant degree by application of appropriate mechanical loads across afflicted joints. Theoretical and experimental support for this concept may be found in the work of Stein[1], Chase[2] and others. The long range goal of this project is the design of a practical multi-degree-of-freedom orthosis or fixed-base compliant restraint system.

## EXPERIMENT

A transduction and loading device, TLD, pictured in Figure 1, restrains the subject's forearm while allowing normal wrist flexion and extension. A "handle" passively coupled to the hand permits goniometric measurement of wrist angle and application of torque generated by a magnetic particle brake controlled electronically as a function of wrist angle. Present control circuitry differentiates goniometer output producing a viscous load with externally controllable damping constant up to 5 (in-lbf)/(rad/sec). The TLD also incorporates a torque transducer to measure the applied brake moment.

Subjects are asked to perform a pursuit tracking task by matching target and wrist angle traces on a CRT. To date, target functions have been ± 28° sinusoids at frequencies between 1/8 Hz and 3/4 Hz. At each tracking frequency, trials were performed with no load and with five values of damping. The recorded wrist angle signal from each 3 min trial was processed to extract ensemble average tracking performance and an RMS tremor (between 2 and 20 Hz) index. Five subjects have been tested so far, three with neuropathologies and two normals.

Fig.1: Photo of TLD.

## DATA

The average undamped tremor of our abnormal subjects was 4.76° ± 1.86° with individual tremor excursions of > 30° common. The value of the normals' tremor averaged over an identical set of undamped trials was 0.42° ± 0.19°. The abnormal subjects (whose etiologies were diverse - mid-brain stroke, MS-like degenerative disease, and idiopathic essential tremor) all showed approximately sinusoidal average tracking in undamped trials even when this intended movement was almost unrecognizable in raw data records. Average tracking amplitude (relative to target amplitude) was ∿ 75% without damping.

Figure 2 displays representative data collected during two trials from abnormal subject J.F. at a tracking frequency of 1/4 Hz. The dramatic decrease in tremor with applied damping of D = 6 (60% of equipment maximum) is apparent. The average tremor attenuation factor at D = 10 for all three abnormals was 0.27 ± 0.19 with attenuation statistically significant at better than 0.1% for $4 \leq D \leq 10$. Normalized tremor data for these subjects is displayed in Figure 3 as a function of D. In addition, as D was increased to 10, no deterioration of average tracking amplitude resulted. An apparent increase in average tracking fidelity was measured in all three abnormals as damping was increased in the range $2 \leq D \leq 6$.

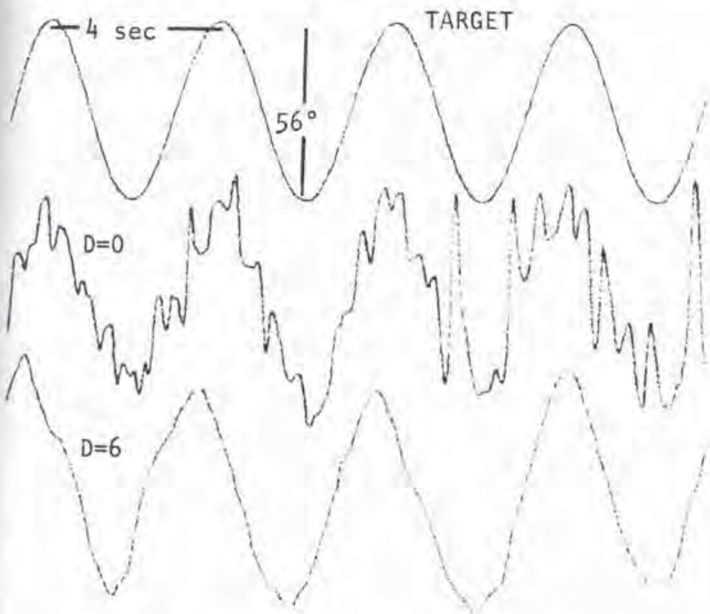


Fig.2: Raw tracking performance of subject JF.

## DISCUSSION AND CONCLUSIONS

While the subjects tested to date are too few in number and heterogeneous in clinical type to permit conclusion of clinical significance, there can be no doubt as to the uniformity of our results. Viscous damping does significantly attenuate tremor in our subjects, in some trials to levels not significantly greater than normal. Further, this result is obtained at values of damping which do not attenuate the intended component of movement. Note that 1/2 Hz 28° sinusoids involve angular slew rates of 90°/sec. - a speed representative of many practical activities. Note too that our working assumption that tremor-disabled people are capable of voluntary control which is masked by tremor was validated by the undamped <u>average</u> tracking performance of our subjects.

As our work continues, testing and data-processing techniques will be expanded in several ways. Psuedo-random target functions and movement in a horizontal plane will be incorporated to extend the generality of our results. Flexor and extensor EMG will be processed to determine the effect on tremorogenic EMG modulation of the applied load and to draw conclusions concerning its effect on muscle fatigue. Variation of tremor power spectra with damping will be used to test alternative models of tremor mechanism.

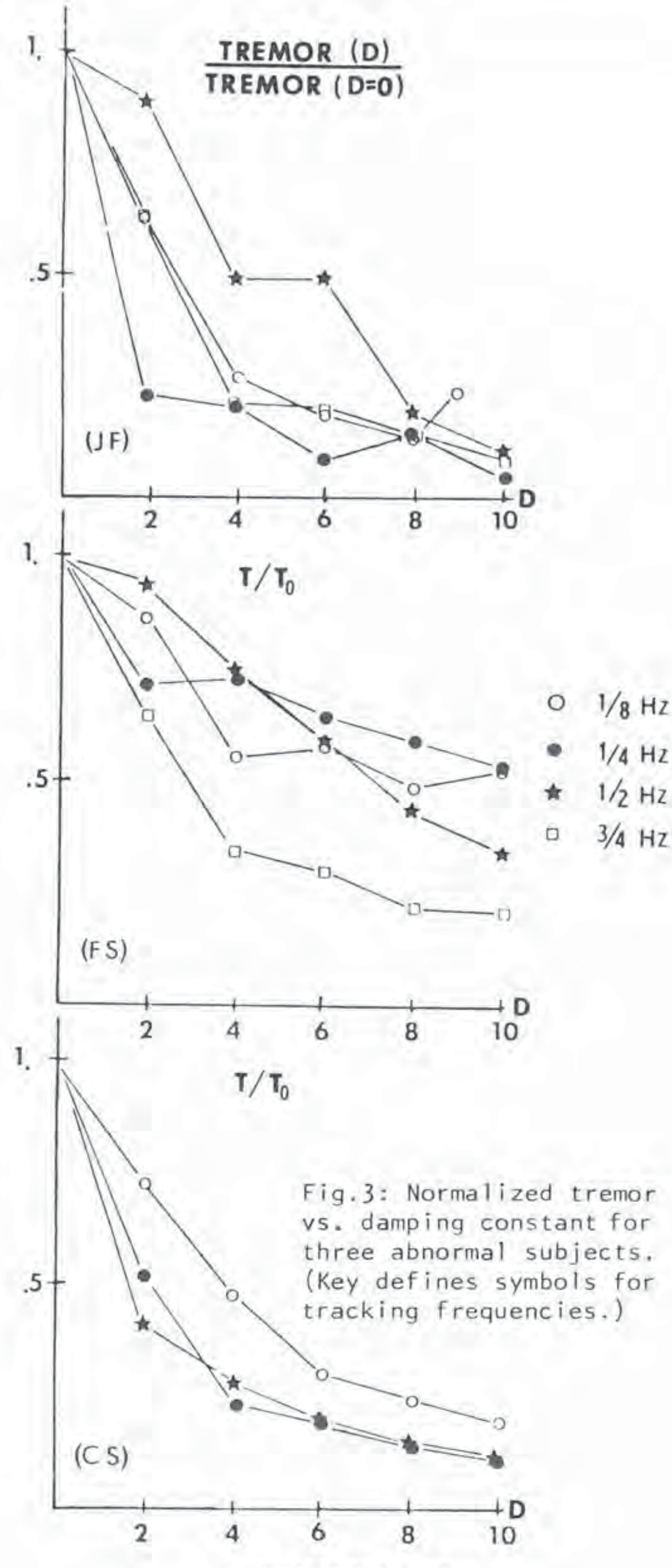


Fig.3: Normalized tremor vs. damping constant for three abnormal subjects. (Key defines symbols for tracking frequencies.)

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## ACKNOWLEDGMENT

This investigation has been carried out as part of the Harvard-M.I.T. Rehab. Engineering Cntr. Project, supported in part by Research Grant No. 23P 55854/1 from the Rehab. Services Admin., Dept. of H.E.W., Washington, D.C.

THE EFFECTS OF RUNNING ON PHYSIOLOGICAL ACTION TREMOR IN NORMAL SUBJECTS

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## INTRODUCTION

A tremor which occurs during voluntary motion in normal non-fatigued subjects has been recently documented (1, 2). This tremor is called Physiological Action Tremor (PAT) and is of high amplitude and of a similar frequency range to that of pathological situations. The present study tested how fatigue from running altered PAT. Unlike previous studies, the total body is fatigued rather than the one limb on which measurements were taken.

## EXPERIMENT

Thirteen normal subjects (6-45 years of age) were used in both parts of the study. The first part of the study consisted of gathering control data. The subjects were seated comfortably in a chair so that their lower legs were perpendicular to the floor and their knees formed right angles. Two sets of bipotential surface EMG electrodes were placed on the soleus and anterior tibialis muscles of each subject's dominant leg. The EMG signals were amplified by a tektronix low level preamplifier and a locally constructed D.C. amplifier. In addition an AVR-250 accelerometer was taped just proximal to the patella and was used to detect vertical motion (cm/sec$^2$). Once the recording instruments were in place the subjects elevated their heels off the ground, while the balls of their feet made contact with the floor, for a 20 second period (Heel Elevated Tremor). Next they were asked to raise and lower their heels while again keeping the balls of their feet on the floor. This one continuous voluntary motion was repeated for a two to three minute period. At a consistent position of this maneuver, a higher amplitude acceleration signal is present, Physiological Action Tremor. All of the data was recorded for later analysis on a PDP-12 digital computer.

In the second part of the study the same subjects were put through the above procedures immediately after they finished running. The length of the run varied with the individual; he or she (4 female subjects) was asked to run until they were fatigued.

## DATA ANALYSIS

The recorded analog accelerometer signals were transformed to digital values by playing the recordings through the PDP-12 analog-to-digital converter while an interrupt driven machine language program sampled the signal at a computed rate. The digitized signals were displayed, and stored on the disk. These sampled values were then retrieved and displayed on a graphic screen at which time the operator selected out desired portions of the tremor type of interest. Once all desired segments were chosen, they were stored. A Fortran IV program was engaged to join the segments together, forming a resultant continuous record which was stored for analysis.

Spectral analysis was done by a Cyber 171, to which the PDP-12 was directly linked. The analysis was written in Fortran IV and conforms exactly to the algorithms for autocorrelation and power spectra, developed by Bendant and Piersol (3). RMS acceleration values were also computed. A printout of a plot of variance vs. frequency, and the RMS acceleration values were provided for each subject.

A one way analysis of variance test was then performed between each tremor type on the RMS acceleration for the subject pool. Also the means and standard deviation of the observed major frequencies of each tremor type was computed. A frequency was considered major if it was the maximum variance peak or had a value of one half that of the maximum.

## RESULTS

Three tremor types are dealt with in this study: heel elevated tremor (HET), physiological action tremor (PAT), and the PAT effected by running, which is termed fatigue altered action tremor (FAAT). The major frequencies of the three tremor types all fall into the same range of 3-8 Hz. The mean, variance, and standard deviation of the frequency ranges for each tremor type are as follows: HET ($\bar{X}$ = 5.6 Hz, $\sigma^2$ = 4.48, $\sigma$ = 2.12), PAT ($\bar{X}$ = 5.6 Hz, $\sigma^2$ = 2.36, $\sigma$ = 1.53) and FAAT ($\bar{X}$ = 6.5 Hz, $\sigma^2$ = 2.71, $\sigma$ = 1.65).

Although the frequency ranges are similar, the amplitudes of the RMS accelerations are significantly different. Fig. 1 shows how they vary for each subject. The mean pooled RMS accelerations for each tremor type are FAAT 16.4 cm/sec$^2$, PAT 12.0 cm/sec$^2$, and HET 3.6 cm/sec$^2$. The Duncan, Tukey-HSP, and Scheffe procedures of one way variance tests all indicate that each tremor type is significantly different from the others at the .050 level.

A definite correlation between the EMG signals of both muscles and motion existed.

## DISCUSSION

Fatigue induced by running altered the amplitude and frequency of PAT. It is yet uncertain if the increase in acceleration is due to alterations of the sensitivity of reflex loops, a change in the physical properties of the muscles, or a combination of these and/or other factors. If mechanical changes alone were responsible for the increased acceleration, a decrease in the mean frequency might be expected. The slight increase in mean frequency and higher RMS acceleration values for FAAT as compared to PAT may indicate a change in the neural input to the limb. Mechanism(s) for the production of PAT is still uncertain. However the effects of fatigue produced by running on PAT may provide some insight to what the controlling factors are. An understanding of PAT is important because relationships and similarities between PAT and pathological tremors (clonus) have been reported (1, 2).

## CONCLUSION

When a normal person runs and becomes fatigued, there is a change in the mechanism controlling PAT. This is concluded from the observed increase in RMS acceleration while the frequency range (3-8 Hz) is maintained.

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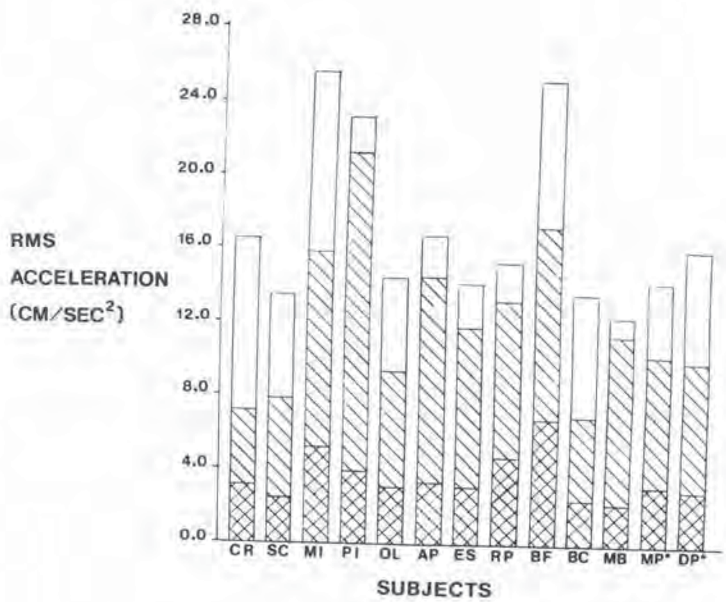


Fig. 1 RMS acceleration of each tremor type for each individual tested. Heel elevated tremor (HET) is represented by ▩, pathological action tremor (PAT) by ▧, and fatigue altered action tremor (FAAT) by □.

(*ages 7-11 years)

# PHYSIOLOGICAL ACTION OF THE ELBOW AND THE WRIST

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## INTRODUCTION

It has previously been noted that a high frequency oscillation occurs during low frequency voluntary motion (1, 2). In this study tremors of the wrist and elbow joint were analyzed during voluntary motion to gain further insight into the relationship between the high frequency oscillation and concomitant EMG activity. Understanding this relationship could lead to possible insights into various pathological situations, i.e. Parkinsonian hand tremor and cogwheel rigidity.

## EXPERIMENT

Sixteen normal subjects ranging from 7-60 years of age were involved in this study. Nine individuals were observed for forearm tremor and ten for hand tremor. In the forearm tremor study, subjects were observed while standing with an accelerometer (AVR-250) taped to a small aluminum splint 25 cm from the proximal end of the ulna. While the upper limb was in the anatomical position, the subjects were instructed to extend and flex the elbow, maintaining the wrist stable. In addition to acceleration, EMG recordings were also taken from surface electrodes placed over the belly of the biceps muscle.

Records of acceleration of the hand were taken by taping the accelerometer to a rigid styrofoam sandwich which helped to minimize finger tremor. The subjects were seated and instructed to extend and flex their wrist with the forearm supported. Surface EMG's were recorded from the extensors and flexors of the wrist.

## DATA ANALYSIS

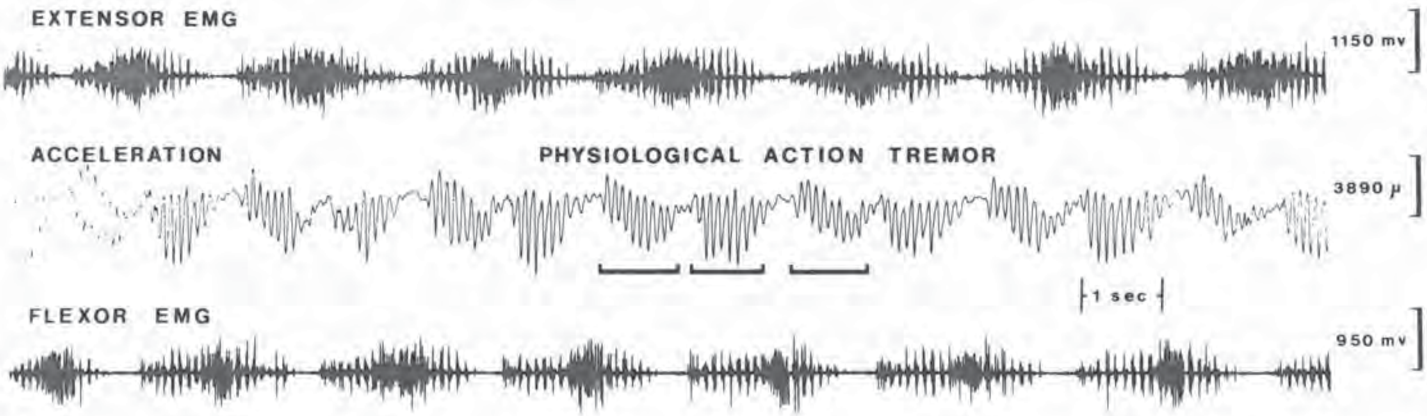
The analogue data was digitized on a PDP-12 computer (Digital Equipment Corp.) linked to a Cyber 171 (Control Data Corp.). The data was sampled at a rate of 60 pts/sec and low pass filtered at 30 Hz. The low frequency associated with voluntary motion (< 2 Hz) was removed by low band pass filtering. Since the high frequency oscillation occurred periodically in the record, a program was written to analyze selected segments of the data. Plots of variance vs. frequency were printed for each subject and RMS acceleration values were computed at the subject's major tremor frequency. Peaks on the plots of variance vs. frequency were considered significant if they were at least ½ the value of the maximum variance peak on a given plot.

## RESULTS

Analyses of both hand and forearm tremor show that a high frequency oscillation accompanies voluntary motion. Since these oscillations occur during voluntary motion they have been called physiological action tremor (PAT). It should be stressed that PAT only occurs at certain positions during the movement of the limb (Fig. 1). Both the forearm and hand showed major peaks of PAT at 2-4 Hz and 8-12 Hz. A majority of the maximum acceleration values of the forearm occurred at 2-4 Hz and those of the hand at 8-12 Hz. Analysis of EMG activity vs. motion revealed a strong correlation between EMG activity and PAT. Distinct bursts of EMG activity were readily evident (Fig. 1).

## DISCUSSION

The presence of PAT during voluntary motion which occurs over a wide range of limb movement substantiates previous reports (1, 2). In 1956, Marshall & Walsh (2) reported on a "physiological tremor" associated with voluntary motion but reported on no synchronization of EMG activity with the tremor. The difference between our results could be due to the increased sensitivity of our recording system. Stiles reported on a high frequency oscillation which accompanied voluntary motion of the hand but in his experimental design the hand was moving less than $10^{\circ}$ whereas in our experiments the hand is moving over $120^{\circ}$ (1).




PAT may be a position dependent amplification of the high frequency reported by Stiles.

The presence of two major frequency peaks may suggest more than one mechanism is responsible for these tremors (3). Since PAT can be detected in the upper limb, it may be altered in certain pathological states like cogwheel rigidity.

## CONCLUSION

In this study, 2 bands of high frequency oscillation (2-4, 8-12 Hz) were recorded in normal subjects who were voluntarily oscillating their upper limb at 0.5-1.0 Hz. This high frequency oscillation which was most evident at certain positions was called physiological action tremor (PAT). Synchronized EMG recording accompanied PAT in all cases.

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Fig. 1 Example of physiological action tremor of hand (area above bar). Motion was detected with an accelerometer taped to the hand while the wrist was extended and flexed. Surface EMG's show electrical activity correlated with physiological action tremor.

ACTION TREMOR DUE TO UNDERDAMPED MOVEMENT WITH SLOWLY DECAYING OSCILLATION.

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## INTRODUCTION

Most cases of significant action tremor encountered clinically are due to essential tremor which is almost always characterized electrophysiologically by synchronous bursting in antagonist muscles. Alternating bursting in antagonist muscles is unusual. The patient to be described here had a tremor with alternating bursting for which the mechanism seems apparent.

## CASE REPORT

A 50 year old man presented with action tremor involving arms, legs and head. He had been in an auto accident 20 years before which resulted in a basilar skull fracture and loss of consciousness for two days. Subsequently he developed a mild spastic quadraparesis, dementia and the tremor which was made worse by anxiety. A CT Scan showed only mild cortical atrophy. His tremor was dramatically improved with propranolol 30 mg orally three times a day.

## PHYSIOLOGICAL OBSERVATIONS

EMG was recorded with surface electrodes from biceps and triceps. There was a 6 Hz tremor, much worse with action, with alternating bursts in the two muscles. Voluntary monophasic ballistic elbow flexion movements were underdamped with slowly decaying oscillation (Fig. 1). The initial agonist burst length was often prolonged compared to normal.

## DISCUSSION

In making ballistic movements there is a burst in the agonist muscle to start the movement and a subsequent burst in the antagonist to help stop it (Hallett and Marsden, 1979). Normal movements may overshoot slightly, but are quickly damped. The underdamping shown by this patient leads to tremor since each movement in the normal flow of successive primary movements and corrective movements sets up a new oscillation. It should be noted that this type of tremor is clearly different from tremor-at-rest in Parkinson's Disease where there are alternating bursts in antagonist muscles, but where the tremor disappears with voluntary movement. Prolongation of the first agonist burst to this degree in a ballistic movement is suggestive of a deficit in cerebellar function (Hallett, et al., 1975).

## SUMMARY

A patient is presented with an action tremor characterized by alternating bursts in antagonist muscles which seems to be due to underdamping of voluntary movement. At least in this circumstance the abnormality is associated with signs of cerebellar malfunction.

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SEE FIGURE ON NEXT PAGE

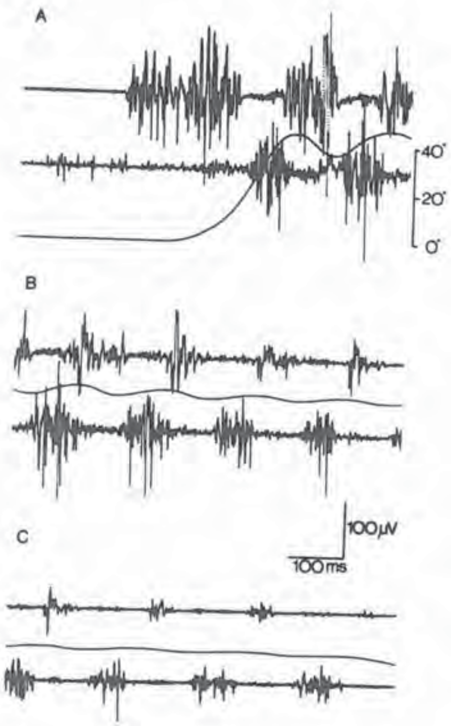



Fig. 1 Attempted 40$^{o}$ rapid elbow flexion movement from initial angle of 120$^{o}$. Parts A, B and C are a continuous record. The top trace is EMG of biceps, the second trace is EMG of triceps and the third trace is elbow angle.

NOTES

# KINESIOLOGY I

## Session 1B

ELECTROMYOGRAPHIC FEATURES OF AN AGONIST MUSCLE

Nathaniel H. Mayer, Gunilla Wannstedt, and Henry Rosenholtz, Rehabilitation Training Center #8, Dept. of Rehabilitation Medicine, Temple U. Health Sciences Center, Phila, Pa.

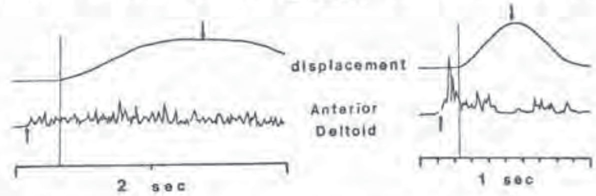
In 1926, Wachholder and Altenburger described different EMG patterns for an agonist muscle depending on movement velocity. Many investigators have subsequently confirmed their observation that slow movement is associated with a continuous pattern of agonist activity while fast movement is featured by a burst of EMG which gives way to a period of silence before maximum displacement occurs.[1] Recent studies have examined burst duration under conditions of fixed amplitude and without regard to variations in movement time.[2] This study explores electromyographic features of an agonist muscle when movement time and movement amplitude are allowed to vary.

METHOD

A total of 483 movements in four normal men (ages 21-34) were studied. While sitting with arm vertical, elbow at 90$^{o}$ and forearm in neutral, each person gripped a pivoted rod which had a potentiometer mounted at its point of rotation to measure movement displacement. Subjects were asked to move the rod forward at widely differing self-generated movement speeds, all the while being careful to terminate forward movement at specified visual targets alongside the rod. Surface EMG recordings were obtained from the anterior deltoid by using an Ag-AgCl electrode with a built-in pre-amplifier to eliminate wire noise. The raw EMG signal was band passed filtered, rectified full wave, and subsequently fed to a PDP 11 computer (sampling rate = 100 Hz) along with the displacement signal. An interactive computer program enabled us to specify the onset and termination of EMG activity and the onset and peak of movement displacement. The computer then calculated maximum velocity and amplitude of displacement, movement time (interval between onset and peak displacement), and the duration of EMG bursts which were observed to end before maximum displacement had occurred. Distinction between burst patterns and continuous EMG patterns were made by visual inspection of each record.

RESULTS

Fig.1 illustrates the two basic agonist features observed in this study: (1) a continuous pattern where EMG activity begins prior to displacement and continues through the peak of movement; and (2) a burst pattern where EMG activity begins prior to displacement but gives way to a silent period before the movement peaks.




Independently of amplitude, burst patterns appeared to be associated with short movement times, the majority of bursts occurring at movetimes less than 450 msec (See fig. 2).

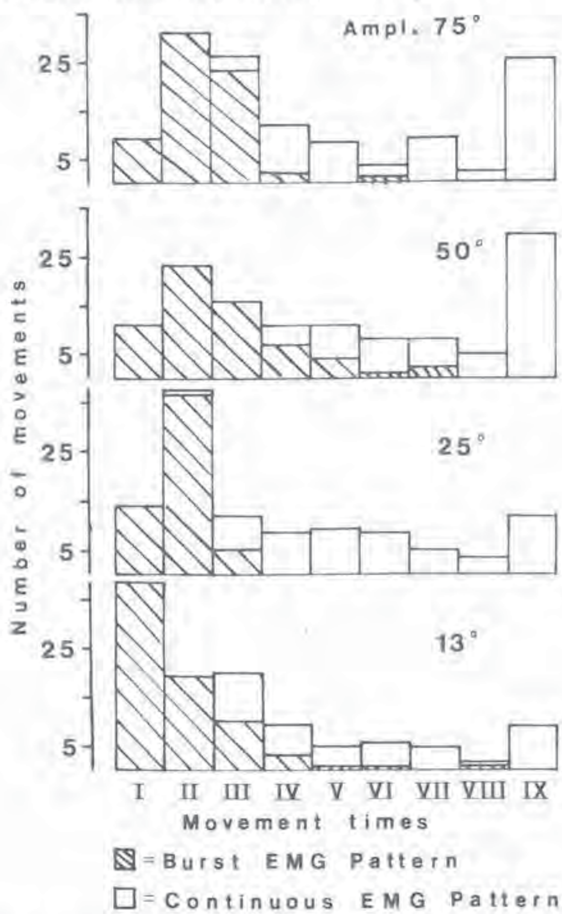



Fig.2 Histogram of burst pattern and continuous pattern frequency as a function of movement

time intervals. Bin I includes all of movement times from 0 through 150 msec; II=151-300; III=301-450; IV=451-600; V=601-750; VI=751-900; VII=901-1050; VIII=1051-1200; IX=>1201msec.

The continuous pattern was associated with longer movement times and only 1 out of 182 trials with movement times <300 msec showed a continuous pattern. Burst duration time appeared to vary directly with movement time as illustrated in figure 3. Quick movements with shorter movement times had smaller burst durations (across all amplitudes). At any one amplitude, the

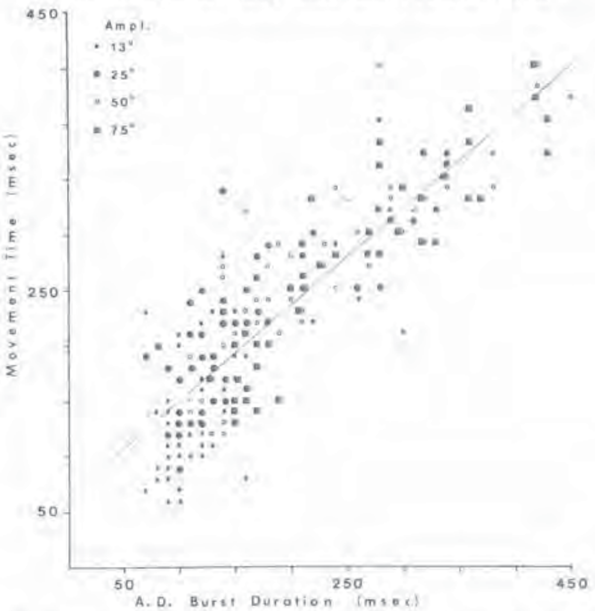



Fig.3 Magnitude of the burst duration of anterior deltoid versus movement time across all movement amplitudes. The best fit straight line has a correlation coefficient of .81. Note that the shortest burst durations are associated with the smaller amplitudes and vice versa.

number of burst patterns increased as the mean maximum movement velocity (V max) increased (compare Table 1 with fig.2). It should be noted, however, that the magnitude of V max was larger in many cases of continuous patterns compared to burst pattern values (compare V max "burst" at 13° amplitude with V max "continuous" at 75° in Table 1).

## DISCUSSION

Many papers in the literature have related burst patterns to movement velocity. Velocity, of course, is a measure relating distance and time. This study suggests that burst patterns are more tightly associated with movement time than velocity. The duration of burst activity varies linearly with movement time (fig.3) even though amplitudes are quite different. The majority of burst patterns occur at movement times <450 msec while much overlap of V max magnitudes exists between continuous and burst patterns (Table 1). Fitts pointed out the close relation between speed, amplitude, and acuracy.[3] When movements do not have to be terminated precisely, movement time is essentially independent of its extent. However, when amplitude accuracy is specified (as in this study), movement time and movement amplitude must be centrally linked. The relationship between agonist burst duration and movement time may reflect this linkage strategy. Velocity may be of secondary importance to what may really be a temporal motor control strategy based on movement time for rapid, amplitude specified displacements.

TABLE 1: MEAN MAXIMUM VELOCITIES (°/SEC) - FOUR AMPLITUDES - DIFFERENT MOVEMENT TIME BINS

| AMPL.=75° | I | II | III | IV | V | VI | VII | VIII |
|---|---|---|---|---|---|---|---|---|
| CONTINUOUS EMG | – | – | 319 n=3 | 249 n=9 | 195 n=7 | 156 n=3 | 152 n=10 | 127 n=2 |
| BURST EMG | 872 n=9 | 520 n=32 | 347 n=21 | 254 n=3 | – | 176 n=1 | – | – |
| AMPL.=50° | | | | | | | | |
| CONTINUOUS EMG | – | – | – | 156 n=4 | 137 n=7 | 112 n=7 | 104 n=6 | 94 n=5 |
| BURST EMG | 586 n=11 | 361 n=30 | 219 n=16 | 178 n=7 | 132 n=4 | 137 n=1 | – | – |
| AMPL.=25° | | | | | | | | |
| CONTINUOUS EMG | – | 156 n=1 | 114 n=7 | 95 n=8 | 74 n=9 | 76 n=5 | 59 n=3 | 59 n=2 |
| BURST EMG | 371 n=14 | 198 n=38 | 121 n=5 | – | – | – | – | – |
| AMPL.=13° | | | | | | | | |
| CONTINUOUS EMG | – | – | 78 n=3 | 59 n=6 | 49 n=4 | 47 n=5 | 43 n=5 | 39 n=1 |
| BURST EMG | 208 n=38 | 108 n-19 | 67 n=5 | 59 n=2 | 49 n=2 | 39 n=1 | – | 39 n=1 |

(Time bin values are the same as in fig.2)

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## ACKNOWLEDGEMENT


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ELECTROMYOGRAPHIC FEATURES OF AN ANTAGONIST MUSCLE

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## INTRODUCTION

Charles Beevor in his Croonian Lectures on Muscular Movements (1903) defines antagonists of a movement as "Those muscles which move the joint in a direction which is diametrically opposed to this movement".[1] Some authors have described antagonist muscles as being elect romyographically relaxed during movement, acting only at the end of the whip-like movement, or active only as a sign of increased tension or inept skill.[2] Others have described and measured timing features of alternating agonist and antagonist EMG patterns during "fast movement" without examining corresponding kinematic variability.[3] This study was undertaken to explore the action of an antagonist muscle when movement time and movement amplitude are allowed to vary.

## METHOD

Four normal men participated in the study. Sitting with the arm vertical and elbow at $90^{o}$, each subject gripped a lightweight lever which he was instructed to move forward to specified visual targets representing amplitudes of $13^{o}$, $25^{o}$,$50^{o}$ or $75^{o}$. Movements were made over a wide range of self-generated velocities. A potentiometer was mounted at the rod's rotation point to enable measurement of amplitude and average rate of movement. Surface EMG recordings were obtained from the anterior and posterior deltoid muscles because moving the lever forward required shoulder flexion of the humerus as agonist and antagonist respectively. The recording electrodes were silver-silver-chloride, contained in a package 2 cm apart with pre-amplifiers included for elimination of wire noise during movement. The EMG signal was band-pass filtered, full wave rectified and recorded on magnetic tape along with the displacement signal from the potentiometer.

An interactive computer program was used to specify the times of onset and the termination of EMG activity and the times of onset and peak of movement displacement. The estimated observer reading error for this program was less than $\pm$ 15 ms for the scaling factors selected. The computer then calculated amplitude, movement time, duration of EMG bursts, onset of EMG with respect to displacement, and average velocity (movement amplitude divided by movement time).

## RESULTS

Activity in the antagonist posterior deltoid was seen most commonly in association with a silent period in the agonist anterior deltoid following a burst period.(See fig.1) Antagonist firing was observed in 215 of 483 movements, corresponding to movement times ranging from 60-550 msec across all four amplitudes. During the larger amplitude movements ($25^{o}$,$50^{o}$,$75^{o}$), antagonist activity was consistently seen when movement times were smaller than 250 msec while activity in some movements was noted when durations were as high as 550 msec. At the shortest amplitude movement ($13^{o}$), antagonist activity was seen in some records with movement times of 320 msec but was consistently observed at duration shorter than 190 msec. Antagonist EMG appeared to be linked more closely to certain ranges of movement time rather than movement velocity because we noted many records without antagonist activity whose velocities were greater than many records with antagonist EMG.

The beginning of antagonist activity with respect to the onset of movement appeared to be linearly related to movement time (correlation coefficient=.92)(See fig.2). As movement time shortened, the beginning of antagonist EMG shifted closer to the beginning of movement. A similar shift was noted as a function of movement amplitude. The earliest antagonist onset times occurred at amplitudes of $13^{o}$. In a number of the quickest movements, antagonist activity preceded the start of movement by up to 40 msecs (See fig.2). Figure 2 also implicitly illustrates that the time between the onset of antagonist EMG and the peak of movement is fairly constant across amplitudes and movement times. The mean value of this parameter for 212 movements was 118 msec $\pm$ 36 msec. No clear relationship was found between antagonist burst duration and either movement time or velocity.

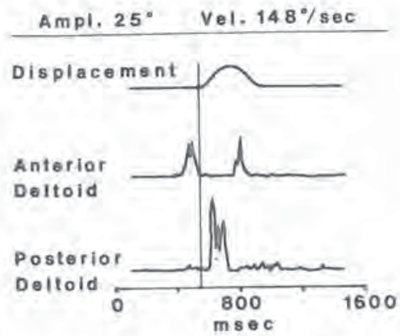


Fig.1 EMG activity pattern of a fast type movement with a movement time of 190 msec.

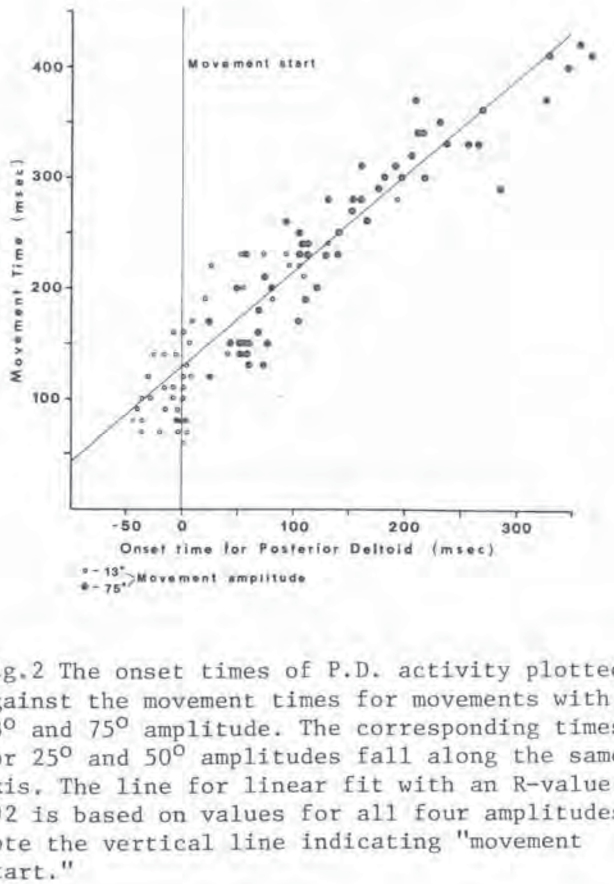


Fig.2 The onset times of P.D. activity plotted against the movement times for movements with 13° and 75° amplitude. The corresponding times for 25° and 50° amplitudes fall along the same axis. The line for linear fit with an R-value of .92 is based on values for all four amplitudes. Note the vertical line indicating "movement start."

## DISCUSSION

When a subject performs an amplitude specified movement at different self-generated speeds, the antagonist muscle appears to function as a brake, enabling the person to arrest forward progression at predetermined locations. The onset of antagonist activity with respect to the beginning of movement appears correlated with movement time (rather than velocity). The time interval between antagonist onset and the peak of movement appears relatively fixed. Our data suggest that antagonist action is orchestrated according to relatively precise timing features. Since antagonist activity was observed exclusively during rapid movements, some authors have discussed the possibility that antagonist EMG is induced by a peripheral velocity sensitive stretch reflex mechanism.[4] Our findings indicate that the onset of antagonist activity can and does take place prior to the onset of movement (when stretch of the antagonist has not as yet taken place) under certain conditions of amplitude and movement time. We, therefore, think that antagonist timing patterns in an amplitude specified movement probably reflects central programming.

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## ACKNOWLEDGEMENT

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# A KINESIOLOGICAL STUDY OF SHOULDER FLEXION

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## INTRODUCTION

Musculotendinous trauma of the shoulder joint is a common injury encountered by the health professional. Treatment of these injuries has traditionally involved wall pulley exercises to maintain and increase range of motion and strength during rehabilitation. This study was undertaken in an effort to provide data to improve patient care by analyzing the activity of the major muscles involved in shoulder flexion against varying levels of resistance.

## METHODS

The sample population consisted of 23 normal male subjects with a mean age of 23.2 years. Two parallel metal arcs attached to the right side of a special chair formed a channel for the hand insuring sagittal motion only. A cable passed through a system of pulleys was attached to the resistance at one end and a wrist cuff at the other. The wrist cuff was placed at the head of the ulna and a padded plexiglass splint restricted elbow flexion. Three straps comfortably secured the subject to the chair.

Twenty-five and fifty percent of the maximum weight each subject could lift through the full range of motion was calculated. Two trials, each at 100, 50, 25 and 0 resistance levels were performed with ample rest between trials. The subjects were instructed that the entire movement of $180^o$ should be performed in 3 seconds. A large face timer provided visual feedback. Trials which deviated more than $\pm$ .2 seconds were repeated.

Bechman miniature (bipolar surface) electrodes were placed over the bellies of the anterior (AD), and middle (MD) portion of the deltoid, clavicular head of the pectoralis major (CHPM), and the biceps brachii (BB). Measurements from bony landmarks insured accurate electrode placement and in addition the skin was prepared to maintain resistance below 3K ohms. The electrical activity was amplified and recorded by a Beckman RM Dynograph with four 9852A Integrator couplers. The amplifiers were set at 1mV/CM and the paper speed at 50mm/sec. The analog record was reduced to digital form by a Summagraphics X/Y coordinate digitizer.

A 16 mm Bolex camera operated at 16 frames/sec. was used to gather the kinematic data. Elapsed time and synchronizations of the EMG and film record was achieved with an electronic timer. Identifiable marks on the center of the shoulder joint and head of the ulna determined joint position for data reduction by a Hewlett-Packard digitizer and calculator.

The data were reduced to four variables of interest for each muscle at each resistance level; angle of the shoulder joint at beginning of activity (ABA), and peak activity (APA); area to peak activity (ArPA); and total area (TA). Means were calculated and submitted to an ANOVA corrected for repeated measures. Post Hoc analysis of significant means was accomplished using the Student-Newman Keuls test. The .05 level was set for statistical significance.

## RESULTS & DISCUSSION

Figure 1 illustrates the respective means and significant differences. The variables of ArPA and TA increased as resistance increased for all muscles supporting earlier works (1,2,3) The data supported previous authors in that once the AD began its activity it remained active throughout the movement (1,3)The ABA for the AD was between $5^o$ and $9^o$ at resistances of 100, 50, 25%. These were significantly less than the ABA of $24^o$ with only gravity providing resistance.

The BB was the only muscle that exhibited a significant decrease in ABA and APA between 0 & 25, 50, 100% resistance. BB also reached peak activity earlier in the movement than the other resistances. Others have reported TM activity in resisted extension.[1] The activity of TM was also closely paralleled to TB activity at the two higher resistances. It appears that with the elbow fixed in extension, the TB can contribute quite early to shoulder extension. The graph also indicates that neither the PD nor the LD is active before 25 degrees of extension. This study, therefore, only partially supports the work of others concerning the range of activity for these muscles.[2]

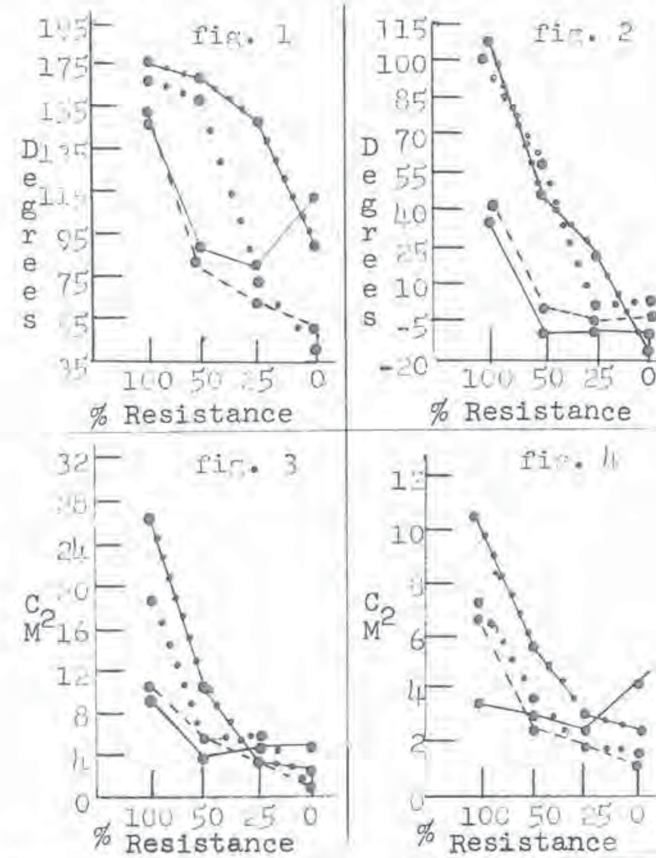
Figure 2 reveals that peak activity of each muscle, except for the LD, occurred at a significantly greater angle with maximum resistance than at lower levels. Peak activity of the TM and TB occurred at significantly greater angles at 50 and 25% resistance than at 0% resistance. These data further support the initiator function of the TM as well as the contribution of the TB to resisted shoulder extension.

As expected, the total area of activity for each muscle at 100% resistance was significantly greater than at lower resistances as displayed in figure 3.

Figure 4 shows that the LD, TM and TB exhibit significantly greater activity to peak at maximum resistance than they do at lesser values. Also, the ArPA for these muscles at 50% resistance was significantly greater than at 0% resistance. In contrast, the PD exhibits no significant differences in its ArPA regardless of resistance level. These data suggest that before reaching peak activity, the PD contributes evenly to shoulder extension, making it an important and consistent shoulder extensor.

## CONCLUSIONS

This study indicates that muscles are recruited earlier, exhibit greater total activity and, with the exception of the PD, display greater ArPA with maximum resistance. In addition, the TM and TB appear to initiate shoulder extension with high resistances. Finally, the PD is a consistent shoulder extensor regardless of resistance level.




Legend: (PD —) (LD...) (TM-+-+) (TB---)

Fig. 1 ABA
LD,TB 100/50,25,0
50/0
TB 50/25
PD,TB 25/0
PD 100/50,25
TM 0/100,50,25

Fig. 2 APA
PD,TM,TB
100/50,25,0
LD 100/25,0
TM 0/50,25
TB 50/25,0

Fig. 3 TA
PD,LD,TM,TB
100/50,25,0
LD 0/50,25
TM 50/25,0

Fig. 4 ArPA
LD,TM,TB
100/50,25,0
50/0

(slash mark denotes significant difference)

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# A KINESIOLOGICAL STUDY OF SHOULDER EXTENSION

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## INTRODUCTION

To improve motion after shoulder injuries, effective exercise programs may be hampered by musculotendinous pain. In attempting to avoid painful muscles during shoulder exercises it would be advantageous to know when specific muscles were active at various resistances. The purpose of this study was to investigate the function of four muscles during shoulder extension and hyperextension at various levels of resistance.

## METHODS

Twenty-three normal, male college students, with a mean age of 23 years, served as subjects. Each sat in a specially designed chair with his trunk comfortably stabilized by three straps and his right elbow splinted in full extension. From a wrist cuff attached at the level of the ulnar head, a cable ran first to a pulley fixed approximately 60cm above the right shoulder and then to a weight carriage which provided resistance. The upper right extremity was positioned between two parallel, semicircular bars fixed at both ends; these bars restricted shoulder motions to the sagittal plane. Subjects performed two trials of shoulder extension and hyperextension against 100, 50, 25, and 0 percent of their predetermined maximum resistance. To avoid fatigue, ample rest was allowed between trials. A large-faced timer provided feedback so that the full arc of shoulder motion was completed in three seconds. After proper skin preparation, pairs of Beckman miniature electrodes were attached over the bellies of the posterior deltoid (PD), latissimus dorsi (LD), teres major (TM), and the long head of the triceps brachii (TB). Electrode placement was determined by muscle palpation and reference lines drawn from bony landmarks. Surface markers on the center of the shoulder joint and the head of the ulna provided necessary reference points for kinematic data reduction. A Beckman type RM dynograph produced a printout of each muscle's activity after amplification and integration of myoelectric signals. Amplification was 1mv/div., and chart speed was 50mm/sec. Analog data were converted to a digital form by a summargraphic coordinate digitizer and then analyzed by computer. Films of each trial were taken by a 16mm Bolex camera operated at a film speed of 16 frames/sec. Synchronization of the film, muscle potentials and timing of each motion was accomplished by an electronic timer. Single film frames were analyzed with a Hewlett Packard digitizer and calculator to determine joint angle measurements. For each muscle, four variables were analyzed at each resistance: (1) shoulder angle where activity began (ABA), (2) shoulder angle at peak activity (APA), (3) total area of activity (TA), and (4) area to peak activity (ArPA). Means were calculated and submitted to an analysis of variance corrected for repeated measures. Post Hoc analysis of significant means was determined by the Student Neuman-Kuels test. The .05 level was set for statistical significance.

## RESULTS AND DISCUSSION

Figure 1 shows that TM, LD and TB began activity at 100 and 50% resistance at a significantly greater angle than they did at 0% resistance. The PD at maximum resistance began its activity at a significantly greater angle than it did at 50 and 25% resistance. Figure 1 also shows that the TM became active sooner than any muscle at 100, 50 and 25% resistance. This finding suggests that the TM could be an initiator of shoulder extension at these muscles at all resistances which may indicate its role as a shoulder joint stabilizer.

The PM yielded results that were expected except its ABA increased significantly between 25 and 0% resistance. The late beginning of the activity with zero resistance seems to differ from the classical shoulder flexion function usually assigned the AD and this muscle.

The MD was the only muscle that exhibited significant differences between all possible combinations of resistance for ABA. It was expected that the beginning activity of muscles would vary according to resistance. However, the data seems to indicate the MD was more sensitive to these changes than the other flexors. The APA for the MD was just the opposite being significantly different only between maximum and 0 resistance.

In summary the TA and ArPA of the muscles increased with resistance maintaining the same relative position. The ABA and APA increased as resistances decreased except the BB which decreased between 50 and 0 levels for both variables. The angle of the shoulder joint at which the MD begins its activity is very sensitive to resistance changes, whereas the angle at which it reaches peak activity is insensitive to changes.

## CONCLUSION

Since the angle of the shoulder joint at which muscles begin and reach peak activity vary the activitation of painful muscles can be avoided by careful attention to resistance and the extent of shoulder flexion.

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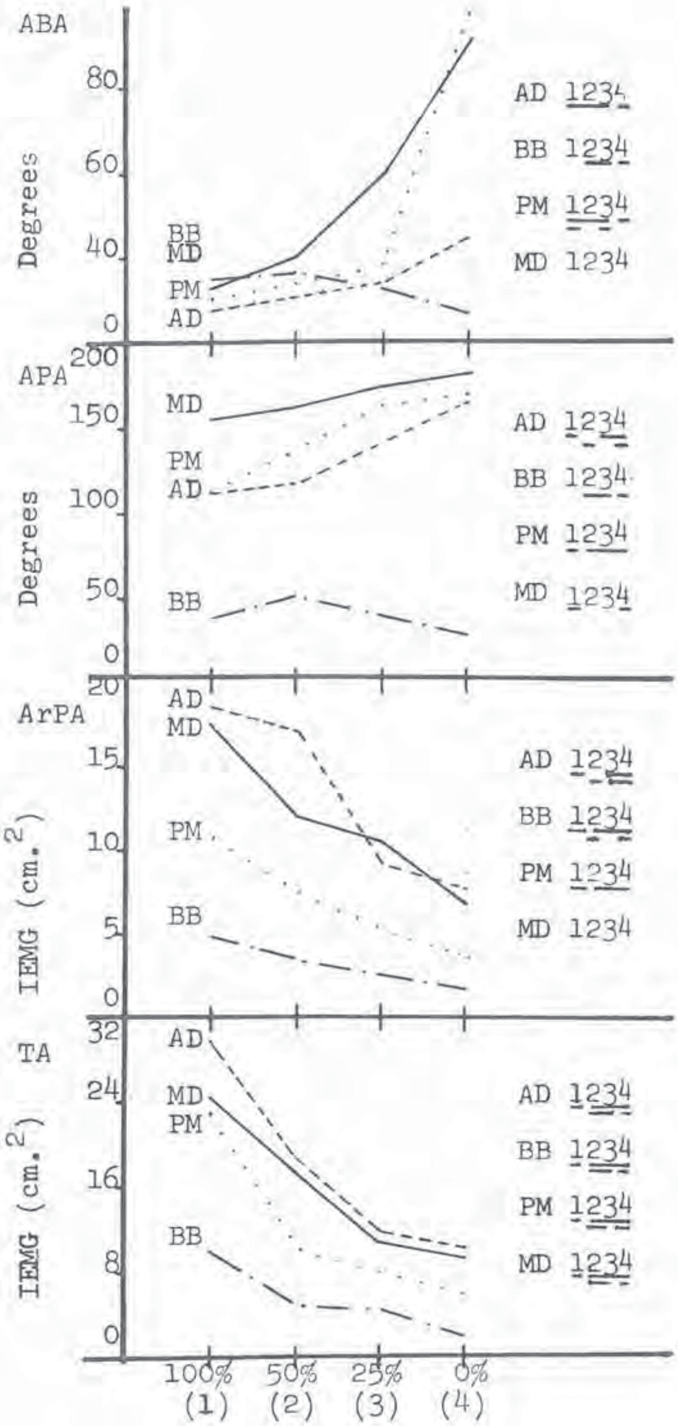


Figure 1. The muscle relationships across resistances including significant changes indicated by dash (initial) and underline (from).
* = all possible pairs significant.

ELECTROMYOGRAPHIC STUDY OF SHOULDER JOINT

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## INTRODUCTION

The shoulder complex is composed of four articulations,-sternoclavicular,acromioclavicular,scapulothoracic, and glenohumeral joints. These joints contribute to the elegant motion of the shoulder complex. But the role of muscles which take part in the shoulder movements are not yet settled completely.

The purpose of this study is to make clear the mechanism of shoulder complex; ie,(1) Analysis of movements of the scapulothoracic and glenohumeral joints during abduction in the scapular plane. (2) Electromyographic study of the shoulder girdle muscles during flexion and abduction in the scapular plane.

## EXPERIMENT

Fifteen normal men were used for this study of the movements of scapulothoracic and glenohumeral joints during abduction in the scapular plane. Two electric goniometers were placed in order to measure the degree of elevation of the arm(angle A) and movement of the glenohumeral joint(angle B). The scapulothoracic movement was calculated by 'A-B'.

Furthermore the role of the muscles during flexion and abduction were examined on ten of above persons. Six pairs of fine wire electrode were inserted into the right trapezius ( ûpper fiber), deltoideus(middle fiber), supraspinatus, infraspinatus, and pectralis major muscles. Action potentials of the muscles and change of the angle A and the angle B during elevation and depression of the arm were simultaneously recorded using a data recorder.

## DATA ANALYSIS

As the angle A and angle B measured by the electrogoniometers correlated with those measured with X-ray pictures (correlation coefficient; $r=0.994$). The formers were considered to show the real values. Although the patterns of integrated action potentials(I.A.P) were somewhat different from an examinee to examinee, those from the same muscle of the same person were almost always simillar in several trials. The I.A.P obtained from ten examinees were averaged at each ten degrees of elevation or depression.

## RESULTS

During abduction in the scapular plane, movement of the glenohumeral joint was accompanied by that of the scapulothoracic joint. During $30^\circ$-$130^\circ$ of elevation, every 15 degrees of motion was composed of 10 degrees at the glenohumeral joint and 5 degrees at the scapulothoracic joint. During elevation the glenohumeral angle began to increase more intensively than the scapulothoracic angle, also during depression the former began to decrease more strongly than the latter.

The mean I.A.P of ten subjects showed that all the muscles examined were active during flexion and abduction. During flexion, the I.A.P of trapezius and rhomboideus muscles showed the increasing activity, however that of the supraspinatus muscle showed rapidly increasing activity during $0^\circ$-$30^\circ$ of elevation and thereafter gradual increasing. In the deltoideus muscle the I.A.P increased maximally during $60^\circ$-$150^\circ$ of elevation of the arm. During abduction in the scapular plane, the increasing curves of the I.A.P of the trapezius,supraspinatus,rhomboideus, and deltoideus muscles were parallel to the degree of the arm. On the other hand, I.A.P of the infraspinatus and pectralis major muscles showed merely weak activity during abduction.

## DISCUSSION AND CONCLUSION

Using electrical goniometers the glenohumeral and scapulothoracic angle were measured exactly in the scapular plane. The curves obtained from both the continuously changing angle during elevation were different from those during depression. It may be due to flagging of the joint capsule. During 30 -130 of abduction of the arm the range of motion of the glenohumeral joint and the scapulothoracic joint were in the ratio 2:1. As the fine wire electrodes were used the examinees could move the shoulder complex with the electric goniometers smoothly without pain. To analyse the muscle functions, use of the integrated action potentials were more convenient than of the action potentiais themselves. It was interesting to find that all the muscles examined showed action potentials during flexion and abduction. Until now generally, the pectoralis major has been considered to be silent during abduction motion, however in this study, action potentials were observed from this muscle. It would be due to the fact that by integrating action potentials finest electric excitment of muscles were able to be picked up. All the above muscles examined participated in flexion and abduction of the shoulder. The trapezius, supraspinatus, rhomboideus, and deltoideus muscles showed the increasing activity with elevation, but the infraspinatus and pectralis major muscles showed only a little activity while elevation. Muscle activity during depression was less than that during elevation, but they both resembled each other in discharge pattern.


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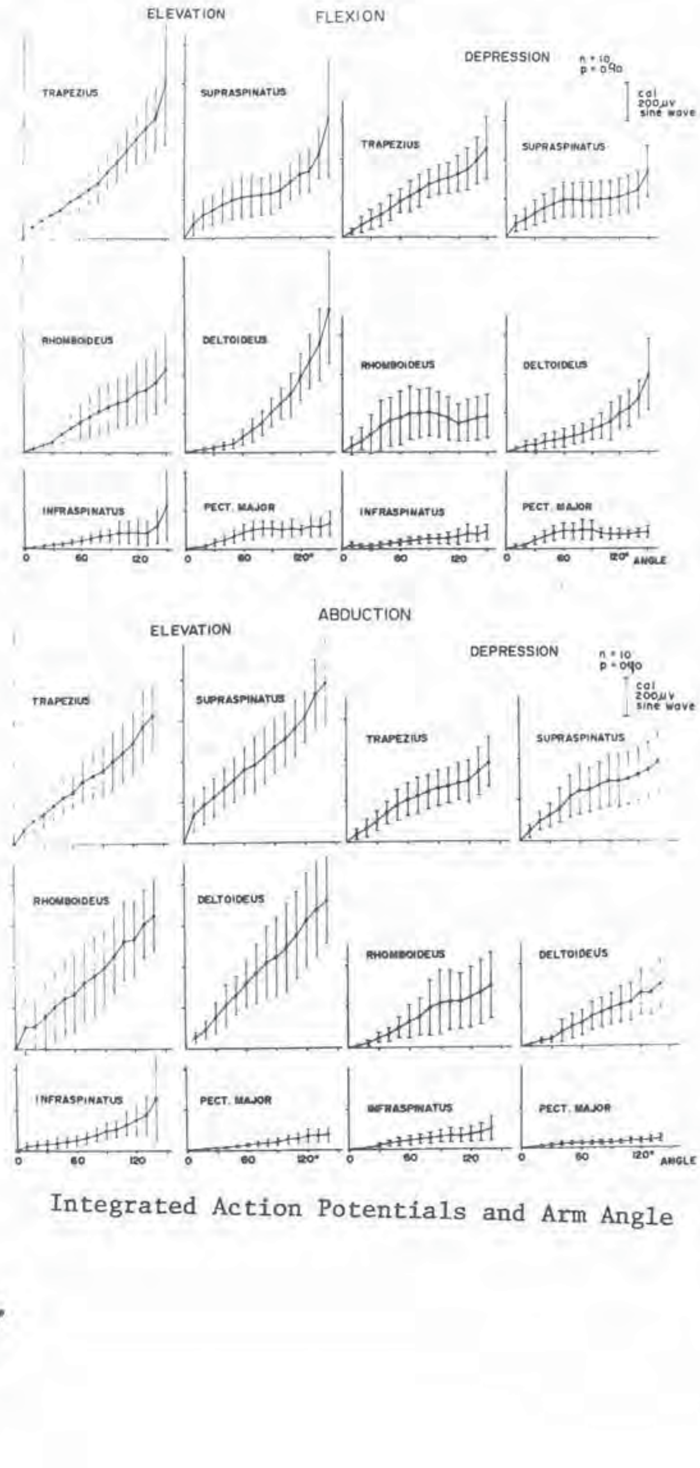
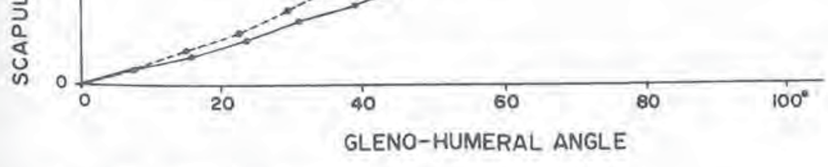



Integrated Action Potentials and Arm Angle




Correlation of glenohumeral angle and scapulothoracic angle (•—• ;elevation •---• ;depression)

NOTES

# NEUROPHYSIOLOGY

## Session 2A

# REINNERVATION OF THE TIBIALIS ANTERIOR TIBIAL MUSCLE OF THE RABBIT AFTER GRAFTING OF THE SCIATIC NERVE.

P.Pinelli,A.Moglia,M.Zanlungo,G.Sandrini,I.Pagani and A.Arrigo(University of Pavia,Italy)

In earlier research we cut the sciatic nerves of rabbits and immediately sutured them with neurorraphy or by a grafting technique, with the aim of investigating the histologic features of regeneration and related functional parameters of reinnervation of the tibialis anterior muscle,measured by means of EMG.
In the present study the main interest has been focused on the course of reinnervation,to obtain a model for the study of different factors to be taken into account as inducers of positive or negative effect on the date and modality of reinnervation after removal of a stump of the sciatic nerve and its substitution with a Cialit homo-graft.

## Material and Methods

Ten rabbits of the New Zealand variety of the same age (90-100 days) and height (2,4-2,7 Kgs) have been used.
From all the animals under a general anaesthesia (mixture of $N_2O$ and $O_2$ and ethilic ether through inhalation) 10 mm of the sciatic nerve were removed from the 3rd mean of the thigh and substituted with a corresponding fragment of a sciatic nerve (taken previously from another rabbit and preserved in Cialit liquid (sodium salt of 2 ethil mercury-mercapto acid 5 carbonic benzoxasol) for at least 30 days). The suture was a perineural neurorraphie carried out under optical magnification, using a nylon string of 8-0, 10-0 calibre suturing material depending on the size of the nervous section. The presence of fibrillation activity on the 1st,2nd,14th,30th,60th, and on the 90th day after the operation was checked,by an Adrian needle electrode inserted in the tibialis anterior muscle.
The maximum Motor Conduction Velocity of the sciatic nerve was studied before and 90 days (in 5 animals) and 180 days (in the remaining 5 animals) after the operation,always under general anaesthesia. A sleding stimulation electrode placed in two points of the sciatic nerve 4 cm from each other , proximally with respect to the biforcation , was used. Supramaximal stimuli square waves 0,1 sec.duration were emploied. The detecting wire electrode was placed in the tibialis anterior muscle.
In all the animals the sciatic nerve and the tibialis anterior muscle were removed for the anatomo-pathological staining after 90th or 180th day.

## Results

The fibrillation activity was homogenous in all the animals : it was scarse on the 2nd day (+--) high on the 7th (+++) high and constant on the 14th and 30th days (++-),later decreasing till the 60th day (++-).In 2 out of 5 animals it was still present,even though scarse, on the 90th day.The conduction velocity (average of 5 animals killed on the 90th day)was of $51,9 \pm 3,8$ m/sec with 70% recovery.
The histological study of the nerve with a traditional staining methods (hematoxyline-eosin,Van Giesen silver and simple luxol) has shown, at the level of the graft, the presence of myelinated fibres reduced in size and sinuously developped, on the 90th day. They were present, even if dissociated at the level of the suture because of oedematous swelling of the interstitial tissue, which had a myxoide appearance. On the 180th day,the graft looked like a normal nerve with a double break at the two ends, where the connecting tissue, by now organised , was crossed by rare strips of nerve ,prevalently on the edges.
The developed fibres were always markedly sinous.
The muscle showed regressive alterations of the fibres with interstitial cells on the 90th day, while they appeared normal on the 180th day. The results of the histoenzymological and ultrastructural studies are still under way. From the clinical point of view all the animals have begun to show a recovery of the motory function between the 30th and 60th day after the operation.

## Discussion

The temporal course of the fibrillation is comparable to that reported in our previous studies on neurorraphy and on the homo-graft of the sciatic nerve in rabbits (Zanlungo et al. 1978; Moglia et al. 1978).

It underlines the inversely proportional ratio of such phenomenon with respect to the max. motor conduction velocity.
In fact with serial controls we had already seen the appearance of fibrillation on the 2nd day, on which the end of the nervous conduction was observed.
It was shown that the recovery of nerve motor conduction between the 30th and 60th day was associated with the decrease in fibrillation , which is in agreement with the results reported by Pinelli (1965).
The values of conduction velocity obtained on the 90th and 180th day in nerves with Cialit-homografts are more than satisfactory, even though slower then tose found after neurorraphie.This difference can be justified after the double interruption that the reinnervation must overcome.
From the histologic point of view, the investigation with the standard staining confirms the good outcome after Cialit homo-graft, strictly corresponding to that got from the preceding experiments with neurorraphie and auto-graft, where all the intermediate stages of the denervation and reinnervation were studied.This work will be completed by the histoenzymological and ultrastructural studies underway.
Finally one must note the absolute lack of phenomenon of rejection and of each cellular reaction to the transplant in all animals observed.
Once these experimental functional and morphological parameters have been confirmed and statistically evaluated it seems to us interesting to investigate the eventual effects of the electro-stimulation of the denervated tibialis anterior muscle by applying the same EMG and histologie methods. A study on 10 rabbits is underway, in which the tibialis anterior muscle is stimulated twice a day with square wave 60 msec duration , 5/sec until the 90th day in 5 animals und until 180th in the re= maining 5, after the Cialit homo-grafting of the sciatic nerve.

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# COMPRESSION EFFECTS ON THE COMPOUND ACTION POTENTIAL OF IN SITU SEVERED NERVES

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## INTRODUCTION

An implantable nerve-electrode interface capable of recording neuroelectric signals for prolonged periods of time is being developed (1). Electrode units implanted around the distal ends of severed sciatic and peroneal nerves in rabbits have successfully recorded neuroelectric activity associated with volitional intentions of the animal for periods up to 142 days. However, the neuroelectric signals recorded from the electrode units exhibited degradation during the beginning of the implant period.

In order to improve signal quality and prolong the period for successful recording of neuroelectric activity, an investigation was undertaken to determine underlying causes for the degradation of the signals with time. Several preliminary experiments were conducted with rabbits and subhuman primates. These indicated that a possible cause for the decrease in the amplitude of the neuroelectric response recorded from the implanted electrodes was due to the forces generated by the musculature surrounding the implant. Application of an external pressure cuff around the limb of the animal at the site of the implant produced similar decrease in signal amplitude.

The results of these experiments suggested the possibility that compressive forces acting on the implant, such as those resulting from contraction of the adjacent musculature during movement of an animal's limb were causing injury to the nerve. The present study was undertaken to verify this assumption.

## EXPERIMENT

New Zealand white rabbits weighing 3.5 - 4.0 Kg were chosen for the experiments. The peroneal nerve was exposed and freed of connective tissue and a 3-channel differential recording electrode unit was installed on the nerve (1). A hook-type stimulation electrode was placed on the nerve approximately 5 cm proximal to the recording electrode unit. The nerve was stimulated supermaximally with 0.05 ms square wave pulse at 30 pps. The resulting compound action potentials (CAP) from each channel were amplified and observed on an oscilloscope and recorded on a multichannel FM tape recorder. During the stimulation, an apparatus incrementally applied compressive forces to a 3 mm length of the exposed nerve approximately 1 cm proximal to the implanted recording electrode. The rate and duration of the applied compressive forces were controlled. In four experiments forces were rapidly applied to the nerve in increments of 10 grams every 5 seconds until no observable CAP was present. In two experiments a sustained 10 or 20 gram weight was applied and the decrease in CAP observed as a function of time. In two experiments with similar preparation no weight was applied for purposes of control. In one control experiment, the animal was sacrificed and the CAP observed as a function of time.

## DATA ANALYSIS

Data obtained from the different experiments were digitized at an effective sampling rate of 160 kHz. Experiments in which the compressive forces were rapidly applied were sampled at the beginning, middle, and end of each incremental force application. An average of 30 CAP's was taken for each sample. Experiments with sustained 10 or 20 gram forces were sampled every sixty seconds and an average of 150 CAP's taken for each sample.

While compressive forces were applied to the nerve, changes in the amplitude and waveshape of the CAP occurred. The waveshape of the CAP was analyzed in both the time and frequency domains. In the time domain, the peak-to-peak amplitude of each averaged CAP was calculated and plotted as a function of compressive force or time. The propagation times, relative to the stimulus pulse, were taken for each averaged sample. The signal was prefiltered at 8 kHz and a time window was used to eliminate the stimulus artifact from the transform.

## RESULTS OF DATA ANALYSIS

Direct compressive forces acting on the nerve produced similar changes in the waveshape of the CAP for both rapid and sustained constant-force applications. The initial positive phase of the signal decreased in amplitude as a function of compressive force. The other phases of the signal remained relatively unchanged throughout the course of the experiments. (See Figure 1.)

The onset of these changes depended on the rate of force application. Experiments in which the force was incrementally applied every 5 seconds exhibited a threshold value of 25 to 40 grams, above which further increases in force caused a rapid decrease in the amplitude of the CAP. The average slope of this decrease was approximately 25 µv/gram. The signal decreased to 50% initial value after application of 50-70 grams. Forces at and below the threshold level were selected for experiments in which a sustained constant force was applied to the nerve. The observed changes in waveshape during sustained threshold and subthreshold force application occurred more gradually over a time span of 2-80 minutes. The signal decreased in amplitude to 50% initial value in 60 minutes for the sustained application of the 10 gram weight. The sustained application of the 20 gram weight required 2 minutes for a 50% decrease in amplitude. The peak-to-peak amplitude of the CAP before force application ranged from 1 mv to 2.5 mv for the experiments.

There was no observable change in the amplitude and waveshape of the CAP in the two control experiments where no forces were applied.

For both rapid and sustained force application, the frequency spectrum of the CAP exhibited a shift in the energy content to the lower frequencies as compressive forces and time increased. (See Figure 2.) A corresponding increase in the propagation time measured from stimulus application to the initial positive peak of the CAP was noted.

## DISCUSSION

The decrease in the amplitude of the CAP, along with the selective decrease in the energy content of the frequency spectrum, accompanied by the increase in propagation time suggest damage to the axons responsible for the higher frequency components of the CAP. This indicates that the population of larger fibers having faster conduction velocities are more susceptible to the effects of direct compressive force. Control experiments in which the animal was sacrificed indicate that termination of blood supply and surgical preparation are not variables for periods of time up to 2.5 hours. Ischemia and nutriant blockage seem to play a minor role in the decrease of the CAP during the observation period. The compressive force appears to primarily affect the structural integrity of the fibers. The rate and extent of damage is dependent on the magnitude and duration of the compressive forces.

The results show that short term application of relatively small forces on the distal portion of severed nerve caused substantial decrease in CAP and loss of large diameter fiber population. Forces of this magnitude are commonly encountered during manipulation of the nerve in a surgical procedure, and during contractions of adjacent musculature. Both occurrences would explain the rapid post-surgical decrease in the neuroelectric signal observed in our previous experiments.

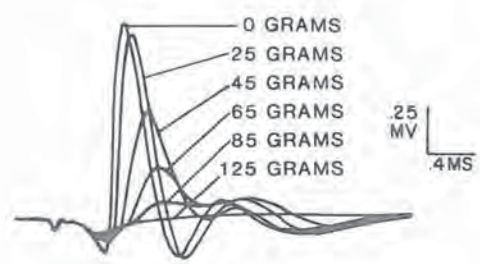


*Fig. 1. Compound Action Potential Recorded as Function of Compressive Force Applied to the Nerve at 5 Second Intervals*

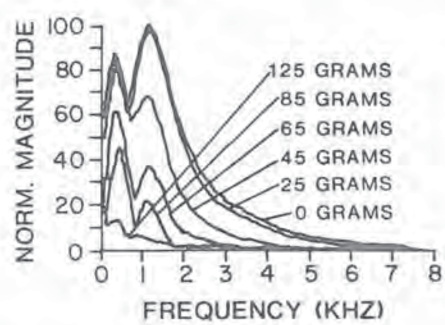


*Fig. 2. Frequency Spectrum of Compound Action Potential as a Function of Applied Compressive Force*

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## ACKNOWLEDGMENT


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CONSEQUENCES OF REMOVAL OF THE INHIBITORY EFFECTS OF RENSHAW CELLS AND Ia INHIBITORY INTERNEURONS ON THE LOCOMOTOR RHYTHM OF ALPHA MOTONEURONS.

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## INTRODUCTION

Interneuronal circuits coupled to flexor or extensor motoneurons (MNs) and linked by mutual inhibition have been suggested as possible locomotion generators (3) which are capable of producing rhythmic MN activity in the absence of any phasic afferent input (2). The identity of these circuits is unknown at present.

Renshaw cells (RCs), interneurons (INs) excited by recurrent collaterals from MNs, have been shown to inhibit agonist MNs, as well as agonist-coupled Ia inhibitory interneurons (IaINs) mediating reciprocal inhibition of antagonist MNs. The fact that both RCs (5) and IaINs (1) are rhythmically active during locomotion in the absence of cyclic afferent input has prompted the suggestion that antagonistic systems of $\alpha$ MNs, RCs and IaINs constitute the spinal locomotion generator (7). Such a model assumes that the rhythmic membrane depolarization that underlies MN firing during locomotion results from the periodic recurrent inhibition of IaINs and subsequent disinhibition of antagonistic MNs.

In the present study, intracellular recordings of $\alpha$ MNs were obtained in fictively locomoting cats. MN membrane potential changes that occurred during locomotion were compared with those observed after the inhibitory effects of IaINs and RCs were experimentally removed. Strychnine, which blocks the postsynaptic inhibitory effects of IaINs and RCs (4), and hyperpolarizing current injected into $\alpha$ MNs, which reverses the postsynaptic inhibition of both INs, were used to test the importance for locomotion of inhibitory synapses involving these cells.

## EXPERIMENT

A tracheotomy, cannulation of a carotid artery and a jugular vein and a laminectomy (L4-L7) were performed under halothane-nitrous oxide anaesthesia. The cat was suspended in a frame, with legs pendant, by clamps on the ilium and the L2 vertebral spine. The skin of the back was tied to form a paraffin-filled pool which was maintained at 38°C by a heating lamp. The left L5-S1 ventral roots (VRs) were cut and placed on platinum bipolar stimulating electrodes. A filament from either the L6 or L7 VR was mounted for recording.

A postmammillary decerebration was then performed and the anaesthetic terminated. After recovery from the anaesthetic, stimulation of the mesencephalic locomotor region (MLR) (30Hz) evoked locomotion (8) as evidenced by the occurrence of rhythmic MN activity in the VR filament and potentiometer recordings of hindlimb excursions. The animal was then paralyzed with a dose of gallamine triethiodide (2.0-3.0 mg/kg, i.v.) and artificially respired. After paralyzation was complete, the MLR was stimulated and locomotion was monitored by recording the activity in the VR filament (fictive locomotion).

Intracellular recordings from $\alpha$ MNs in the lumbar enlargement were obtained during fictive locomotion using glass micropipettes filled with 3M KCl or KAc. An electrometer capable of passing pulses of known current through the recording electrode was used to apply hyperpolarizing current (4sec. duration) sufficient to reverse IaIN-mediated IPSPs. In some animals, strychnine (0.1 mg/kg) was administered intravenously.

## RESULTS

Typical membrane potential oscillations in a quadriceps (Q) MN that were evoked in response to MLR stimulation in a paralyzed cat are shown in the upper trace in Fig. 1. Soon after MLR stimulation was initiated, the membrane potential began to exhibit rhythmic depolarizations (upward deflections) and hyperpolarizations with spiking eventually arising from the depolarized phases. Rhythmic discharges were also produced in a strychnine-treated cat (Fig. 1, lower trace) that were similar to those seen in the control despite the absence of the normal interburst hyperpolarization (IBH). Spontaneous EPSPs, not present in the control, are also apparent after strychnine.

Fig. 2 illustrates the effect of hyperpolarizing current on the locomotor activity of a QMN. The step deflection in a downward direction indicates the period of hyperpolarizing current application. A decrease in the amplitude of the IBH and a facilitation of MN firing were observed during periods of current injection, but the basic locomotor rhythm was not disturbed.

## DISCUSSION

Three lines of evidence now available indicate that disinhibition of $\alpha$ MNs does not adequately explain the depolarization of MNs which occurs during locomotion. It has previously been shown (6) that MN depolarization is not associated with a decrease in conductance as would be predicted by the disinhibition model (7). In addition, the present study demonstrates that strychnine blockage of IaIN and RC inhibitory effects abolishes the IBH but does not perturb the rhythmicity of MNs during locomotion. Reversal of postsynaptic inhibition also does not disturb the normal locomotor rhythm.

The evidence thus suggests that there must be some source of excitation to $\alpha$ MNs that contributes to membrane depolarization. This is supported by the unveiling of EPSPs by strychnine (Fig. 1). IaIN-mediated inhibition may be largely responsible for the IBH, but the purpose of this inhibition may be to reduce the possibility of spurrious firing rather than functioning in rhythm control. The phasing of this inhibition may be determined by periodic recurrent inhibition of IaINs since it is eliminated by strychnine

## CONCLUSION

The locomotor rhythm generated by the spinal cord is not determined by RC-IaIN interaction insofar as removal of the inhibitory effects of this system does not result in any apparent disorganization of locomotor activity. Periodic excitatory input to $\alpha$ MNs must be provided by the spinal locomotion generator since it does not appear that MN depolarization arises from disinhibition.

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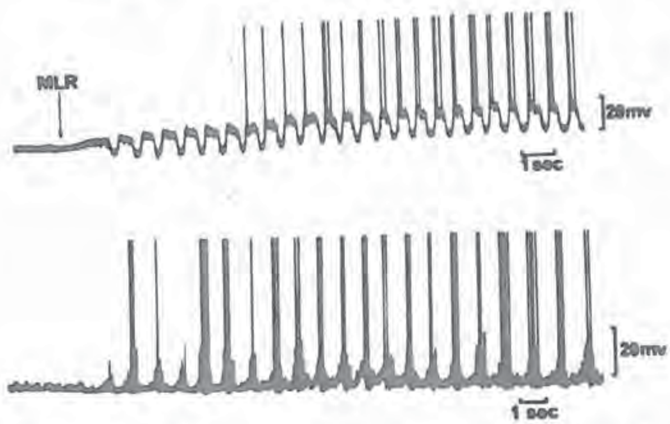


Fig. 1. Upper trace: intracellular recording of a QMN during fictive locomotion. Lower trace: same as above but in a strychnine treated cat. MLR stimulation started at beginning of trace.

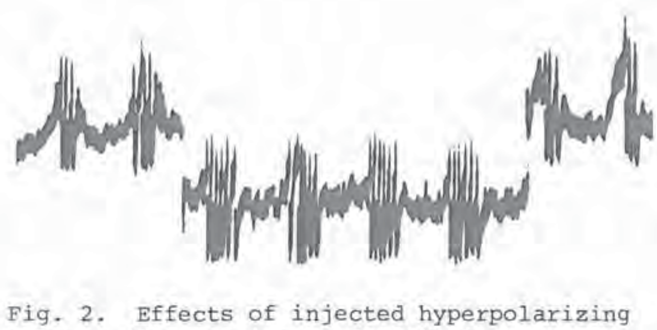

Fig. 2. Effects of injected hyperpolarizing pulses (15 mv., 4 sec. duration) on the rhythmic membrane potential (DC) of a QMN during fictive locomotion. Current injection is indicated by the downward step in potential.

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ONE TYPE OF SPECIFIC EFFECT OF RETICULAR ACTIVITY ON CORTICAL EVOKED POTENTIALS.

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## INTRODUCTION

Theoretical considerations based on experimental evidence suggest that various types of evoked potentials (cortical) are differentially modified by ascending activity of the reticular formation (pontine). In addition to spontaneous activity, some of the pontine reticular cells generate stimulus specific response to afferent sensory stimuli. The differences in response consist of different number of spikes per second and differences in PSP/spike ratios. Under the influence of pacemaker cells the spikes may be grouped into spike bursts. This ascending reticular activity seems to affect differently the various higher cerebral centers. The process of encoding and the cortical effects of the ponto-geniculo-occipital (PGO) waves as a representative ascending reticular process have been addressed in details in the here presented study. PGO waves are prevalently monophasic spike bursts (consisting of a few attenuating 8 per sec spikes recurring with a near 3 sec interburst intervals) that immediately precede and concur with the rapid eye movement (REM) sleep.

## METHODS

The nature and course of PGO waves were studied following the methods of Brooks and Gershon (1971), the activity of cells of the paramedian pontine reticular formation (PPRF) by the method of Segundo et al.(1967), and encoding of PGO waves by methods described by Torda (1979 a,b). Adult cats (either sec, 2.5-4 kg body weight) were implanted under halothane-$NO_2$ anesthesia with stimulating and recording electrodes following stereotaxic coordinates. The gross behavior of PGO waves was studied by bipolar electrodes inserted into the pontine tegmentum, the lateral geniculate (LG) and the marginal gyrus of the occipital cortex. Directly transmitted PGO waves were recorded either after surgical removal of the LG or transsection of geniculo-cortical pathways. Encoding of PGO waves was studied by intra- and extracellular recordings. Silver wire bipolar stimulating electrodes were inserted into various first and second order vestibular neurons vestibular nuclei, the nucleus interstitialis of Cajal(NiC), the nucleus Darkschewitsch(ND), the medial longitudinal fasciculus(MLF) and the pontine tegmentum in front of the PPRF. Rectangular pulses (0.1 msec pulse duration, 0.01-5 V ( 1-500 μA) intensity) were applied either in trains of 20 to 100 pulses per sec, or 1 to 3 pulses in 3 msec intervals. Glass recording microelectrodes (0.5-1 μm tip diameter, 10-20 megaohm impedance at 300 Hz) were filled with 3 M NaCl for extracellular and with 2 M K citrate for intracellular recording, and were inserted into the PPRF, the MLF and various vestibular nuclei. Serotonin(5-HT), norepinephrine(NE), acetylcholine(ACH), atropine sulfate and α-tubocurarine were administered microiontophoretically following the method of Kelly (1975).Parachlorophenylalanine(PCPA) was given for 3 days s.c.(150 mg/kg/day), reserpine (6 mg/kg) in one i.p. injection 80 min before the experiments. The generated bioelectric activity was amplified by conventional methods. The records were measured by hand or by the aid of a PDP-11 compiuter and the statistical significance of the observations was evaluated by conventional methods.

## RESULTS

Ten % of the PPRF cells responded with recurrent spiking ( 8 per sec) to sufficiently intensive spontaneous tonic activity of first order vestibular neurons and Type I neurons of the medial vestibular nucleus(MVN). High frequency phasic MVN pulses activated the inhibitory interneurons ( of the Nucleus interstitialis of Cajal(NiC) and the Nucleus Darkschewitsch(ND) to recurrently disrupt the excitatory pulses generated by the MVN for about 3 sec at a time. Together with other vestibular activity, these cyclic pulse bursts were transmitted through the medial longitudinal fasciculus(MLF) to the pontine reticular formation. The specific PPRF neurons responded selectively to the cyclic MVN-bursts by releasing the pontine PGO waves. The cyclicity of the PGO waves was secured by the cooperation of pacemaker cells located at the midline in front of the PPRF.

The phasic MVN activity required for encoding the interburst intervals is cholinoceptive and thus may be responsible for the appearance of PGO waves only immediately before and concurring with REM sleep. Lack of PGO waves during wakefulness seems to result from the inhibitory effects of norepinephrine, lack during slow wave sleep from the inhibitory effects of serotonin.

The pontine, geniculate and occipital PGO waves are near-synchronous, with the wavelength increasing from pons to cortex. The three waves significantly differ in amplitude & wave shape. Mathematical evaluations suggest that the occipital PGO waves represent a partial vectorial sum of the geniculate PGO and the PGO waves directly evoked by reticular activity transmitted without intervention of the lateral geniculate.

## DISCUSSION AND CONCLUSIONS

A method of encoding PGO waves has been identified. The code consists of two parts: (1) generation of 8 per sec spiking is carried by special activity of either first or second order vestibular neurons.(2)The near 3 sec interburst intervals are encoded by the periodic response of inhibitory interneurons to excitation that originate in the high frequency cholinoceptive phasic bursts of the medial vestibular neurons. The PGO-code is carried, together with numerous excitatory impulses by the medial longitudinal fasciculus(MLF) to the pontine tegmentum. A special PPRF cell group is able to selectively respond to the code by releasing the pontine PGO waves. The cyclicity of the PGO waves is secured by special pacemaker cells located near the midline in front of the PPRF. The ACH-dependence of the phasic MVN activity seems to be responsible for the presence of PGO waves only immediately preceding and concurring with REM sleep. Serotonin is responsible for lack of PGO waves during slow wave sleep, NE for their absence during wakefulness. The differential processing of PGO by the LG and the occipital cortex has a pertinent function in information processing (Torda, 1979 c) and partakes in the overall regulatory function of the reticular formation (Torda, 1979 d).

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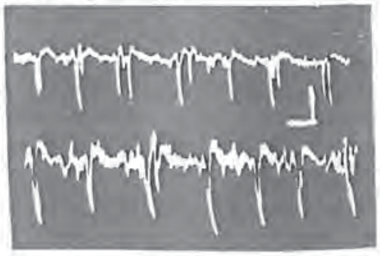
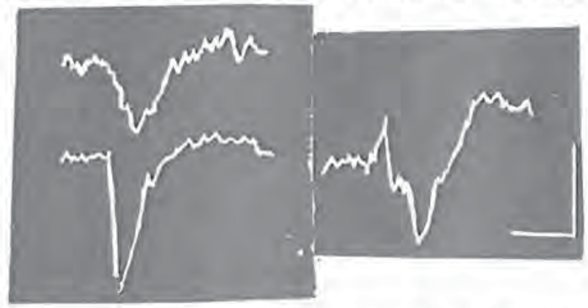
1. PONTO-GENICULO-OCCIPITAL WAVES
ROW I: pontine PGO
ROW II: occipital PGO
Horizontal: 3 sec
Vertical: 450 μV



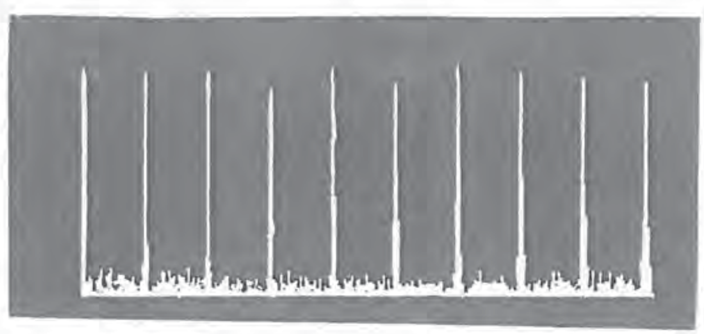
2. PGO WAVES (RIGHT: occipital PGO, intact cat
LEFT I: directly transmitted occipital PGO
LEFT II: LGOPGO)
Horizontal: 110 msec; Vertical: 450 μV



3. RECORD FROM PPRF CELL DURING SLOW WAVE SLEEP
ROW I: unstimulated cat
ROW II: stimulated first order vestibular neuron
Horizontal: 60 msec
Vertical: 50 mV

4. COMPARISON OF RECORDS TAKEN FROM THE MLF AND PPRF CELLS
ROW I: medial long. fascic.
ROW II: pontine PPRF cell
Horizontal: 1 sec
Vertical: I: 20 mV
II: 40 mV

5. AUTOCORRELATION HISTOGRAM FROM A CONSTANT INTERVAL PACEMAKER UNIT



# CONVERGENT VISCERAL (AORTIC OCCLUSION) AND SOMATIC INFLUENCES ON THORACIC SPINAL NEURONS

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## INTRODUCTION

Several investigations have examined the role of peripheral nerves in transmitting either visceral or somatic afferent information. The thoracic viscera is innervated by sympathetic afferent fibers with receptors which are sensitive to a variety of cardiovascular and pulmonary parameters. Receptors have been localized in the aorta, atria, ventricles, pulmonary artery and vein, lung, and coronary artery. Thoracic spinal nerves composed of general somatic afferent fibers supply sensory innervation to the upper forelimb and thorax.

The object of this study was to determine if neurons in the gray matter of the thoracic spinal cord respond to convergent inputs from both visceral and somatic structures. In particular, the effects of aortic occlusion were examined in this study.

## METHODS

Experiments were performed on 22 cats which were sedated with ketamine (10 mg/kg) and anesthetized with chloralose (60 mg/kg). The cats were artificially ventilated to maintain an end $CO_2$ level of 3-4% by connecting a positive pressure respirator to an endotracheal tube. Blood pressure was monitored after cannulation of a femoral artery. ECG was monitored with standard limb leads. Fluid was continually replaced by infusion of lactated Ringer's solution into a cannulated femoral vein.

A left hemithoracotomy in the fourth intercostal space exposed both the aorta and left sympathetic chain. The descending aorta immediately below the arch was cleared and a ligature was looped around the aorta. Aortic occlusions were subsequently accomplished by sliding a polyethylene tube, which was threaded into a ligature, against the aorta. The left sympathetic chain was cleared rostral and caudal to the $T_2$ white ramus communicans. Two electrodes were placed around the sympathetic chain for electrical stimulation of sympathetic afferent fibers.

A laminectomy exposed the $T_2$ to $T_4$ segments of the spinal cord. The spinal cord was stabilized by clamping the $T_1$ and $T_5$ vertebrae to a stereotaxic apparatus. A glass microelectrode filled with 4M NaCl was driven into the left gray matter in order to record the extracellular potentials of single cells.

Each spinal neuron in this study met three criteria. First, the unit responded to sympathetic afferents during electrical stimulation of the sympathetic chain. Second, the unit responded to "natural" activation of the sympathetic afferents during aortic occlusion. Third, the unit responded to somatic afferents by pinching the left forelimb and thorax.

## RESULTS

A total of 53 neurons which responded to electrical stimulation of the sympathetics were further tested for a response to aortic occlusion (AO). Only 16 units (30%) responded to AO. Of the 16 AO-responsive units, 15 also responded to pinching of the somatic field.

The effect of AO on responsive units was either excitatory (11 units) or inhibitory (5 units). The majority of these units (14) responded during AO when systemic arterial pressure was reduced. In 4 of these units, concomitant arrhythmias were observed on the ECG tracing. Only one unit responded immediately to AO (4 sec latency) without correlations to changes in blood pressure or ECG.

Most of the AO-responsive neurons also responded to manipulation of the left forelimb or thorax. The response of the spinal neuron depended on stimulus thresholds. A high threshold stimulus, ie. hard pinch, was generally necessary to affect the cell activity. When the somatic field was pinched, 14 spinal units were excited while the remaining 2 were inhibited. Two units also responded to lower threshold stimuli, ie. blowing the hair or touching the skin, in addition to responding to pinching. In these units responding to multiple somatic stimuli, the response of the spinal neuron was directly related to the stimulus threshold. The receptor was most commonly activated by pinching the skin, but several units also responded when the underlying muscle was pinched. The somatic field was moderate to large in size, generally several square centimeters in surface area.

## DISCUSSION

Somatic afferents had receptors located in the skin and muscle of the upper left forelimb and left thorax. The cell sometimes responded differentially to different thresholds of somatic stimuli suggesting distinguishable encoding of multiple somatic afferent information. In general, the somatic afferents are probably involved in nociception since most spinal neurons responded to somatic pinching.

Due to the selective criteria of this experiment, about a third of the sympathetically-activated neurons also responded to AO. By occluding the descending aorta, it is assumed that blood pressure increases within the aortic arch, carotid sinuses, heart chambers and pulmonary vasculature. Consequently, baroreceptors located in these sites are probably the most commonly activated receptors. The results of this study clearly demonstrate that the majority of spinal neurons responded during alterations in systemic arterial pressure. Other studies in this laboratory have shown that spinal neurons also respond to receptors located in other visceral sites. Perhaps some of these other receptors may be secondarily affected by AO, and there is evidence that some spinal neurons respond to multiple visceral afferents.

The spinal neuron responded to viscerosomatic afferents in a variety of ways. Most common was a similarity of the response characteristics (excitation or inhibition) for separate application of visceral or somatic stimuli. Oftentimes the response was indistinguishable between visceral and somatic stimuli, however, a few units did have distinct frequency differences. Other units distinguished between visceral and somatic afferents by excitation to one and inhibition to the other. Of interest is the large proportion of units which were inhibited by AO. This was unexpected since the protocol required cells which were excited by electrical stimulation of the sympathetic chain. This is a differential response to different sympathetic afferent fibers, ie. excitation by one fiber group during electrical stimulation and inhibition by other convergent fibers during AO. Alternatively, the thoracic spinal neuron may be indirectly affected by vagal afferents which first affect medullary cardiovascular neurons. The axons of these cardiovascular neurons may descend in the spinal cord to secondarily influence the response of thoracic cells.

Spinal neurons may be involved in several functions. Probably the majority of units were interneurons transmitting viscerosomatic information in a polysynaptic pathway. First, the neurons might be involved in spinal reflexes. This may be separated into somatic sensory and motor reflexes via spinal nerves or cardiovisceral reflexes via the sympathetics. Secondly, the neurons may carry information ascending to higher levels of the CNS. This may be subcategorized to those ascending to bulbar cardiovascular nuclei in the reticular formation or those ascending via the spinothalamic tract. This latter pathway may be involved in referred pain if the encoding of convergent visceral and somatic information is indistinguishable.

## ACKNOWLEDGMENT

This work was supported by NIH Grants HL05670 and HL22732.

NOTES

# MOTOR CONTROL

## Session 2B

SINGLE MOTOR UNIT CONTROL AND THE NUMBER OF MUSCLE SPINDLES.

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## INTRODUCTION

According to some authors (e.g. Birbaumer 1976/77) EMG-feedback has proven to be the most promising form of Biofeedback. The fine discrimination made possible by this technique is illustrated by the ability to exert volontary single motor unit control (SMUC).
Work of Van Ravensberg et al. (1978) showed that a positive relationship exists between the ability of a subject to control single motor units in a muscle and the relative number of muscle spindles in that muscle.
This finding is based on experiments in which SMUC was compared for the abductor and flexor pollicis brevis muscles. Even though the relative content of spindles of these muscles differ substantially (resp. 29.3 and 11.5 spindles / g of muscle (Schulze, 1955), both can be considered high with respect to other muscles. The present experiment was designed to compare two muscles on the extreme ends of the scale of relative spindle content.

## METHOD

Ss were 30 students who had never participated in EMG-experiments and who were naive w.r.t. the tested hypothesis.
The electromyograms of the m. abductor pollicis brevis (APB) and the combination of m. mylohyoideus and m. digastricus pars anterior (MD) were registered with the use of Beckman miniature surface electrodes (type 650950) and a DISA (15C01) amplifier. Conventional methods were used to insure good electrical contact with the skin. The electrodes were placed 15 mm apart in the middle of the belly of the APB and on the chin on the midline approximately 1 cm behind the posterior aspect of the mandibula.
The electromyogram was presented to the subject through an osciloscope (national VP 52G0A) (raw signal) and a set of headphones (filtered signal 0-500Hz) (Krohnhite filter 3750). All electromyograms were recorded on tape (Philips analog 7 type EL 1128-21B) for purposes of post-experimental checks and analysis.
The experiments were monitored with the aid of a storage osciloscope (Philips PM 3234) and a loudspeaker.
After having been exposed to a taped demonstration of different phases of SMUC and familiarized with the feed-back signals of their own muscles the subjects were asked to isolate a single motor unit for one minute. Ss were free to chose a position of hand arm and jaw but no contact of the thumb with other fingers or objects was allowed and no contact of the teeth occured. Half the subjects started the experiment with the APB and the other half with the MD as determined by chance. For each muscle the experiment lasted maximally 30 min. with a 30 min. break between the two sessions.

## RESULTS

SMUC performance was operationalized through time scores indicating the time needed to attain isolation of a motor unit. Significantly ($p<.01$, Sign test) more Ss were able to isolate a single motor unit in the APB (15 out of 30, in a mean time of 8 min.) than in the MD (4 out of 30, mean time 19 min.).

## DISCUSSION

Considering the location of our electrodes over the chin muscles it is likely that most activity registered originates from the m. mylohyoideus. It is generally accepted that this muscle contains no muscle spindles (Karlsson, 1976). However the possibility that the m. digastricus (anterior head) contributes to the EMG-signal cannot be completely excluded. For this muscle contradictory evidence exists on muscle spindle content: Kubota and Masegi (1977) found none in the anterior belly, while Voss (1956) found 6 spindles in this part of the muscle (approx. 2 spindles / g). No detectable response can be shown in these depressor muscles however, when the chin is tapped from underneath, while the jaw jerk (sudden downward motion of the jaw) illicits strong reflex activity in the masseter muscle (Matthews, 1976), which is richly endowed with muscle spindles (Freimann, 1954).
The relative spindle content of the APB is 29.3 spindles / g of muscle (Schulze, 1955).

In the light of this discussion one might conclude that our results once more confirm the hypothesis of the dependence of SMUC on relative number of muscle spindles.
However it was noticed that the amplitude of the registered motor unit potentials frequently was substantially lower in the MD compared to the APB, thus influencing the signal to noise ratio and thereby the quality of the feedback. Consequently this conclusion may only be drawn under the assumption that this lower signal to noise ratio of feedback would not have an effect on the final result. Even though the effect of this phenomenon can not be quantified as yet, the assumption seems to have a reasonable base. On the one hand one would expect some effect on time to learning while on the other hand it is clear from the litterature (e.g. Kato and Tanji, 1972) that is is frequently much easier to isolate small amplitude units in comparison to large ones. The size principle (Milner Brown et al. 1973) which has been shown to apply to cranial muscles as well (Yemm, 1976; Goldberg, 1976) leads to the same expectation, especially since for some cranial muscles a positive, though by no means perfect, relationship has been shown between the twitch tension of a motor unit and its amplitude on the surface EMG. (Yemm, 1976). If one accepts the above mentioned assumption the following can be concluded.

Single motor unit isolation is strongly influenced by the presence of muscle spindles in a muscle.

Direct involvement of muscle spindles of antagonist muscles in the process of isolation is unlikely in the light of the present finding, as most of these muscles have substantially higher spindle contents (5-14.7 spindles / g) (Voss, 1956, 1958; Freimann 1954) than MD. One would expect better SMUC-performance in such a case of the chin muscles.

Receptors of the joint capsule could play a role, but to explain the present results and those of Van Ravensberg et al. (1978) one would have to postulate unlikely distributions of receptors in the tissues of the joint parallelling the muscle spindle distribution of the muscles studied.

From the fact that it is possible for some subjects to attain SMUC in MD one can conclude that it is likely that direct involvement of muscle spindles is not a prerequisite to make isolation at all possible. Simular arguments can be presented w.r.t. tendon organs since it is generally accepted that such receptors, nor activity attributable to them, have been unequivocally demonstrated in masticatory muscles (Karlsson, 1978). Involvement (indirect) of muscle spindles from synergists or antagonists can not be excluded.

The conclusion that muscle spindles play an important role in SMUC is in agreement with the finding of Fukushima et al. (1976) that the control of motor units firing at a frequency of 10 Hz is abolished by xylocaine administration even though isolation is maintained, and their conclusion that functional γ-neurons are indispensable for the control of the motor unit at such a frequency.

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## INTRODUCTION

Regulation of man's upright posture is known to involve the complex interaction of visual, proprioceptive and vestibular systems operating in both feedforward and feedback modes of control. While much attention has been paid to experimental evidence of the role of sensory feedback consequences of unexpected perturbations, little empirical evidence is available on the nature of the feedforward (anticipatory) operations. In the present study, anticipatory stabilizing activity of muscles of the trunk and lower limbs has been investigated by recording the surface electromyograms prior to and during a series of rapid arm movements.

## THEORETICAL BASIS

As a first approximation to modeling the anticipatory stabilizing activity of the trunk and leg muscles, we have treated the musculoskeletal system as a 5-segment kinematic linkage and postulated that any anticipatory activity that occurs will be localized in those muscles where dominant net torques are initially required a) to counteract the reactive (inertial and gravitational) torques associated with the arm movement and b) to maintain the alignment of "non-moving" segments with the total body center of gravity located over the base of support. Order in anticipatory postural activity is thus considered to be determined by, and predictable from, the dynamics of the forthcoming movement. The dynamics may be quantified through the application of basic Newtonian mechanics:

$$\Sigma F_x = m \cdot a_x$$

$$\Sigma F_y = m \cdot a_y$$

$$\Sigma M = I \cdot \alpha$$

Our experiments were designed to enable the simultaneous recording of EMG activity in a number of postural muscle groups prior to and during movement and to conduct a dynamic analysis of the dominant muscle torque requirements of the task.

## PROCEDURES

Seven healthy female subjects took part in this study. Each subject was required to rapidly move her extended arms in response to a visual stimulus for three series of ten trials. The first series consisted of simple bilateral flexion. Series two involved bilateral flexion with a 1 kg. weight in each hand, and series three involved bilateral extension. EMG activity (linear envelopes with 6 Hz cutoff) in the following muscles was monitored by Beckman surface electrodes: anterior and posterior deltoid, rectus abdominus, erector spinae, rectus femoris, biceps femoris. tibialis anterior,soleus. Selected trials were filmed at 50 fps using a Bolex cine camera. Kinematic data were combined with anthropometric information and force-plate data (recorded on magnetic tape), in a computerized five segment model, to yield joint torque patterns at the ankle, knee, hip and shoulder.

## RESULTS

Table 1 shows the mean EMG onset times for the flexion (no weight) and extension conditions for all subjects. Individual subject torque time histories at the shoulder, hip, knee and ankle joints of one subject are shown in Figures 1 and 2.

TABLE 1

Means and Standard Deviations of EMG Onset Times

| | Flexion (msec.) | Extension (msec.) |
|---|---|---|
| Ant. Deltoid | 290 ± 21 | |
| Post. Deltoid | | 274 ± 13 |
| Rec. Abdom. | 291 ± 47 | 237 ± 14 |
| Erect. Spin. | 248 ± 17 | 279 ± 10 |
| Biceps Fem. | 268 ± 27 | 283 ± 19 |
| Rect. Fem. | 301 ± 29 | 285 ± 21 |
| Arm Accel. | 310 ± 54 | 292 ± 69 |

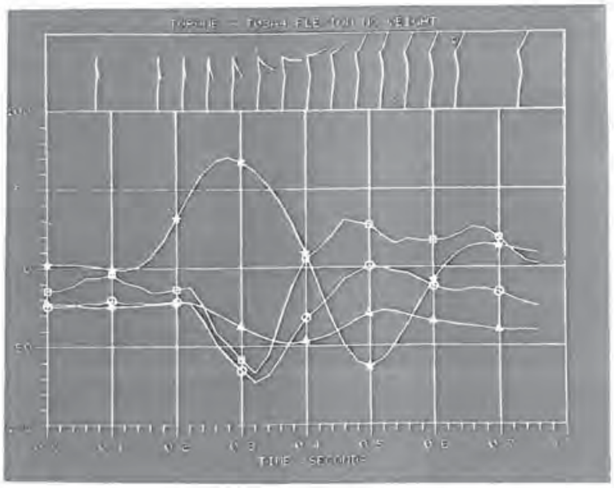

Figure 1. Torque Time Histories for Subject T03A4 during arm flexion. The y-axis is torque calibrated in Newton-meters. *-shoulder, ☐-hip, O-knee, Δ-ankle.

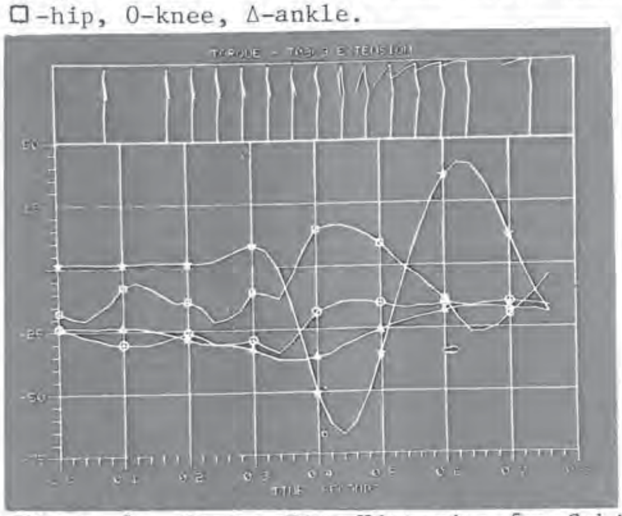

Figure 2. Torque Time Histories for Subject T03C3 during arm extension. Key as in Figure 1.

## DISCUSSION

In all six subjects postural EMG activity occurred prior to the arm acceleration (which was used to indicate the onset of the equilibrium disturbing movement). This evidence (see the onset times of erector spinae and biceps femoris in the flexion condition and rectus abdominus in the extension condition, Table 1) supports the concept of a preparatory stabilization of the trunk and lower limbs. Since postural activity in trunk and lower limbs precedes the upper limb motion, it would appear to be the result of centrally generated commands and not the result of sensory feedback. Similar evidence has been presented elsewhere and it has been speculated that this may correspond to an energy minimizing process (Belenkii, et al,1969).

A unique pattern of postural muscle activity has been identified in the lower limbs and trunk for each type of movement of the upper limbs. The sequence of activation of the trunk and limb muscles prior to arm movement corresponds to the order of the dominant muscle torques (see Figures 1 and 2) required for stabilization of the kinematic linkage. These patterns are consistent across trials.

Examination of the torque curves in Figures 1 and 2 helps to explain the early onset of specific muscles. An extensor torque at the hip and flexor torque at the knee during the acceleratory phase of the movement (as indicated by their negative values), in Figure 1, are related to early erector spinae and biceps femoris activity. Similarly, in Figure 2, a hip flexor torque (positive value) is related to early rectus abdominus activity.

Onset times related directly to the visual stimulus were difficult to discern for the distal muscles (soleus and tibialis anterior). Continuous activity was present in these muscles prior to and during the upper limb movement. It has been suggested that mechanisms other than preprogrammed responses prevail in distal muscles (Litvintsev, 1973).

One of the practical ramifications of these results is that there is a need for therapists to reconsider the therapeutic implications of equilibrium maintenance mechanisms. Evidence of anticipatory muscle activity may necessitate a re-assessment of the role of peripheral afferent feedback in neuromuscular facilitation techniques. Such feedback could conceivably be only minimally significant if the pattern of muscle activity is centrally preprogrammed. In addition, the fact that any voluntary movement consists of acceleratory and deceleratory phases, must be considered. Different postural muscle groups will be facilitated or inhibited during each phase and the therapist must be aware that such movements can increase the tone in both the agonist and antagonist. Utilization of the predictable patterns of anticipatory activity in postural muscles during voluntary movement may provide a useful supplement to conventional passive facilitation techniques that are feedback dependent.

## CONCLUSION

The postural activity patterns that occur in trunk and limb muscles, prior to rapid flexion or extension of the arm, are predictable from the dynamics of the forthcoming movement.

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# THE EFFECT OF EXPECTED AND UNEXPECTED MOVEMENT PERTURBATION ON EVENT-RELATED BRAIN POTENTIALS AND MUSCULAR RESPONSES

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## INTRODUCTION

The organization of load correction strategies is thought to involve spinal and supraspinal mechanisms, including the supposed long-loop reflexes. Whilst much research effort has been directed at the effects of prior instruction, little has been said about the effects of stimulus probability on brain and muscular correlates of load correction.

## THEORETICAL BASIS

Evidence derived from recordings of single cortical neurons in monkeys (Conrad et al., 1975) has clearly demonstrated "early" responses following transient load changes to self-paced movements which were tightly coupled to EMG changes. Ablation of the sensory cortex abolished the M2 response of the EMG (Tatton et al., 1975) suggesting a functional relationship between the sensorimotor cortex and responding muscle.

The present study utilized the averaged electroencephalogram (Event-related Potential or ERP), time-locked to perturbation onset, to investigate the possible cortical manifestation of both the long-loop reflex and volitional motor response under conditions of high and low stimulus expectancy.

## PROCEDURES

Eleven healthy male subjects, aged 18-34 years were used in this study. The criterion task consisted of a learned elbow extension movement through 90°. In the low expectancy condition 50 of a total of 250 trials were suddenly loaded "mid stream" by a breaking device attached to the task apparatus, and the subject responded by attempting to correct the load disturbance. In the high expectancy condition each of 50 trials was loaded.

The EEG (bandpass .05Hz-3KHz) was derived from an active electrode located on the contralateral parietal scalp referred to the opposite ear, and the EOG was monitored throughout. EMG data (bandpass 3Hz-5KHz) were recorded from the responding muscle (m. triceps brachii) and the signal was full-wave rectified and smoothed. Velocity of the lever arm was recorded together with a perturbation signal which provided the averaging trigger. All data were stored on magnetic tape and edited off-line for EOG artifact, following which the "legitimate" trials were averaged and digitized on a PDP 8/E lab computer for digital detrending, filtering and plotting.

## RESULTS

Table 1 shows the mean values for ERP, EMG and kinematic data.

## DISCUSSION

Typically the ERP response to load perturbation consisted of two components, an inconsistent early potential (N14) and a later positive-negative-positive complex (N140). The effect of changing the stimulus probability had no effect on N14, but all components in the N140 complex commenced and peaked earlier in the high probability condition, although the response amplitudes were not significantly different. The fractionated EMG response followed a typical 3-burst pattern with M1 peaking earlier, M2 with greater amplitude and M3 commencing earlier in the high probability condition. The data suggest that the N140 ERP component is related to the motor response and is influenced by stimulus probability. This, coupled with the significantly augmented M2 component of the EMG in the high probability condition seems to indicate a preprogramming of the response which takes into account the anticipated perturbation and subsequently, perhaps by feed-forward mechanisms, increases the gain on the spinal motoneuronal pool at the appropriate time.

## CONCLUSION

For condition in which there is a high expectancy of load change the ERP is less well defined and occurs earlier when compared to the ERP under conditions of low expectancy.

TABLE 1

Means and Standard Deviations of ERP, EMG and Velocity for High and Low Stimulus Probabilities

| | | | | |
|---|---|---|---|---|
| N14 | Lat. to Peak (msec) | 13.5± 6.3 | 14.3± 6.6 | 22.5 |
| | Peak Ampl. (μvolt) | 5.0± 4.6 | 6.0±10.0 | 34.5 |
| N140 | Onset Lat. (msec) | 53.5±25.6 | 77.7±34.0 | 12.5* |
| | Lat. to Peak (msec) | 91.5±24.7 | 137.5±24.8 | 0.0** |
| | Peak Ampl. (μvolt) | 9.5± 9.5 | 14.6±21.0 | 22.5 |
| P200 | Lat. to Peak (msec) | 176.9±22.0 | 217.4±57.2 | 15.0* |
| | Peak Ampl. (μvolt) | 17.4±12.9 | 26.4±27.9 | 23.0 |
| Reaction Time (msec) | | 145.4±22.1 | 166.9±24.9 | 3.0* |
| M1 | Lat. to Peak (msec) | 13.4± 5.3 | 17.1± 7.1 | 13.5* |
| | Peak Ampl. (m volt) | 15.2± 9.5 | 17.3±16.6 | 27.5 |
| M2 | Onset Lat. (msec) | 46.1± 9.7 | 46.4± 8.2 | 26.5 |
| | Lat. to Peak (msec) | 67.6±15.7 | 61.1±10.2 | 35.5 |
| | Peak Ampl. (m volt) | 65.7±55.0 | 26.4±16.1 | 8.0* |
| M3 | Onset Lat. (msec) | 109.5± 7.2 | 129.8±24.3 | 2.0** |
| | Lat. to Peak (msec) | 191.1±40.0 | 209.1±37.2 | 21.5 |
| | Peak Ampl. (μvolt) | 91.5±86.8 | 69.9±78.5 | 42.0 |

1 Lowest sign total in the Wilcoxen matched-pairs signed-ranks test
Significance level *.05 **.01

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VARIATION OF SECOND-ORDER PARAMETERS IN HUMAN HAND TARGET TRACKING

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## INTRODUCTION

The transfer function of a second order, low-pass, linear system can be represented by

$$G(S) = \frac{\omega_n^2}{S^2 + 2\zeta\omega_n S + \omega_n^2}$$

where A - low frequency gain
$\zeta$ - damping coefficient
$\omega_n$ - natural frequency

This model has been suggested as a reasonable model for characterizing the time course of twitch contractions by motoneurons of the human soleus muscle (Bawa and Stein, 1976) and the response of the human arm to external mechanical disturbance while the arm is maintaining a fixed posture (Crago et al., 1976). The possibility of extending this model to represent the human target tracking ability and how the parameters $\zeta$ and $\omega_n$ may change as a function of the motor task is the purpose of this investigation.

## METHODS

Three adult male subjects were studied. The subjects sat comfortably in a chair with their hands rested on a table. The right forearm (dominent side) was held in place by a splint. All fingers, except the index, were also restrained. A strain-gauge (Grass FT03) was attached to the lateral surface of the index finger (near the thumb). The forces controlling the lateral motion of this finger were measured isometrically.

By controlling the forces exerted on the strain-gauge by the index finger, the subject manipulated one of two beams on an oscilloscope (response, f(t) in Fig. 2). The target, either a sinusoidal (k sin $\omega$t) or a ramp (k t $u$(t), where $u$(t) is the unit step function) function, was displayed on the accompanying second scope trace. The magnitude of the target (k) was adjusted to solicit approximately a 100 gm force variation from the subject. The magnitude was kept constant for the entire duration of each experiment. The targets and responses were time averaged, 8 or 32 sweeps depending on the subject, with an electronic averager synchronized by a trigger pulse from the function generator.

## RESULTS

In a target tracking situation, the low-frequency gain (A) is 1. That is, at low frequencies, the magnitude of the response is equal to the magnitude of the target.

1. Sinusoidal targets

The lowest frequency used in these studies was 0.05 Hz. At this frequency, all subjects can easily follow the target faithfully. Starting at 0.1 Hz, the target frequency was increased logarithmically in a stepwise fashion (i.e. 0.1, 0.2 ..., 0.9, 1, 2, ...Hz). The maximum frequency was determined when the subject's average response amplitude is less than ½ of the target. The gain at each frequency was determined by

$$\text{Gain} = |G(j\omega)| = 20 \log_{10} \left| \frac{\text{p-p response}}{\text{p-p target}} \right|$$

(note $\omega = 2\pi f$)

The natural frequency (by sinusoidal method), $\omega_{ns}$, was found empirically by

$$|G(j\omega_{ns})| = -6 \text{ db}$$

The sinusoidal damping coefficient $\zeta_s$, was determined by plotting gain versus $\omega/\omega_{ns}$ on semilog paper and reading off its best-fit value from a set of parametric curves.

Figure 1 shows an example for one subject. A previous plot determined $f_{ns}$ ($\omega_{ns}$ = 2.5 x $2\pi$ rad/sec) for this subject. Four parametric curves representing 4 different values of $\zeta$ are also shown in this figure. The damping coefficient that provided the best fit is $\zeta_s = 0.8$. The range of $f_{ns}$ for the subjects studied is 2-4 Hz (average 2.8 Hz) and the range of $\zeta_s$ is 0.8-1.0 (average 0.87). All subjects have over-damped response.

2. Ramp targets

A sweep speed of 50 msec/cm in a 10 cm scope was found desirable for the target tracking of a ramp function. Oscilloscope sweep speeds faster than this were too rapid and slower sweep speeds slower than this were easy for the subject to follow. In retrospect, since all subjects can perform well up to 2 Hz in sinusoidal tracking, a full screen of approximately 500 msec is, indeed, the speed of choice.

The response of one subject tracking a ramp function is shown in Figure 2. The slope of the ramp was approximately 250 gm/sec. At time t=0 ($\omega_{nr} t$ =-2), the ramp begins to rise. There was a delay of approximately 80 msec--corresponding to the time the subject noticed a change in the target's baseline to a response by the subject to follow it. Again, by matching up the response to a set of parametric curves characterizing the behavior of a unit gain, second-order, low-pass linear system, $\zeta_r$ and $\omega_{nr}$ ($\zeta$ and $\omega_n$ to ramp target respectively). For the specific example in Fig. 2, $\zeta_r$ = 0.4 and $\omega_{nr}$=24 rad/sec (3.8 Hz). The range of $\zeta_r$ for the subjects studied is 0.1 to 0.4 (average 0.3) and the range of $\omega_{nr}$ is 4-6 Hz (average 4.6 Hz). All subjects have under-damped response.

DISCUSSION

The mean twitch contraction time of motor units of the first dorsal interosseus (a primary muscle for the lateral motion of the index finger) is approximately 70-80 msec, and the mean half-relaxation is 60-70 msec. Working with these numbers it is not likely that the natural frequency of human motor responses, $\omega_n$ , can be higher than 10 Hz. This is consistent with this study and that of Bawa and Stein (1976).

Different values of $\zeta$ and $\omega_n$ were found depending on the task of the subject. It is suggested that the motor system may be able to vary $\zeta$ and $\omega_n$ in accordance with the intended performance. In this study, it was found that as $\omega_n$ increase, $\zeta$ decrease. These observations lead us to suggest that $\zeta$ and $\omega_n$ may not be independently adjustable by the motor system.

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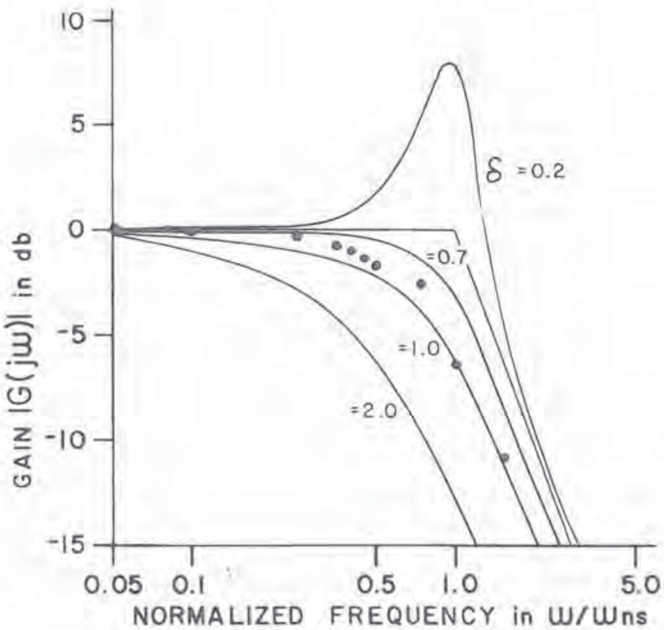


Fig. 1 Frequency response characteristic of one subject to sinusoidal targets.

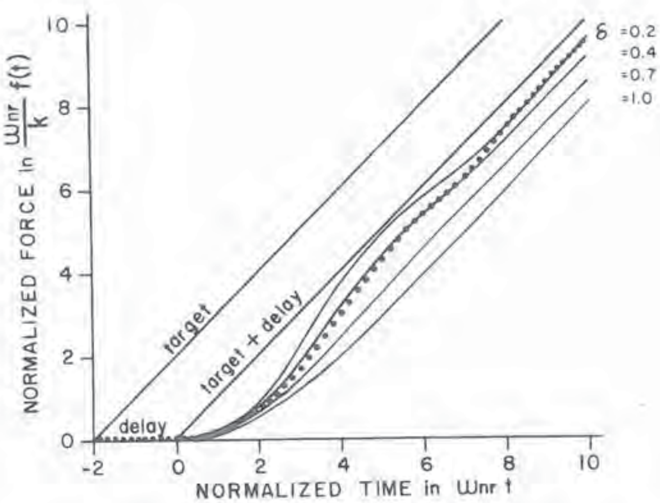


Fig. 2 Response of the same subject shown in Fig. 1 to ramp targets.

ACKNOWLEDGMENT

We are grateful for the help of Dr. Charles Abzug and Dr. Richard Tomasulo.

# THE EFFECTS OF THE TM-SIDHI PROGRAM ON THE PAIRED HOFFMAN REFLEX RESPONSE

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During the last decade, numerous studies have been conducted on the physiological effects of meditative techniques and especially on the Transcendental Meditation (TM) technique as taught by Maharishi Mahesh Yogi (Wallace, et al., 1971, Orme-Johnson & Farrow, 1977). A number of researchers have reported that during the practice of the TM technique, as taught by Maharishi Mahesh Yogi, subjects achieve a physiologic state which is characterized by a quiescence of metabolic and autonomic activity as measured by such changes as decreased breath rate, $O_2$ consumption, plasma cortisol, and phasic skin responses, and by a state of restful alertness as measured by specific changes in EEG activity such as increased alpha and theta coherence, especially in frontal and central areas of the brain. Studies comparing TM meditators with non-meditating controls have also reported a wide variety of changes such as: faster habituation to stressful stimuli, faster reactions, increased field independence, increased self-regard and self-actualization, decreased anxiety and neuroticism, increased intelligence, and improvements in conditions such as hypertension, asthma, and a number of psychiatric disorders. Studies have recently been conducted on participants of an advanced TM program known as the TM-Sidhi program which involves learning the performance of certain traditional "sidhis" or "powers" such as friendliness, knowledge of the body, disappearing, and flying, and which are designed to develop maximum neurophysiological integration. Subjects with clearer subjective experiences of the TM-Sidhi program have higher intra and inter hemisperic EEG coherence in alpha and theta frequencies than subjects with unclear experiences. Further, subjects with clearer experiences also score higher on the Torrance Test of Creative Thinking and show a faster recovery of the paired Hoffman (H) reflex response.

The following study on the paired H-reflex response was conducted as part of an ongoing longitudinal study on the TM-Sidhi program.

## SUBJECTS AND DESIGN

A total of 13 male and 9 female students at Maharishi International University participated in the study. The experimental group consisted of 14 subjects who were tested before and after beginning a three-month TM-Sidhi course and the control group of 8 subjects who were practicing the TM technique but not the TM-Sidhi techniques and were tested before and after the same three-month period.

## PROCEDURE

The H-reflex response was elicited by percutaneous electrical stimulation of the posterior tibial nerve by a $2cm^2$ silverplated electrode in the region of the popliteal fossa and a $20cm^2$ electrode plate placed on the patella. The $2cm^2$ stimulating electrode was fixed securely only after a position was found which produced the maximum H wave (H-reflex response) preceded by the smallest M wave (direct motor response). Stimulus intensities were also set so as to obtain maximal H wave and the lowest M/H ratio. The stimuli were square wave pulses, 1 ms. in duration, delivered from a Grass stimulator. Electromyographic signals were recorded from bipolar surface electrodes (2cm apart) located longitudinally over the soleus muscle approximately 6cm below the gastrocnemius. The EMG signals were amplified and displayed and peak to peak amplitudes recorded from a Tektronix storage oscilliscope.

Subjects were seated in a comfortable chair, their left leg extended with the knee flexed about 120°, and their foot resting on a support. The experimenter and stimulating and recording apparatus were located in another room so as not to disturb the subject who was asked to sit comfortably with eyes open and merely look at the wall. Pairs of stimuli were given at intervals of 50, 70, 100, 150, 200, 333, 500, and 1000 ms. Each pair was given three separate but consecutive times (10 sec. apart) and the results were taken as the average of the three trials. At the beginning of each run on a subject, the M/H ratio, the M and H wave forms and the latency of the H wave was recorded, and at the end of the run the M/H ratio was again recorded to insure stability of electrode placement.

Questionnaires were given to subjects to assess the clarity of experience in the days' TM or TM-Sidhi practice. Further, at a separate time the subjects were given a standard neurological examination by a trained physician in order to test for any signs of abnormal motor responses.

## RESULTS

The H-reflex data for each individual was analyzed by calculating the ratio of the amplitude of the test response (H2) over the conditioning response (H1) for each time interval (delay between the conditioning and test stimuli).

An analysis of covariance was performed comparing pre and post measurements of the experimental with the control group. The pre score was used as the covariant. Significant differences in recovery of H-reflex were found for male but not female subjects at the following time intervals: 70, 150, 250, and 330 ms. The differences were in the direction of a faster recovery of H-reflex response after beginning the TM-Sidhi program. Further, male experimental and control subjects who independently reported clear experiences during their TM and TM-Sidhi program showed a significantly faster recovery of H-reflex response as compared to those who reported unclear experiences.

The results of the neurological examination showed all subjects to be normal.

## DISCUSSION

The results of the study clearly demonstrate that H-reflex recovery increases significantly in males as a result of participating in the TM-Sidhi program. The reason why the changes were only seen in male and not female subjects is unknown.

The precise physiological mechanism for the recovery of H-reflex is still very much under investigation, however a number of researches have suggested that changes in H-reflex recovery are due primarily to influences from central processes. Van Boxtel (1974) in an attempt to determine the importance of central processes on spinal reflexes measured EEG alpha activity in subjects at rest as an indication of supraspinal activation and amplitude of H-reflex as an indication of reflex activity. The results showed that a constant alpha index (percent of time alpha was present) was accompanied by stable reflex amplitude and that a decreasing alpha index (indicative of drowsiness) was accompanied by decreasing reflex amplitude.

Whether the longitudinal changes in H-reflex in TM-Sidhi subjects is due to local spinal influences or more central supraspinal alterations is unknown. The fact that the H-reflex recovery response in TM-Sidhi participants has been correlated with higher levels of EEG coherence is suggestive of a common supraspinal influence. The mechanism of this common supraspinal influence would necessarily be different from those mechanisms which have been suggested for the increased motor excitability seen in certain clinical conditions, since neurological examinations showed all subjects to be normal. Further, since H-reflex recovery in TM-Sidhi participants has also been correlated with higher levels of creativity, and since numerous studies have demonstrated improved physical and mental functioning as a result of the TM technique, we would suggest that the changes reported here are indicative of a more optimal style of neurophysiological functioning, one in which the nervous system is less biased by prior conditioning and more uniform and accurate in its responsiveness.

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## ACKNOWLEDGMENTS


The authors wish to express their thanks to Dr. Stanley Kater, Dr. Doyne Lloyd, Dr. Michael Dillbeck, Chris Haynes, Paul Mackowiah, Robert Druker, Rivca Rosenfeld, Nancy Rosenberg, Sylvia Jansen, Jim Jackson, and Rita Toibman for their assistance in conducting the research. This study was supported in part by the Rogers Foundation.

NOTES

# KINESIOLOGY II

## Session 3A

CO-CONTRACTION OF BIFUNCTIONAL MUSCLES : A KINESIOLOGICAL STUDY

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## INTRODUCTION

According to classical data, the effect of bifunctional muscle contraction is never limited to one joint (see BASMAJIAN, 1957 ; JOSEPH, 1973 ; FUJIWARA and BASMAJIAN, 1975). Therefore, when a bifunctional muscle is involved in producing a monoarticular motion, a simultaneous activation of antagonistic muscles (namely a "co-contraction") is required to prevent the unwanted movement. In the present investigation, an attempt was made to obtain a quantitative picture of this phenomenon by studying the behavior of two biantagonistic muscles during voluntary efforts.

## METHODS

The subjects were seated with the right foot held on the plate of an ergometer, the knee being flexed at 90 degrees and the ankle slightly plantarflexed. The horizontal axis through the medial maleolus coincided approximately with the rotation axis of the mechanical system. During the course of each experiment, the subjects were required to perform combined efforts. First, they were asked to maintain a given isometric torque of abduction by means of a weight suspended from a wire running over a pulley, the other end of the wire being attached around the tarsus. Then, after a few seconds, an effort of plantarflexion was associated with the isometric torque. For this purpose, the plate of the ergometer was linked to a piezoresistive dynamometer. The signal from this transducer was used to provide a visual reference to help the subject maintain a predetermined level of plantarflexion.

Surface EMGs of two bifunctional muscles (tibialis anterior : dorsiflexor, adductor ; peroneus longus : plantarflexor, abductor) were detected and integrated using an analogical numerical converter. This apparatus delivered a number of impulses per second (or "pips") which was proportional to the total area of the EMG. The EMG activities and the output signal of the dynamometer were simultaneously recorded on photosensitive paper by means of moving-magnet oscillographs of appropriate frequency.

Five subjects were given two tests each. From one experiment to another, the value of the isometric torque of abduction was held constant (4 N.m) and the effort of plantarflexion varied from 2.5 N.m to 12.5 N.m. For each intensity of combined efforts, about ten trials were done by each subject.

## RESULTS

1) Initial phase of abduction. For each subject, the peroneus longus and the tibialis anterior were simultaneously active. Despite the constant value of the isometric torque, both muscles exhibited a level of electrical activity which varied from one trial to another. However, a linear relationship was found between the integrated EMGs of the tibialis anterior (QTA) and the peroneus longus (QPL){Fig. 1}. In all cases, the scatter was small (Bravais-Pearson coefficient $r = 0.75\text{-}0.82$, $df = 50$). Furthermore, as attested by the mechanical record, a slight extension was observed during this phase.

2) Plantarflexion associated with abduction. With some latency after initiation of plantarflexion, the discharge of the tibialis anterior disappeared while, in most cases, the activity of the peroneus longus was increased. In some trials (i.e. when the tibialis anterior was highly co-contracted in abduction), the peroneus longus was paradoxically less active than in abduction.

## DISCUSSION

The present results and those previously reported (PEROT and GOUBEL, 1978) show that the tibialis anterior is normally active in abduction. These findings are in close agreement with those of O'CONNELL (1958) which remained "unexplained and unconfirmed" (BASMAJIAN, 1974) In fact these data are consistent with the idea that a bifunctional muscle underlies a process of co-contraction : during the initial phase of abduction, the activation of the peroneus longus leads to a plantarflexion torque and this unwanted effort is prevented by the recruitment of dorsal flexor muscles (i.e. the tibialis anterior) working as antagonists of plantar flexors.

This interpretation, in terms of "anti-extension" activity, is supported by the fact that when plantarflexion is authorized, the tibialis anterior becomes silent. Furthermore, this muscle is also an adductor. Then, during abduction, it also works as an "anti-abductor" and induces a conflicting situation which can explain i) the variations of $Q_{TA}$ and $Q_{PL}$ for maintaining a given torque of abduction, ii) the development of a slight extension torque and iii) a decreasing activity of the peroneus longus in plantarflexion when the tibialis anterior is highly active in the initial phase and develops an important torque of adduction.

Finally, the activations of both the biantagonistic muscles follow a similar evolution, as attested by the $Q_{PL}$-$Q_{TA}$ linear relationship. This relationship could reflect a property of the motor-program used in such a complex situation.

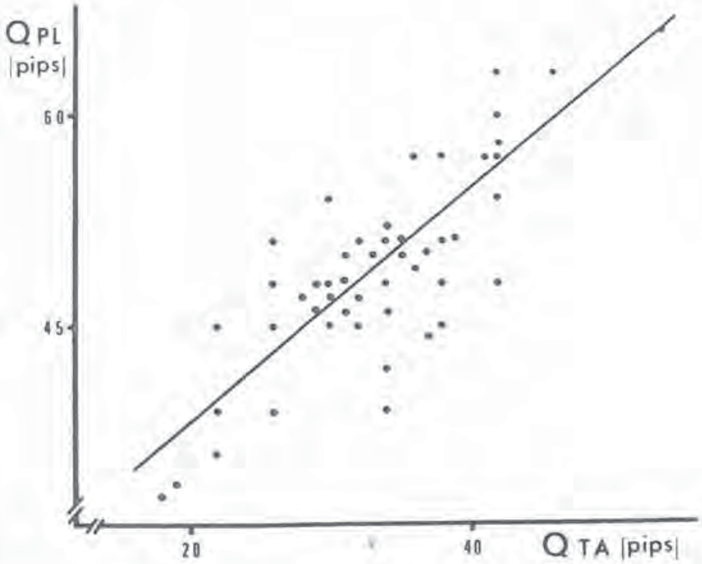


Fig. 1. Relationship between integrated EMGs of the peroneus longus ($Q_{PL}$) and the tibialis anterior ($Q_{TA}$). Results for one subject for constant isometric torques of abduction (4 N.m). $Q_{PL}$-$Q_{TA}$ regression line is shown (r = 0.79)

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# ELECTROMYOGRAPHIC AND ANATOMICAL STUDY OF THE HUMAN TENSOR FASCIAE LATAE MUSCLE.

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## INTRODUCTION

The tensor fasciae latae muscle is generally believed to be active during movements involving either flexion, abduction, or medial rotation of the thigh. It has also been suggested that the muscle may act via the iliotibial tract to extend or maintain extension of the leg and to produce outward rotation of the leg when the knee is flexed. To date, however, electromyographic studies do not agree with regard to the relative level of activity exhibited by this muscle during each of these simple movements. Thus, there exists a variety of views on its actual functional role during normal gait. In preliminary tests in which electrical activity was recorded from one indwelling bipolar fine-wire electrode, we observed variation between experiments on the same subject for relative activity levels during different simple movements. The present study was performed to investigate the cause of this observed intrasubject variation.

## EXPERIMENTS

### Electromyographic

Ten normal adults (five men and five women aged 20 to 36 years) volunteered for this investigation. Since the observed intrasubject variation might have been due to electrode placement, three or four 50 um nylon insulated wires prepared in a manner similar to that suggested by Basmajian and Stecko (1962) were inserted along a line transecting the muscle belly of the right thigh. The wires led to a 4-channel FM telemetry transmitter package (Bio-sentry Telemetry, Inc., Torranic, CA) which was affixed to a belt worn by the subject. Our receiving and recording apparatus (Stern et al., 1977) provided videotape records of electrical activity perfectly synchronized with subject behavior. Each subject performed a minimum total of sixty-one planned simple exercises from the following positions: stance on left limb-right lower extremity free, stance on right lower extremity only, stance on both lower extremities, seated, and supine. In addition, records of sixteen various bipedal locomotor behaviors were obtained. Total time required for each experiment, including subject preparation was two and a half to three hours.

### Anatomical

Dissections of the lateral aspect of one or both thighs of six cadavers were performed. The specimens were fifty-five years or older and of varying body build. The outer transverse layer of the fascia lata was first removed as well as the fascial layer covering the tensor fasciae latae muscle. This allowed tracing the tendinous fibers from the muscle to their insertions.

## DATA ANALYSIS

### Electromyographic

Frame by frame playback of videotape records (60 frames/sec) allowed observation of relations between subject movement and presence or absence of muscle activity. An estimate of the degree of activity for each electrode channel during the planned simple exercises was made by comparing any particular burst to what has been established as the maximum for that channel during the experiment. A somewhat similar analysis was carried out for muscular activity noted during the various bipedal behaviors. For walking, jogging and running, 10-step cycles of each behavior for each subject were used to prepare graphs of activity versus position that reflect the variability between subjects. Such graphs were constructed for only the most anteromedial and most posterolateral fibers since they were found to have phasic activity periods vastly different from each other.

## RESULTS

### Electromyographic

A distinct difference in activity pattern is apparent between the most anteromedial and most posterolateral fibers of the muscle. The anteromedial fibers act during simple movements requiring hip flexion; whereas, the posterolateral fibers are active in exercise requiring thigh abduction and medial rotation. Simple non-strenuous exercises designed to distinguish abduction activity from medial rotation activity in the posterolateral fibers indicated that these fibers ordinarily act in abduction only when medial rotation occurs concurrently. Activity in the most posterolateral fibers is always present during medial rotation with or without abduction. No significant activity is present in any region of the muscle during isolated knee movements. For each subject activity patterns during walking, jogging and running were similar, but at faster speeds there occurred an increase in relative degree of activity and an increase in percent of cycle activity. Variability between subjects was more pronounced. Curves for phasic activity during walking showed no significant activity in the most anteromedial fibers. In contrast, the most posterolateral fibers are active shortly before and continuing until slightly after heel strike. They again became active near midstance in most subjects. Three subjects showed a short burst of activity at toe-off. As speed of progression increased from walking to jogging to running, the most anteromedial fibers became increasingly active in all subjects after toe-off until patellar-cross of the swing phase. The most posterolateral fibers became increasingly active both in relative amplitude and duration at heel strike. A few subjects again showed lesser activity for variable periods either slightly before or after toe-off.

### Anatomical

The tendinous fibers of the anteromedial portion of the tensor fasciae latae course longitudinally down the thigh ending as they curve anteriorly just distal to the patella. None of these fibers continue to the lateral tubercle of the tibia. Tendinous fibers from the remaining posterolateral part of the tensor fasciae latae course distally and somewhat posteriorly to join a portion of the fascia lata known as the iliotibial tract which originates from the iliac crest. The iliotibial tract then continues to its insertion on the lateral tubercle of the tibia. When the tensor fascia latae, gluteus maximus and intervening iliotibial tract are transected and reflected, the distal course of the tendon fibers of the most posterolateral portion of the tensor fasciae latae may be traced. These tendon fibers at first lie between the inner transverse layer and middle longitudinal layer of the iliotibial tract and soon join a strong tendon bundle from the cranial portion of the gluteus maximus which inserts on the femur.

## DISCUSSION AND CONCLUSIONS

It is well known that certain large muscles in the human body exhibit different activity patterns in different fiber areas. This possibility has been largely neglected in previous functional electromyographic studies of smaller muscles since it has been commonly assumed that bipolar fine-wire electrodes record unfiltered electrical activity from a relatively large expanse of muscle except when fascial planes or relatively extensive connective tissue separations exist between fiber bundles. Our findings that the most anteromedial and most posterolateral fibers of the tensor fasciae latae exhibit different relative activity levels during simple planned exercises and different phasic activity patterns during locomotor activities suggest that previous variable descriptions for this muscle in the literature might be due to electrode placement. However, some true intersubject variability does indeed exist in our data. Future reports of experiments now in progress will address intersubject variation.

Our data suggest that the most anteromedial fibers of the tensor fasciae latae play no active role during walking other than to assist in flexion of the thigh during the initial period of acceleration. As speed increases from jogging to running, we believe this portion of the muscle plays an increasingly important role in rapid deceleration and acceleration of the thigh. The most posterolateral fibers, however, have an active role during walking. We suggest that their activity during support in walking through running is related to abduction and medial rotation at the hip. Dissections of anatomical material do not disagree with these suggested functions. The longitudinal course of the anteromedial tendon fibers is quite different from the distal and progressively more posterior course of the posterolateral tendon fibers. The revelation that the most posterolateral tendon fibers join a tendon bundle from the cranial portion of the gluteus maximus inserting on the femur argues strongly for a medial rotatory function in addition to an abductor function for the posterolateral portion of the muscle.
Supported by NSF Research Grant BNS 7683114A01.

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KINEMATIC AND ELECTROMYOGRAPHIC ANALYSES OF THE STRENGTH OF THE ABDOMINAL AND BACK MUSCLES

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## INTRODUCTION

In recent years, much attention has been given to the role of the abdominal and back muscles in the supporting ability of the spine especially in the field of research of lumbar spine diseases. Although electromyographic and kinesiologic studies have been useful in clarifying the dynamic features of the trunk muscles, a better method of quantitative evaluation of their strength is still being looked for. This paper reports a new method of quantitative measurement, which uses a cybex machine, and also a comparison of the measurements by it and conventional electromyography.

## MATERIALS AND METHODS

One hundred normal subjects, 50 men and 50 women, without any complaints referable to the spine were used for the study. The subjects were evenly distributed among the second through sixth decades in both sexes.

A cybex machine, which has been used for the extremity muscles, was used for the trunk muscles. A newly devised bar was connected to the lever arm of a dynamometer. The axis of rotation of the arm was adjusted at the hip joint, and the bar was placed on the xyphoid process for the abdominal muscles and on the lower pole of the scapula for the back muscles. Each subject was ordered to make both isometric and isokinetic motions to obtain a muscle torque curve. Then, for muscle fatigue curves the subjects produced a sustained isometric contraction and repeated isokinetic contractions. Strength decrement index ( SDI ) proposed by Clarke et al.(1954) was calculated from the formula:

$$\frac{\text{Initial muscle torque - Final muscle torque}}{\text{Initial muscle torque}} \times 100 = \text{SDI}$$

Simultaneously, electromyographic recordings were taken with surface electrodes and subjected to quantitative analysis both by a reset type integrator and by counting the number of discharge spikes.

## RESULTS

The results of maximal muscle torque determinations suggested that the abdominal muscles might usually be weaker than the back muscles. Influence of aging was seen in both muscles; it was slightly greater in the abdominal than in the back muscles of the women (Table 1). Aging did not have much influence on muscle fatigue curves. Generally speaking, the abdominal muscles were fatigued more easily than the back muscles, and both were fatigued more easily by a sustained contraction than by repeated contractions(Table 2). A high correlation was found between the muscle torque value and integrated-electromyographic value until fatigue occurred in the examined muscles. The influence of fatigue was greater in the abdominal than in the back muscles (Fig. 1 and 2).

## DISCUSSION

Various methods of quantitative measurement were proposed for the skeletal muscles, but a cybex machine, based on the idea of isokinetic excercise, was the first to permit continuous measurement and recording of muscle strength. So far, this machine has been used mainly for the extremity muscles, but in this study it was adapted for use in measuring the trunk muscles. The machine is simple to use and gives fairly good results, and actually is being utilized at our clinic for low back pain patients.

This study indicates that the muscle torque value is closely related to the integrated electromyographic value, suggesting that the machine is useful in quantitative meaurement of the trunk muscles.

## CONCLUSION

A cybex machine is suitable for quantitative measurement of the trunk muscles in clinical cases. The abdominal muscles are apt to be weaker than the back muscles. Influence of aging is definitely seen in both abdominal and back muscles. Both are fatigued more easily by a sustained isometric contraction than by repeated contractions. The abdominal muscles are apt to be fatigued more easily than the back muscles.

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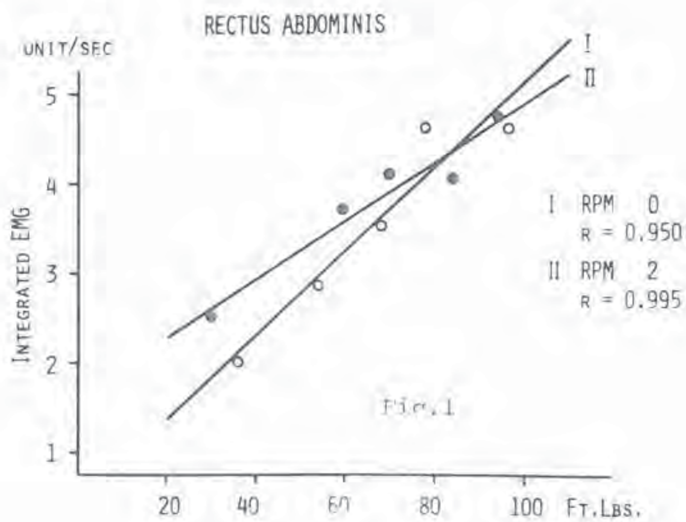
Table 1.

| Age | Abdominal Muscle isomet. | Abdominal Muscle isokinet. | Back Muscle isomet. | Back Muscle isokinet. |
|---|---|---|---|---|
| male | | | | |
| 10-19 | 121.1 | 119.8 | 151.4 | 140.9 |
| 20-29 | 131.0 | 130.3 | 158.6 | 149.0 |
| 30-39 | 124.6 | 116.7 | 154.5 | 141.5 |
| 40-49 | 87.6 | 87.1 | 115.5 | 116.8 |
| 50-59 | 74.1 | 79.4 | 106.1 | 100.8 |
| female | | | | |
| 10-19 | 77.0 | 72.4 | 110.0 | 87.9 |
| 20-29 | 63.6 | 68.1 | 96.5 | 99.0 |
| 30-39 | 66.4 | 65.6 | 101.3 | 97.1 |
| 40-49 | 52.2 | 57.0 | 94.7 | 87.9 |
| 50-59 | 36.6 | 30.3 | 66.7 | 54.6 |

Table 2.

| Age | Abdominal Muscle isomet. | Abdominal Muscle isokinet. | Back Muscle isomet. | Back Muscle isokinet. |
|---|---|---|---|---|
| male | | | | |
| 10-19 | 59.4 | 12.6 | 34.3 | 10.6 |
| 20-29 | 27.6 | 14.0 | 18.9 | 7.9 |
| 30-39 | 55.3 | 17.7 | 16.1 | 2.1 |
| 40-49 | 62.4 | 14.6 | 16.4 | 11.5 |
| 50-59 | 57.6 | 19.6 | 32.8 | 8.5 |
| female | | | | |
| 10-19 | 46.6 | 22.7 | 21.5 | 15.7 |
| 20-29 | 63.0 | 28.9 | 21.6 | 21.5 |
| 30-39 | 81.8 | 41.1 | 16.6 | 14.7 |
| 40-49 | 58.4 | 41.6 | 49.2 | 31.5 |
| 50-59 | 76.5 | 24.0 | 32.5 | 19.3 |



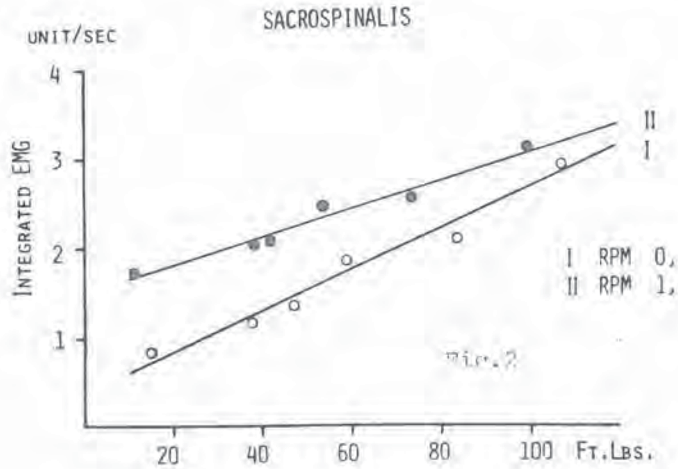




Table 1 Mean values of maximal muscle torque related to age.
Table 2 Strength decrement index related to age.
Fig. 1 Correlation between integrated-electromyographic values and muscle torque values of the rectus abdominis muscle.
Fig. 2 Correlation between integrated-electromyographic values and muscle torque values of the sacrospinalis muscle.

# EMG VERSUS ISOMETRIC FORCE AND MUSCLE LENGTH

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## INTRODUCTION

It is essential to know the relationship between the myoelectric signal and variations in muscle parameters due to joint motion before utilizing the myoelectric signal as a measure of muscle tension under anisometric conditions. For a constant level of excitation the isometric tension developed by a muscle depends on its length, [1]. The surface EMG, which measures the level of excitation, is related to the force exerted by the muscle, [2]. The present work was undertaken to investigate the relationship between EMG activity, isometric force and the muscle length. The results will be used to modify Patla's dynamic equation, [3], given below as Equation 1.

$$K\sigma = bMg + (Mg + a)V + bM\dot{V} + MV\dot{V} \quad \ldots \quad 1$$

where M is the mass of the load on the muscle, g is the acceleration due to gravity, V is the velocity of shortening of the muscle, a and b are constants determined by the muscle physiology, $\sigma$ is the RMS value of the myoelectric signal, and K is a constant of proportionality.

This equation describes anisometric contraction variables in terms of the myoelectric signal at a given muscle length. The proposed modifications will make the equation valid over the entire range of flexion, to facilitate continuous measurement of EMG and mechanical data.

Under isometric conditions, Equation 1 reduces to

$$K\sigma = bMg$$

and letting K and b become functions of muscle length (joint angle) gives

$$\sigma = K_t(\theta)F_t \quad \ldots \quad 2$$

where $K_t(\theta)$ is b/K and is the coefficient relating the RMS value of the EMG signal to the isometric force, $F_t$, exerted by the muscle.

The dependence of $K_t(\theta)$ on joint angle is due to its three component terms given by Equation 3.

$$K_t(\theta) = K_m(\theta)K_L(\theta)K_c(\theta) \quad \ldots \quad 3$$

where $K_m(\theta)$ is the mechanical gain of the lever configuration.
$K_L(\theta)$ is the efficiency of the muscle given by the length-tension curve.
$K_c(\theta)$ is the variation of the muscle's contribution to force production when more than one muscle is involved in the joint motion.

The main objective of this work is to experimentally obtain $K_t(\theta)$. However, if the contribution of the biceps brachii is assumed constant over the range of elbow flexion studied it is possible, knowing $K_m(\theta)$, to determine the length-tension diagram, $K_L(\theta)$, for the biceps brachii. This also is done in this work with interesting results.

## EXPERIMENT

The apparatus consisted of a horizontal platform, on which the right arm of the seated subject rested. In this position the palm of the hand was turned inwards to the subject. A known mass was applied, via a pulley system, at the wrist to give a load force in a direction approximately parallel to the humerus. The EMG signal was obtained with Beckman surface electrodes in a bipolar configuration. These were placed over the biceps, transversely to the longitudinal direction of the muscle. The transverse arrangement was chosen in order to reduce the effect of electrode displacement relative to the muscle due to changes in the joint angle. The EMG activity was differentially amplified to a suitable level and then stored on FM tape for future processing.

Data was collected from three male subjects. For each trial the subject was required to hold a known mass at a given angle of elbow flexion for a period of 10 seconds, during which time EMG data was collected. This procedure was repeated for a number of known masses and over a range of elbow flexion from $40^\circ$ to $140^\circ$ at $5^\circ$ intervals, (full extension is $180^\circ$). For each mass the angle data was collected in a pseudo-random fashion and sufficient rest was given between measurements to avoid the effects of muscle fatigue. The recorded EMG was processed and the RMS value of the signal was used as a measure of the level of muscle activity.

## RESULTS

For any joint angle, $\theta$, the relationship between the RMS value, $\sigma$, of the EMG signal and the resultant torque acting at the wrist was found to be linear with correlation coefficients ranging from .90 to .99. The slope of the regression line, which gives $K_t(\theta)$, decreased approximately exponentially with increasing joint angle as shown in Figure 1. Knowing the mechanical gain of the lever configuration, the normalized force developed in the muscle may be calculated. Figure 2 is a plot of the RMS value, $\sigma$, of the EMG signal versus the normalized force developed in the muscle for a number of muscle lengths. A family of length-tension curves may be constructed for different levels of constant excitation. Such a relationship is shown in Figure 3. It is observed to have the classic shape of a length-tension diagram with force decreasing linearly below the rest length, a plateau region of constant force about the rest length and an exponential increase in force beyond the rest length.

## CONCLUSION

The coefficient relating the RMS value of the EMG activity to the force exerted under isometric conditions was shown to decrease exponentially with increasing joint angle. This result provides information for suitable modifications to Patla's dynamic equation to account for variations due to joint angle.

Based on the assumption of a constant contribution to force production at all joint angles, a family of length-tension curves may be constructed for the biceps from the surface EMG data. The length-tension curve expresses a fundamental property of the contracting muscle and finds applications in functional electromyographic studies.

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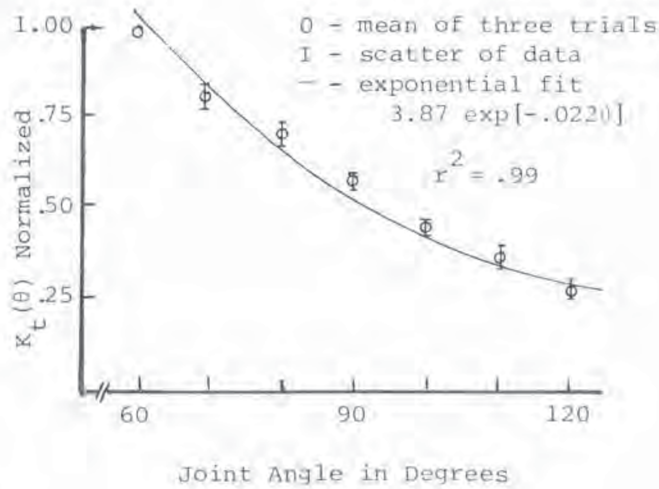


FIGURE 1

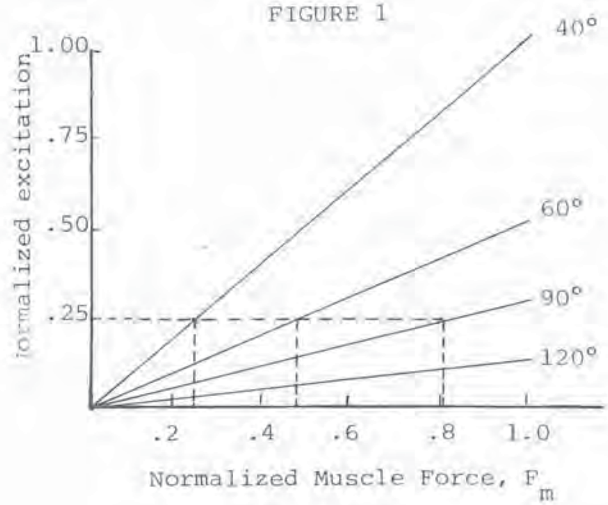


FIGURE 2

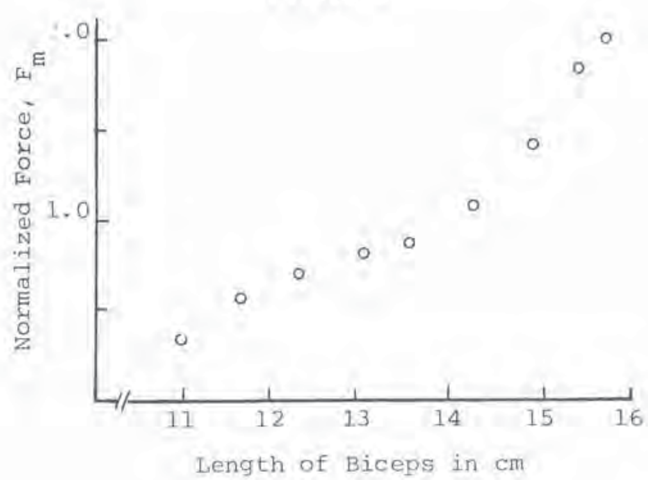


FIGURE 3


## ACKNOWLEDGEMENTS

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NOTES

# PROSTHESES

## Session 3B

# THE INTERFERENCE BETWEEN ELECTRICAL STIMULATION APPLIED FOR PROSTHESES' SENSORY FEEDBACK AND MYOELECTRIC CONTROL SYSTEMS

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## INTRODUCTION

It is generally recognized that access to sensory feedback will increase the performance of a motorized prosthesis (Mann & Reimers, 1970). Many authors have suggested the use of electrical stimulation of the skin or the nerves for this purpose as electrical stimulation possesses several advantages in relation to mechanical ones (Shannon, 1976; Anani et al., 1977). Electrical stimulation will cause interference with the myoelectric control systems. The purpose of this investigation is to determine if electrical stimulation can be applied to nerves in the amputation stump without causing interference to such a degree that it will disturb the prosthesis control system. This is important for the possibility to make the prosthesis self-contained. Self-containment is according to Childress (1973) vital for the patient acceptance of a prosthesis.

## MATERIALS AND METHODS

Two groups of experiments were performed. In one group the interference was studied "in vitro", i.e. the stimulation and pick-up situation of the amputee's forearm was simulated by an experimental set-up consisting of a bowl with saline solution on the surface of which a porous cloth was mounted and soaked with the solution. The other group of experiments were performed on forearms of non-amputated subjects.

The stimulating and recording equipments were the same in both groups of experiments. The stimulating electrodes were two 50 µm insulated wires with a deinsulated, hook-shaped tip of 2.5 mm. They were introduced to the proper position in the arm or cloth through hypodermic needles. The stimulating equipment delivered square-shaped pulses. The tested stimulation parameters included current intensities from 0.65 to 1.3 mA and frequencies from 10 to 80 Hz with a constant pulse duration of 100 ms. This selection of parameters was based upon a study of afferent electrical nerve stimulation for sensory feedback purposes (Anani et al., 1977).

The interference was studied by recording the output signal from two kinds of pick-up electrodes. One was the Otto Bock electrode characterized by an on-off response mode and the other was an electrode designed for use in a pattern recognition control system. In this electrode the myoelectric signal was amplified, band-pass filtered and rectified and the output was a DC voltage proportional to the myoelectric signal.

The procedure in each experiment was to study the interference of the electrical stimulation with the output of the EMG pick-up electrodes at various stimulation parameters as a function of the distance between the stimulation electrodes and the pick-up electrodes (the inter-electrode distance). Identical experiments were done on the human subjects and "in vitro". The "in vitro" experiments were done in order to obtain greater accuracy in the measurements than was possible with the human subjects.

## RESULTS

The pattern of interference with the electrode providing the proportional output for a 1 mA, 30 Hz current is shown in Figure 1, where the interference voltage is plotted as a function of the inter-electrode distance. When the stimulation parameters were altered it was noted that within the range relevant to intraneural stimulation the interference was more affected by a variation of the frequency than of the amplitude. This is due to the pick-up electrode filtering characteristics. The curve in Figure 1 was obtained from an "in vitro" experiment. The pattern of interference obtained from the subjects was identical although less reproducible due to practical difficulties.

Figure 2 is a summary of the experiments performed on the human subjects using the Otto Bock electrode. The possibility to adjust the amplification of that electrode was taken into consideration. For this electrode the effect of increased frequency was smaller than for the proportional output electrode.

## DISCUSSION

The efforts to develop artificial feedback systems for motorized prostheses have during the latest years been increasingly directed towards the use of electrical stimulation instead of vibrators. The reason for this is the possibility to contain an electrical stimulator and its control circuits in the prosthesis itself in addition to reliability and low energy consumption. Electrical stimulation, however, causes interference with the prosthesis control system. It is therefore necessary to investigate the extent and nature of this interference prior to clinical application of an electrical feedback system to a motorized prosthesis.

The results of the measurements of the interference on the subject and "in vitro" were compared to computer calculated patterns of interference based on the electric field theory. This comparison showed that with some approximations it was possible to predict the pattern of interference for different electrode arrangements and stimulation parameters. Thus, unexpected trouble with interference between the stimulating current and the prosthesis control system in clinical applications can be avoided.

The practical tests and theoretical calculations indicate that for stimulation parameters causing maximum possible interference relevant to afferent electrical nerve stimulation, an inter-electrode distance of 60 mm is enough to ensure total lack of interference with the pick-up electrode providing the proportional output in an on-off operated conventional control system. If it is not necessary to utilize the whole accessible range for the nerve stimulation parameters, this distance can be considerably reduced. Thus, for 1.0 mA and 30 Hz the critical inter-electrode distance for interference is less than 30 mm (Figure 1). Together with pattern recognition control techniques with compensation of the control algorithm for the interference the inter-electrode distance can be 75 mm without any limitations of the stimulating parameters. The Otto Bock electrode with the commonly clinically used amplification of 4 allowed an inter-electrode distance of 60 mm without any interference.

The interference voltage is dependent upon the stimulation current applied. For electrical nerve stimulation this current can be about 1/10 of the current necessary for electrical skin stimulation. Thus, from this point of view nerve stimulation is superior to skin stimulation.

## CONCLUSION

This investigation shows that with regard to the interference with the prosthesis control system, the use of electrical nerve stimulation for prostheses sensory feedback is compatible with prostheses self-containment in most amputation stumps.

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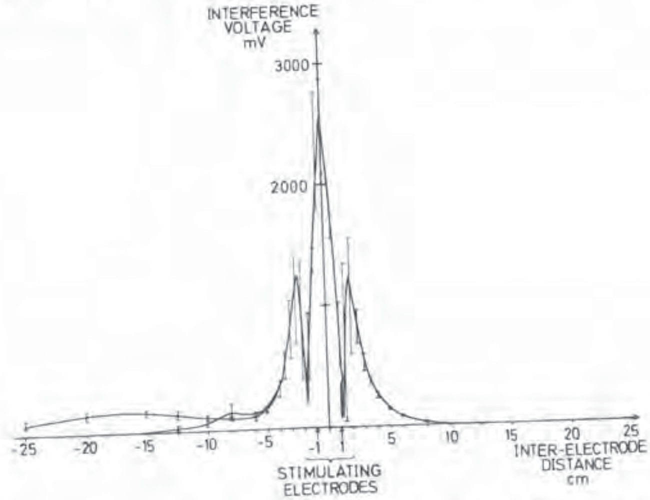


Fig. 1. The interference voltage is plotted as a function of the inter-electrode distance for various reference electrode positions. Stimulating reference at: -10 and -25 cm. Pick-up reference at: -25, -10, 10, and 25 cm. The location of the pick-up reference does not affect the interference, but around the stimulation reference there is an apparent increase in interference voltage.

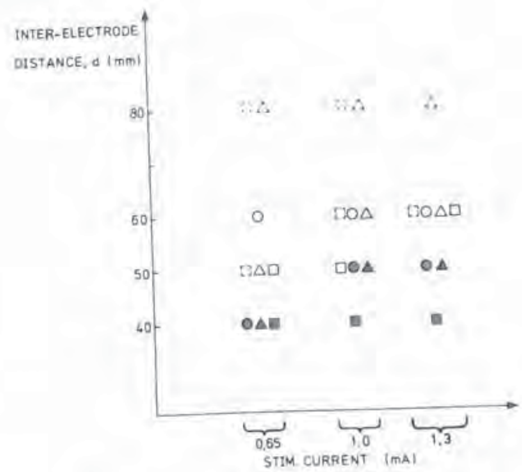


Fig. 2. Interference of electrical stimulation on the Otto Bock electrode. Symbols denote the shortest inter-electrode distance where an off response of the electrode was maintained during stimulation at various stimulation parameters and electrode amplifications.
Circles indicate stimulation at 80 Hz, triangle stimulation at 30 Hz, and squares stimulation a 10 Hz. Dotted symbols indicate electrode amplification 6, open symbols amplification 4 and filled symbols amplification 2.

# A NEW CONCEPT FOR KNEE ENDOPROSTHESIS

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## INTRODUCTION

Replacement of the natural joints of the limbs by artificial substitutes has become an acceptable technique in the treatment of deformed and painful joints. The experience with total replacement of the hip joint is long and successful enough so as to perform the arthroplasty on patients with life expectancy of 20 years and more. Knee arthroplasty has been initiated more recently and is gradually gaining popularity. However, the knee is more complicated joint from the kinematic and anchorage points of view. The loosening the wear are therefore more severe and more likely to happen in the knee than in the hip joint replacement. Since the age of the potential receivers of artificial knees is currently reduced, it is becoming very important to devote more attention to the mechanical wear of the prostheses.

The present work deals with special type of knee joint which performs the natural kinematics of the leg segments by pure rolloing of the mating surfaces one on top of the other. This is a conceptual change in the design approach which may reduce the wear several folds due to the lack of sliding.

## KINEMATIC CONCEPT

Since the natural knee is a polycentric joint the kinematics of the femur and tibia can be described with the aid of instantaneous center of rotation. The locus of these centers is a line termed centrode. If the tibia is held fixed and the femur rotates, then the instantaneous centers of rotation forms the fixed centrode. If the femur is held fixed and the tibia rotates, then the movable centrode is obtained. If artificial mating joint surfaces were made in the exact shape and location of the two centrodes, then the precise kinematics would be obtained by rolling the two surfaces one on top of the other.

## EXPERIMENTAL WORK AND ANALYSIS

Data on the natural centrodes of the knee joint is available from literature (Frankel, 1971; Walker, 1972). However, since the results of the different investigators were not in full agreement, a series of tests were performed to identify the natural centrodes. Five normal subjects were used, three males (ages 25, 38, 58) and two females (ages 25 and 50). Their knee joints were x-rayed while bearing weight in six different positions of flexion. Two of the subjects were also x-rayed with the knee not bearing weight for comparative purposes.

The radiographs were then analyzed and the centers of rotation were located by a finite displacement method. The results obtained by this procedure were scattered in a small region of the femoral condyles and could not be fitted into a smooth and simple curve. These results were in agreement with those obtained by some other investigators and indicated that the measuring technique was not accurate enough for this purpose. This, however, is the only non-invasive technique for in vivo assessment of the knee centrode.

As a result of the failure to obtain centrodes which are suitable to serve as joint surfaces it was decided to reverse the approach, postulate mating surfaces and analyze their resulting kinematics.

## SYNTHESIS OF THE ROLLING SURFACES

The procedure of selection of rolling surfaces for the knee replacement was based on both good reproduction of the joint kinematics and surgical feasibility of the suggested layout. Since surfaces of high curvature are inadequate for this purpose, it was assumed that most possibilities could be covered by using a short arc of a circle or ellipse. Various combinations of circles and ellipses rolling one on top of the other on their concave and convex sides were examined. The postulated surfaces were superposed on the femoral and tibial condyles. A simulation of their rolling was performed with the aid of a digital computer, and the resulting kinematics of the femur in relation to the tibia was examined. The quality of the results was assessed by selecting three arbitrary points on the femur and comparing their trajectory, as a result of the rolling, to their real trajectory as obtained from the radiographs (Figure 1).

A criteria for optimization was used, based on the sum of squares of the deviations of corresponding points on the synthesized trajectory (by rolling) and the points on the normal trajectory. This was represented by one figure of the sum of squares. A systematic search was made by taking each postulated combination of rolling curves and shifting them laterally and vertically so as to obtain their optimal position in the space of the joint. This procedure was carried out with all the combinations and an optimal pair of surfaces was fitted to each of the five subjects whose joint kinematics was radiographically examined. As a result, five different combinations were obtained. The most suitable surfaces appeared to be two circular ones rolling one on the other on their convex sides. The next step in the procedure was to try to fit one pair of surfaces to all five subjects. This procedure could indicate whether the population can be categorized according to several sizes. For the limited population of the present work, one size was found adequate for all five subjects and performed very well in producing results similar to their natural knee kinematics.

Once the layout of the knee surfaces was established, the design had to be tested in relation to possible misalignment of the prosthesis during installation. Basically, it was assumed that linear error of installation in the range of ±5 mm could occur as well as angular errors of up to ±5°. This was again processed through the simulation procedure and the kinematic results obtained deviated very little from the intended design. This proved that the joint model was not very sensitive to possible surgical errors.

Finally, to ensure that no excessive stretch or laxity occurred in the soft tissues as a result of the function of the joint, the hypothetical stretch in the collateral ligaments was examined. A point, assumed to be in the vicinity of the insertion of the collateral ligament at the femoral condyle was chosen. The movement of this point was examined and compared to the natural trajectory of the same point as obtained from the radiographs. Again, the two trajectories were found to be very similar and therefore the ligaments was expected to function in its natural fashion.

## DISCUSSION AND CONCLUSIONS

The idea of producing a knee joint prosthesis based solely on rolling was examined. The basic intention of the present analysis was to suggest a design which will reduce significantly the wear of the prosthesis and reproduce the natural kinematics. The basic considerations which guided the work were:

1. Reproduction of normal kinematics.
2. No radical changes in the knee layout.
3. Utilization of the knee space only without exceeding the condylar contours.
4. Minimum bone cutting during installation.
5. Minimal effect of surgical errors on the performance of the joint.

All these points were considered in the analysis and it appears that the concept of a rolling prosthesis is feasible. The execution of the concept is also possible mechanically. The constraint of rolling without sliding can be introduced by performing a wedging type prominence and eminence on the mating surfaces.

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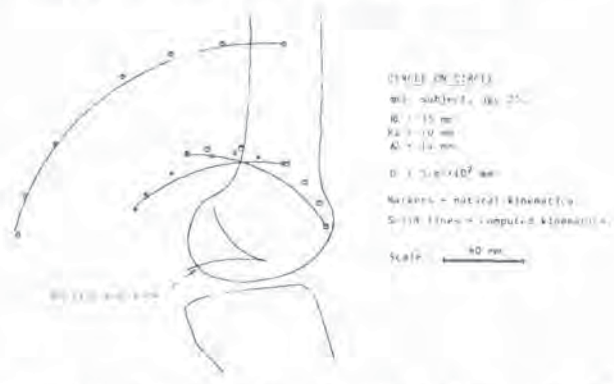


Fig. 1 Comparison between the kinematics of the femur as a result of rolling (continuous line) and the natural one (marked points).

## ACKNOWLEGMENT


The work was conducted at the Julius Silver Institute for Biomedical Engineering at the Technion, Israel Institute of Technology.

# DRIVE MECHANISMS FOR LIMB ORTHOSES

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## INTRODUCTION

Dexterity, harmonious motion and continuous adaptation during effort pertain to our ability to perform complex gestures. We owe this wonderful ability firstly, to the continuous cortical integration of an elaborate and highly complex neuro-muscular system and secondly, to the great potential of our articular kinematics.

This paper deals with the realization of driving mechanisms for a prehension orthosis which exploits one of our articular kinematic features. This feature which utilizes the control information derived from conscious and unconscious feedback pathways optimizes the energy required to perform a given task. This optimization is achieved by relating continuously both the speed of displacement and the force of a given body segment, to the resistance opposing the motion. It is interesting to note that as more force is required to do work, less speed and displacement are needed.

## STATE OF THE ART

Relating speed and force (or torque about an axis) to the resistance opposing the motion, is not a new mechanical concept. Car transmissions, for example, use the step by step gear-shift approach. The continuous approach has not been as widely used but it is found on some machine tools and occasionally on vehicles. For prosthetic or orthotic activation the last approach is certainly more attractive, since by continuously optimizing the speed and torque characteristics, it contributes to a smooth and more harmonious motion of the activated body segment.

With the exception of systems developed at the Rehabilitation Institute of Montreal (1) the above mechanical concept has not yet been utilized in prehension orthoses, although it is used differently in three externally powered hand prostheses. These three designs are based on the two speed approach, namely the high speed low torque in the unloaded state, and low speed high torque in the loaded state.

The Vaduz hand was probably the first prothesis incorporating this feature. In this design, the gear-shift from high to low was obtained only when the fingers reached the object to be grasped at a given minimum velocity. This approach reduces it's real effectiveness and restricts the control to the on-off mode.

The Otto-Bock hand is another example. In this case the automatic gear-shift occurs following the coiling of a spring which begins at a predetermined load. This approach has proved to be efficient despite the delay introduced by the coiling and uncoiling of the spring, especially when grasping flexible objects.

A third approach termed " Synergetic Prehension " has been described by D.S.Childress of Northwestern University. It has one possible drawback in that it requires two sets of motors and gear-boxes. The first set drives one or more fingers to produce rapid excursion at low force and the second set drives the opposing thumb to produce high force with small excursion. This concept is becoming more attractive with the emergence of more powerful magnets which lead to smaller and lighter motors.

## DRIVE MECHANISMS FOR PREHENSION ORTHOSES

For the past few years one of our objectives has been to produce an externally powered orthotic device which will restore some of the primary functions of the upper limb. The restoration of arm function in traumatic quadriplegic patients begins with the reduction of the prehension deficit, not only because it concerns a broad group of patients, but mainly because it is a prerequisite for tackling the other important arm deficits. To date our efforts have been, mostly directed towards the design of an adequate and useful externally powered prehension device. The activating mechanism together with the orthotic structure present a challenging design problem. In fact, the realization of an orthotic device, and particularly of a prehension orthosis, requires design efforts so as to minimize the weight and bulk factors while maximizing the prehension parameters.

The speed-torque relationship of a permanent magnet D.C. motor followed by a straight gear-box does not satisfy the above contradicting factors and parameters. Consequently, it became obvious that the addition of a mechanical " feature " which would relate both the speed of displacement and force of prehension was required. In an attempt to simplify the following description, the above mentioned mechanism will be termed a " Torque Converter ". Adding this unit to a drive mechanism does the following:

1) It minimizes the energy required to perform a given task since, assuming a low coefficient of friction in the moving parts, little energy is demanded for the entire working range of the unit (i.e. from high speed with low torque to low speed with high torque). This means that the motor size and weight can be appreciably reduced.

2) It maximizes the prehension parameters such as the grasping force, opening and closing speed and opening range.

3) Combined with a good control system, it can improve the quality of motion particularly if continuous changes in speed and torque are provided.

## TORQUE CONVERTER DESIGN

Several concepts have been tried and tested in our laboratory. With the exception of a unit having limited working range, continuous systems offering an acceptable level of efficiency, has up to now resisted miniaturization. The limited range unit was produced and installed on a prehension orthosis which was fitted in 1976 on a C5 quadriplegic patient. The mechanism is easy to construct and it's working principle is rather simple. Supplied with constant force from the drive mechanism, the torque converter mobilizes the finger segments through a preloading spring located along proximal phalange of the index finger. This spring elongates as the resistance opposed to finger motion rises. The elongation is then used to increase the leverage (i.e. the moment of force) acting on the metacarpophalangeal joint. The cosmesis of the unit was somewhat affected by the spring dimensions, but it's elongation has proved to be a valuable source of visual feedback about the force exerted. Following this successful fitting it was determined that a torque converter having a torque gain of 5 is probably optimum, considering factors such as mechanical friction and patient spasticity. Consequently a new design based on the two speed approach was initiated.

Working Principle (referring to figure 1)

The input force Fi from the motor and gear-box assembly is applied at P. The output force Fo activating the finger mechanism is derived from O'.

In the unloaded state, the high torque lever L2 is locked with respect to the low torque Lever L1 and the planetary gear G1 rotates freely over the internal gear segment. These conditions cause L1 to pivot around O (fixed axis) and this yields an output velocity and displacement which are twice as great as at the input.

In the loaded state, L2 is unlocked and linked to the gear G1 which now transfers the force to G2 and drives L1 at a reduced velocity but with higher force.

An average torque gain of 6 has been obtained with this system.

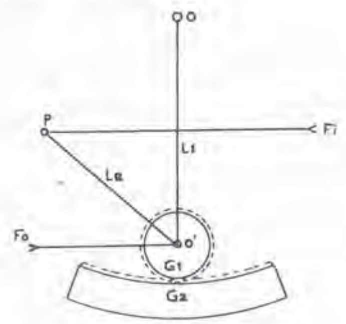


Fig. 1 Working principle of torque converter and layout of the various mechanical parts.

Fig. 2 New drive mechanism and torque converter assembly weighing less than 100gm.

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## ACKNOWLEDGMENTS

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# THE BOSTON ELBOW: A CASE FOR PROPORTIONAL MYOELECTRIC CONTROL

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## THE FIRST FIVE YEARS WITH THE BOSTON ELBOW

For five years the Liberty Mutual Insurance Company and other organizations have been fitting above-elbow and shoulder disarticulation amputees with a proportionally controlled myoelectric prosthesis called the Boston Elbow. We now have 40 patients wearing this device and have accumulated over 100 patient years of experience with the system.

The Boston Elbow project is typical of any technology that affects a relatively small population. A lot of time is required for debugging the system before one can make good observations on its utility. When just five amputees were wearing the Elbow, we might have concluded that good proportional myoelectric control was not possible. We now know that careful attention to all aspects of both electric and mechanical design is required to achieve a device that is the same device on every patient. Only after every patient has had his prosthesis updated with all the debugging modifications can meaningful data be collected.

## MYOELECTRIC HANDS ARE NOT PROPORTIONAL

We hear so often that hands work perfectly well with one speed that a few remarks are in order. The first is that many of the hands are really two speed. The motor operates at one speed but the mechanism shifts to a high torque mode when the fingers first encounter resistance to motion. If amputees find this satisfactory, then it is, ipso facto, all right.

The best way to explain the need for proportional control in an elbow is to discuss the way amputees use a variable speed prosthesis. If they carry out activities that cannot be done with a less sophisticated system, we must assume that they need the more advanced one.

## ELBOW AMPUTEES NEED PROPORTIONAL CONTROL

We have found that our patients both want and use a full range of speeds. Any malfunction that leads to loss of control at slow speed is immediately reported and repairs are sought. This type of problem is reported as "lack of control", and it results in the amputee being unable to position the prosthesis at a precise end point without either over or under shooting. On the other hand, many amputees want the elbow to flex faster than its present maximum rate and all make use of the fastest speed. One second to full flexion is too long when the intact elbow can achieve the same flexion in one-fourth the time. Thus both high and low speed are important to our amputees. These people use high speed to move the terminal device close to the desired position and slow speed to position it exactly.

## ARE PRACTICAL MYOSIGNALS PROPORTIONAL

There is an important distinction to be made between the number of myoelectric levels an amputee can generate from a "cold start" and the continuous gradation of force and myosignal he can achieve with feedback. Systems requiring a cold start seem to be good for only two or three levels since they require an amputee to achieve a desired force output instantaneously with no feedback until after the fact. With a proportional system, the amputee can use visual or audio feedback to achieve good control. With the Boston Elbow it is obvious to even the casual observer that many more than three levels are easily achieved. Most of our amputees can vary the speed of their prosthesis at will over a wide range.

### Recent Electrical Improvements

Differential amplifiers moved to muscle sites
System gain increased from 47,000 to 110,000
Common mode rejection increased to 86-90 dB
More filtering on Op-Amp supply busses

### Recent Mechanical Improvements

Elbow axis to end of frame 21 cm (8.3 in)
(Allows room for electric wrist and hand)
System made completely modular
Shear pins protect mechanism in a fall

## COMBINING HIGH AND LOW SPEED CONTROL

Those experienced in the use of myoelectric signals know that the typical electromyogram is a noisy signal indeed. To be useful, it requires some sort of filtering. Unfortunately, the filter that removes "noise tremor" from the motion of a prosthesis at low speed introduces an unacceptable delay when gross high speed motions are desired.

We have solved the noise problem with a two-level filter. For signals less than 20% of maximum, the time constant is 880 msec (220 msec in the most recent design), while for greater signals the time constant is 18 msec. The circuit shown in Figure 1 gives excellent low speed control with acceptably fast response to large signals.

## HOW NATURAL IS THE CONTROL

While any pair of independently controlable muscles might be used to control an elbow, the 40 patients fitted to date use the most distal agonist-antagonist pair remaining. Thirty-three patients use biceps-triceps, two use anterior and posterior fibers of the deltoid and four use pectoralis major and the muscles on the posterior aspect of the scapula. All patients have been equipped with Bowden cable harnesses using forward motion of the shoulders or the humeral portion of the arm for actuating the terminal device, and all patients have been able to control the terminal device independently while moving the elbow.

There is some argument as to whether the Boston Elbow's velocity feedback is a "natural" way to operate the servo loop; however, several factors make this scheme more natural than it might seem. The first is the simple observation that rapid flexion or extension in the normal subject requires more effort than slow. The second factor has to do with the nature of a "real" versus a hypothetical servo system. With the elbow, the "real" servo is compliant. Thus more muscle signal is required for a given speed if the weight to be lifted increases. This is a "natural" way to handle changes in load. In addition, an amputee can sense an increase in load by a change in the load on the straps and socket.

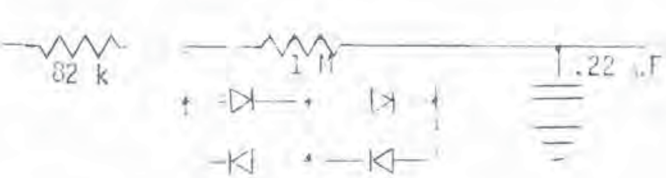


*Fig. 1 Two Level Nonlinear Filter*

### Engineering Specifications

Proportional pulse width modulation control
450 mA-hr battery will last one year
One 15-minute charge gives 350 flexions
Flexion time varies from 1 to 15 seconds
Actively lifts 2.7 kg at 30 cm (6 lb at 1 ft)
Passively holds 23 kg at 3 cm (50 lb at 1 ft)
Weighs 1.08 kg (2 lb, 6 oz) with battery
Self-locking no-back clutch saves power

## PROPORTIONAL CONTROL AFFECTS DESIGN

One reason why many myoelectric prostheses use essentially on-off control is that proportional control adds complexity. It is no longer possible to optimize the combination of battery, motor and gear train to operate at one speed. Pulse width modulation must be used to achieve variable speed control. If speed rather than motor current is to be controlled, a servo loop with a feedback tachometer must be introduced. Finally, the myoelectric system must be capable of differentiating many levels of myoactivity. This in turn implies that the preamplifiers need good common mode rejection and high gain and a good signal-to-noise ratio in situ.

Secondary considerations are that the drive system must combine high torque at low speed with reasonable maximum speed, that an efficient gear train is required and that a reverse locking clutch becomes more of a necessity than a luxury. The Boston Elbow combines all of these advanced technological features into one device that has now proven itself.

## ACKNOWLEDGMENT


Early research on the Boston Elbow was done at Harvard Medical School, the Massachusetts General Hospital and the Massachusetts Institute of Technology. Development has been done at the Research Center of Liberty Mutual Insurance Company with funding provided by the Company.


### Of Interest to the Amputee or Prosthetist

Upper arm socket uses standard technology
Separate electrode assembly easily installed
Drive mechanism sealed into Elbow housing
$30^{\circ}$ free swing or direct drive selectable
Adjustable length forearm holds battery
Terminal device uses below-elbow arrangement
Supplied with fast and slow chargers
Available to any qualified prosthetist
Modular construction for easy maintenance

NOTES

# FATIGUE I

## Session 4A

# THE ELEVATED ARM; MYOELECTRIC AMPLITUDE AND SPECTRAL CHANGES IN SOME SHOULDER MUSCLES

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## INTRODUCTION

Manual work performed at or above acromion height and demanding elevated arms, was found to be a promoting factor in non-inflammatory rheumatic shoulder and neck disorders. As a guide to ergonomic evaluations and vocational electromyography, this laboratory investigation was set up to determine which muscles are likely to develop electromyographic signs of muscle-fatiguing processes and how fast the signs occur in elevated arm positions.

## EXPERIMENTS

The investigation was set up with seven healthy females between 20 and 29 years of age. Two different elevated arm positions were examined for each subject, one with the arm held at 90 degrees forward flexion, the other with 90 degrees of abduction at the right shoulder joint. In both positions the elbow joint was kept at a right angle, the forearm semipronated; in the forward flexion parallell, and in the abduction position perpendicular, to the trunk. Separate from the experiments, the maximal voluntary contraction (MVC) of shoulder flexion and abduction in the two positions was determined for the subjects. In the experiments the subjects were asked to maintain the elevated arm position for as long as they could. The myoelectric activity was FM-tape recorded by bipolar surface electrodes from the descending part of the trapezius muscle, the infraspinate muscle, the middle and anterior part of the deltoid muscle, and from the common belly of the biceps brachial muscle. From the supraspinate muscle, the myoelectric activity was recorded by bipolar wire electrodes.

## DATA ANALYSIS

Electromyographic signs of fatiguing processes in a muscle can be determined by changes either in the amplitude or in the frequency domain. In a sustained contraction, ultimately leading to muscular fatigue, an increase of the EMG level is seen and of low frequency components in the myoelectric spectral content (Kadefors et al, 1968).

Off-line tape analysis was done both in the amplitude and in the frequency domain. The EMG-amplitude signal levels were determined as root-mean square (RMS) values for chosen sample sizes (5 sec.). The amplitude probability distribution function (APDF) was determined via the probability density function of the RMS-detected signal (Ericson and Hagberg, 1978). The mean power frequency (MPF) was calculated from the power spectra (Kwatny et al, 1970) obtained by FFT-analysis of one-second samples every eighth second. The RMS versus time regression (exponential curve) was estimated for the first 30, 60, 90 .... to 300 seconds. The MPF versus time regression (exponential) was estimated for the first five minutes. The significance of the slope estimators (increase in RMS, decrease in MPF) were tested by the t-test ($p < 0.01$). The first minute and the last minute of static myoelectric levels (the amplitude levels at p 0.10 in the APDF) were estimated for the different muscles in the experiments. Difference in torque and endurance between the two positions was tested by the t-test of correlated data ($p < 0.05$).

## RESULTS

The earliest significant increase of RMS was found in the supraspinate muscle in the forward flexion position and occurred for all subjects within 1.5 minutes (mean: 0.9 minutes). In the abduction position this amplitude change was found for all subjects in the supraspinate muscle within 3.5 minutes (mean: 1.8 minutes). For the trapezius muscle the significant increase of RMS was found within 2.5 minutes in both positions. For the other muscles significant increase of RMS was not found within five minutes for all subjects (see figure 1). A total of 11 muscles showed significant decrease of MPF in the forward flexion position, for the abduction position 20 muscles during the first five minutes. Both significant decrease of MPF and increase of RMS was found in the forward flexion position for 11 muscles, in the abduction position for 18 muscles (figure 1).

The mean "static" µV-levels for the first and last minutes of the experiments were high for the trapezius and the supraspinate muscle (figure 2). The mean torque produced by maximal voluntary contraction in the forward flexion was 48.3 Nm and in the abduction 45.2 Nm (significant difference). The mean load level (torque produced by the weight of the arm) for forward flexion was 11.9 per cent of MVC and for the abduction position 12.6 per cent of MVC (non-significant difference). The mean endurance time for the forward flexion position was 21 minutes and for the abduction position 18 minutes (non-significant difference).

## DISCUSSION

For the elevated arm positions, electromyographic signs of muscular fatiguing processes developed rapidly in the supraspinate and the upper part of the trapezius muscle. The findings for the supraspinate muscles are in accordance with vocational studies of elevated arm positions (Herberts and Kadefors, 1976) and with functional anatomical considerations such as the muscle being an abductor and stabilizor in the humero-scapular joint. Both significant RMS increase and MPF decrease are strong evidence of muscular fatigue development. This occurred in five of the seven subjects for the supraspinate muscle in abduction during the first five minutes.

The upper fibers of the trapezius prevents a downward rotation of the scapulae and this stabilizing function in the elevated arm is high enough to result in quick muscular fatiguing processes. The frequently observed neck complaints in workers with overhead work, may be a result of exertion of the trapezius muscle.

The mean "static" EMG level during the first and last minute of the experiments (figure 2) seems similar to the incident of muscular fatigue signs (figure 1). The load levels of the shoulder muscles are high even if only the arm is elevated, giving for all subjects a static load level above 10 per cent of MVC. Furthermore in assembly work demanding elevated arms, tools are often elevated by the arm.

The mean endurance time for the two positions is in agreement with the endurance time for the elbow flexors at corresponding contraction levels (Björkstén and Jonsson, 1978).

## CONCLUSIONS

In the elevated arm position, signs of muscular fatigue development occurs rapidly, in the supraspinate muscle and the trapezius muscle within a few minutes. The load level of the shoulder muscles, in positions demanding elevated arms, is more than 10 per cent of MVC.

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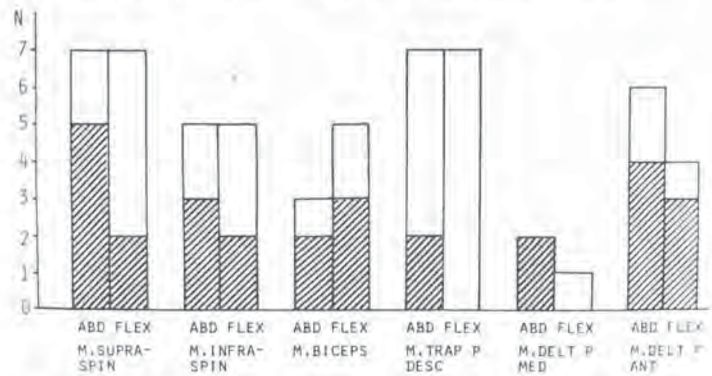


Figure 1 The number of subjects with significant increase of RMS or significant decrease of MPF (blank area) during the first five minutes, both RMS increase and MPF decrease is marked with shaded area. The forward flexion position is marked Flex and the abduction position Abd for the different muscles.

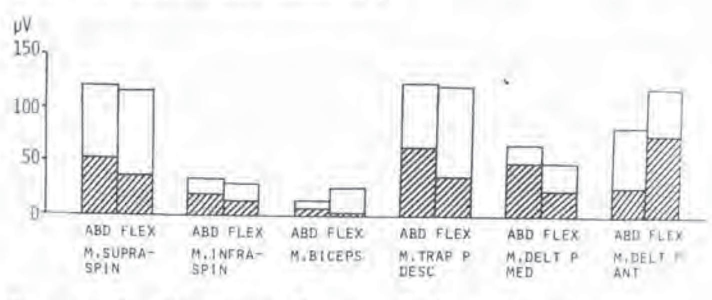


Figure 2 The mean "static" (p .10 for the APDF) µV level for seven subjects during the first (shaded area) and the last (blank area) minute of the experiments.

LOCALIZED MUSCLE FATIGUE IN SHOULDER MUSCLES: A PRELIMINARY STUDY EMPLOYING THE SPECTRAL MOMENT ANALYZER.

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## INTRODUCTION

Localized muscle fatigue - motor impairment and pain confined to muscle - can be quantified using electromyographic methods (Kadefors et al., 1968, Chaffin, 1973). The spectral changes concomitant with the sensation of fatigue are attributable to a progressive decrease in action potential conduction velocity along the muscle fibers (Broman, 1977). Through the mathematical model for myoelectric signal generation and propagation presented by Lindström (1970), a so-called fatigue index (Lindström et al., 1977) could be defined. This index makes possible intermuscular and interindividual comparison of muscle fatigue in different working situations. A new instrument, the Spectral Moment Analyzer (Broman, 1979, Broman & Kadefors, 1979) was developed for the particular purpose of quantifying localized muscle fatigue. The instrument was evaluated in this preliminary study of the fatigue of different muscles of the shoulder in different conformations of the arm.

Shoulder muscle fatigue in welding work has been studied in ship-yard environments (Kadefors et al., 1976), where it was found that localized muscle fatigue is common in overhead work. This laboratory study was undertaken, making use of the fatigue index method, to quantify muscle fatigue in different muscles of the shoulder region as a function of working posture. Such knowledge may form a basis for recommendations in specific job situations.

## METHOD AND MATERIAL

The study was carried out in an experimental situation where the subject was keeping an electrode holder (weight 2 kg) fixed in space in three different positions, according to Figure 1.

In each position different degrees of glenohumeral abduction was assumed as shown in Table 1.

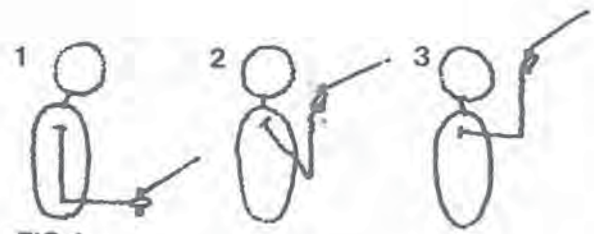

FIG.1

Table 1
Postures investigated

| | | | | | | | | |
|---|---|---|---|---|---|---|---|---|
| Glenohumeral flexion | $0^0$ | $0^0$ | $45^0$ | $45^0$ | $45^0$ | $90^0$ | $90^0$ | $90^0$ |
| Glenohumeral abduction | $0^0$ | $30^0$ | $0^0$ | $45^0$ | $90^0$ | $0^0$ | $45^0$ | $90^0$ |
| Notation | 1:1 | 1:2 | 2:1 | 2:2 | 2:3 | 3:1 | 3:2 | 3:3 |

Each posture was maintained for one minute. After each single loading, several minutes were allowed for rest. The load level was not considered heavy by the subjects.

EMG signals from the anterior and the medial portions of the deltoid, the supraspinatus, the infraspinatus, and the upper portion of the trapezius muscle were recorded by means of single fine wire electrodes inserted by means of a cannula. A common reference electrode was placed on the surface of the skin above the spinal processus at level T3. The signals were fed to a Medelec electromyograph, amplified and filtered 8 Hz-3.2 kHz. The signals were recorded on a Bell & Howell FM tape recorder at 15"/s tape speed.

Analysis of the recorded signals was performed on the Spectral Moment Analyzer (see Broman & Kadefors, 1979), with a low-pass frequency of 1 kHz and a time constant of 1 s. Only the mean frequency (synonymous with relative velocity) and the r.m.s.-channels were considered.

Five normal male subjects aged 20-35 participated in the study.

## RESULTS

For an illustration of the printout, see the parallel abstract at this conference (Broman & Kadefors, 1979)

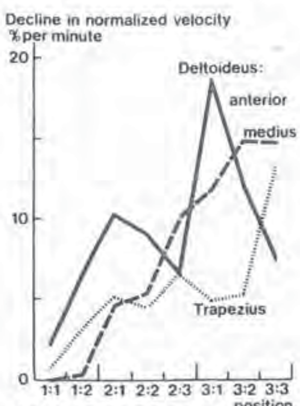


FIG. 2

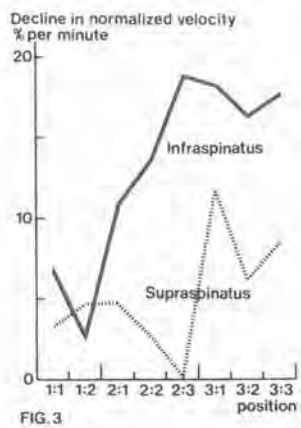


FIG. 3

The results are given in terms of change in relative velocity per minute. This quantity is arrived at by estimation of the regression line in each individual recording. Figure 2-3 summarize the results. For clarity, the dispersion has not been included in the figures; a typical standard error amounts to 2 %/min.

## DISCUSSION

The present preliminary study does not include a thorough statistical analysis. This will be undertaken on the basis of a larger material. Some observations of interest follow.

The anterior portion of the deltoid muscle is the main flexor of the glenohumeral joint. Accordingly, there is an increasing activity in this muscle when the arm is elevated.

With increasing abduction the medial portion of the deltoid takes over, which is illustrated for positions (2:1-2:3) and (3:1-3:3).

If the arm is elevated to shoulder level there is a pronounced muscle fatigue occurring in both portions of the deltoid muscle even at this moderate load.

The fatigue phenomenon in the supraspinatus is less pronounced in this study, at this load and short observation time. The muscle fatigue will appear when the arm is elevated to shoulder level. It has, however, been shown that the supraspinatus muscle is very much affected in more prolonged, static and dynamic overhead working situations.

The striking fatigue in the infraspinatus muscle reflects the important function of this muscle as a stabilizer of the glenohumral joint when the arm is flexed and abducted.

In the arm positions studied there is only a slight muscle fatigue in the trapezius muscle which is not surprising since this muscle is most active when the arm and shoulder is more extremely elevated.

Localized muscle fatigue is closely linked to the subjective perception of work load. The totally perceived muscular pain may be related to contributions from all the individual sources or be set by one or two dominating sources exclusively. Figure 4, finally, shows the result of a linear addition of fatigue contributions from different sources (which does not include all conceivable contributors in the shoulder region) in overhead work. This would indicate that a moderate glenohumeral abduction is advantageous in overhead work similar to the experimental situations studied.

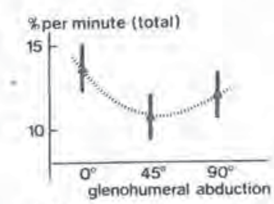


FIG. 4

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# THE FATIGUE AND RECOVERY OF THE HUMAN DIAPHRAGM

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## INTRODUCTION

Studies on diaphragmatic fatigue in man using EMG frequency analysis have shown that the rate of decay in the high frequency/low frequency(H/L) ratio with time is inversely related to the duty cycle for any given muscle tension (1). The rate of recovery in the H/L that takes place during the interval between two successive contractions is thought to be one of the main determinants for the overall rate of decay in the H/L. The present experiment was undertaken to define the recovery time constant (Tr) of the H/L of the diaphragm during inspiratory resistive breathing.

## THEORY

Lindstrom (4) has described the time course of the decrease in conduction velocity of myopotentials during static continuous contractions as a function of muscle load and has developed the so-called "fatigue index model". The model has recently been extended to periodic muscular contractions to include the analysis of the recovery time constant of conduction velocity proposed by Broman(2). (Lindstrom, personal communication).

The extended fatigue index model proposes that the propagation velocity of the action potentials during periodic fatiguing contractions follows on the average an exponentially decreasing course with a final stable value, a plateau. The final velocity (Vf) can be approximated by the following equation:

$$Vf = Vo/((1 + CTr/(1 - C)\ Tf)) \qquad (1)$$

where Vo is the velocity at the beginning of the test, C the duty cycle and Tr and Tf the recovery and fatigue time constants, respectively.

We then define Tr/Tf = K, and solving equation (1) for K yields the following expression:

$$K = (Vo/Vf - 1) \cdot (1/C - 1) \qquad (2)$$

Substituting in equation (2) the conduction velocity for the H/L ratio (expressed as a fraction of the initial value), we find:

$$K = \frac{T_R}{T_F} = \frac{1}{\frac{H}{Lf}} - 1 \cdot \frac{1}{C} - 1 \qquad (3)$$

where H/Lf is the value of H/L at the plateau.

Knowing the values of H/Lf and Tf, their values obtained experimentally from periodic and continuous fatiguing contractions respectively, one can calculate Tr by solving eq. 3.

## METHODS

Three normal male subjects were studied. On each run they inspired to generate transdiaphragmatic pressures (Pdi) ranging between 15 and 90% of the Pdi maximum. Each Pdi was sustained at duty cycles ($T_I/T_{TOT}$) ranging between 0.25 to 1. In all runs, tidal volume, Pdi and $T_I/T_{TOT}$ were monitored by the subject from a storage oscilloscope and kept constant for a maximum of 10 min or until exhaustion. Diaphragmatic EMG was recorded with an oesophageal electrode. The signal was band pass filtered between 150 and 350 Hz and between 20 and 40 Hz for High (H) and Low (L) frequency power content respectively, and recorded on paper from which the H/L ratio was computed and normalized with respect to the value obtained at the beginning of each test.

## RESULTS

The fatigue time constant (Tf) of the diaphragm was calculated from the slope of the H/L-time ratio for the continuous and periodic contractions. It is shown in Figure 1 as a function of the product of Pdi and $T_I/T_{TOT}$. The two parameters shows an inverse power relationship. Tf ranges between 12 sec and 16 min.

In addition, in runs with a duty cycle less than one, the level of the plateau of the H/L ratio was measured. In all runs where a stable value of H/L was found, K was calculated according to eq. 3 The results are presented in Fig. 2. K was found to be linearly related to Pdi. The regression line intersects the abscissa at 15% of the Pdi max which represents the critical pressure below which no fatigue can occur because the Tf approaches infinity and Tr approaches zero.

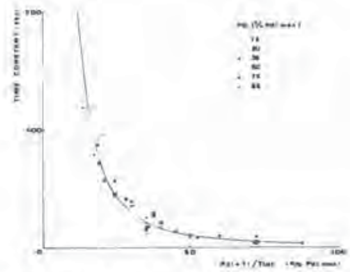

Fig. 1:Effect of the product of transdiaphragmatic pressure (Pdi) and the fatigue time constant (Tf) in 3 subjects. The Pdi · $T_I/T_{TOT}$ product is expressed in percent of the Pdi max of the subjects. Each point represents the result of one test. Different symbols refer to different Pdi.

From Fig. 1 and 2, the value of Tr was calculated for any Pdi greater than 15% of the Pdi max. Fig. 3 shows the relationship between Tr and Pdi. Also shown, is the regression line of Tf for continuous contractions extracted from Fig. 1. Tr is found to increase very rapidly at

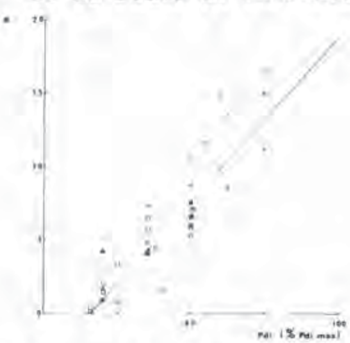

Fig. 2:Relationship between K (recovery time constant/fatigue time constant) and Pdi (expressed in percent of the Pdi max of the subjects). Different symbols represent different subjects. Linear regression equation: K = 0.02 Pdi - 0.34, r = 0.85.

low Pdi to reach a maximum value at about 25% of Pdi max and from then on to decrease almost linearly with increasing Pdi.

## DISCUSSION

It has been reported earlier that the Tf of the biceps is inversely related to tension for any given duty cycle and to duty cycle for any given tension (Lindstrom, personal communication). From the present data, it appears that the Tf of the diaphragm is uniquely related to the Pdi·Ti/Ttot product as shown in Fig. 1 which includes both continuous and periodic contractions of the diaphragm under various Pdi and $T_I/T_{TOT}$ values. This implies that Tf is affected in the same proportion by either Pdi or Ti/Ttot. It should be mentioned that all tests were performed at a constant flow rate (Vt/Ti=0.25 L/sec)

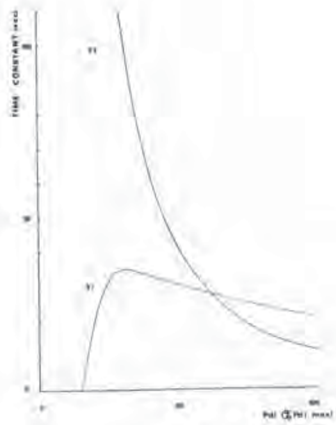

Fig. 3:Relationship between fatigue (Tf) and recovery (Tr) time constants and Pdi (expressed in percent of the Pdi max of the subjects).

so that the above result might apply only for the present experimental conditions.

Both the Tf and Tr for the diaphragm appear to be smaller than the values calculated from the conduction velocity in the biceps (2,4). Whether this observation represents a real biological difference between the two types of muscle, or is due to the different parameters used to evaluate the time course of fatigue development is not clear at the moment. Nevertheless, the function of the Tf vs Pdi curve is similar to that of conduction velocity in the biceps. Tr is clearly less dependent on Pdi than Tf. This finding is in agreement with other reports on recovery data calculated from the conduction velocity. Moreover, Tr and Tf intersect (K = 1) at a relative load of 60% of the Pdi maximum, which fits the combined data of Lindstrom (4) and Broman (2).

The observation that Tr decreases with increasing load beyond 25% of the Pdi max is in accordance with the data of Funderburk et al(3) who found that the relative rate of recovery of endurance time from statically induced fatigue is directly related to the load.


## ACKNOWLEDGMENT

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# FATIGUE AND STRENGTH OF THE TRICEPS SURAE AFTER SURGICAL AND NON-SURGICAL TREATMENT OF ACHILLES TENDON RUPTURE

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## Introduction

The achilles tendon is an enormously strong structure - even in the human body - and still it is one of the tendons which most often rupture as a result of indirect trauma. The treatment of subcutaneous ruptures of the tendon has traditionally been operative since the operation is simple and the results have been satisfactory. On the other hand, good results have also been obtained when the rupture has been treated non-surgically, i.e. with a plaster cast for eight weeks. When reporting the results of these treatment modalities, the frequency of healing of the rupture and the patients gross functional ability are usually analyzed. This is sufficient in many cases, but in order to compare the two methods further, a more detailed functional analysis of the triceps surae was considered helpful.

## Material

The material consisted of 103 consecutive patients from an ongoing prospective study in which subcutaneous achilles tendon ruptures were treated surgically and non-surgically every other day. The minimum follow-up period was one year. The clinical results of both treatment methods were good and comparable. All patients were included in the study of plantar flexion strength and twelve patients from each treatment group were randomly selected for quantitative myoelectric studies.

## Methods

The plantar flexion strength was determined for both legs in the supine position using a Cybex II dynamometer. Studies were made with the ankle in $15^{\circ}$ of dorsiflexion, in neutral position, and in $15^{\circ}$ and $25^{\circ}$ of plantar flexion. In each position the maximum voluntary plantar flexion torque was determined. Dynamic contractions were studied at constant velocity of 30, 90 and 180 degrees per second, and the peak torques were measured. The myoelectric activity of the soleus muscle and the medial and lateral parts of the gastrocnemius muscle was picked up by means of unipolar wire electrodes. Recordings were made in the neutral position and different angles of flexion. The signals were stored on magnetic tape together with information on the generated moment for subsequent analysis. A static contraction at 75 per cent of the maximum torque level on the injured side was performed during 30 s with the ankle in neutral position. For analysis, the myoelectric signals were played back from the magnetic tape, filtered in low pass filters limit frequency 800 Hz, fed to a digital computer and digitized at a conversion rate of 2048 $s^{-1}$. Average rms-values of the signal amplitude during the recording period were estimated. Analysis of localized muscle fatigue was performed using the fatigue index method (Lindström et al, 1977). The method involves power spectrum analysis and yields relative values of action potential propagation velocity along the muscle fibers, decreasing velocity being associated with increasing fatigue.

## Results

A decrease in plantar flexion torque compared to the healthy side was found in all subjects. The greatest difference in moment was recorded when the ankle was in $25^{\circ}$ plantar flexion. No statistically significant difference (5 %) was found between subjects treated surgically and non-surgically.

During the fatigue experiment, the conduction velocity decreased in all muscles. The decrease was faster in gastrocnemius than in soleus. Also, the decrease was faster on the injured side than on the uninjured for all muscles. There were no statistically significant differences in conduction velocity decrease between the treatment groups.

## Discussion

In this investigation an attempt was made to study the function of the muscles attached to the achilles tendon. No functional differences between surgical and non-surgical treatment of tendon rupture could be found. The muscles on the injured side were weaker and more fatigable than those on the healthy side, however. The fact that the main loss in plantar flexion force occured with the ankle in plantar flexion, suggests that it is the function of mainly the gastrocnemius muscle which is impaired. Although it is impossible, at present, to performe a detailed in vivo analysis of the treatment effect on the tendon itself, the findings support the clinical impression that the treatment methods are comparable with respect to residual function.

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NOTES

# CLINICAL I

## Session 4B

# A LONG-TERM ELECTROMYOGRAPHICAL EVALUATION OF INSPIRATORY FUNCTION IN QUADRIPLEGIC PATIENTS


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## INTRODUCTION

The muscles involved in inspiration can be classified as primary, first-order accessory, and second-order accessory. The primary inspiratory muscles, the diaphragm (Diaph) and the intercostals (IC), are defined as having their principal points of attachment in the rib cage. The first-order accessory muscles, such as the pectoralis major (Pect), are those muscles which are attached only in part in the rib cage. The second-order accessory muscles, such as the sternocleidomastoid (ScM) and the upper fibres of the trapezius (UT), are those muscles which have no point of attachment in the rib cage and which can be either paradoxically or synergistically involved in inspiration.

The study of dynamic inspiratory function in quadriplegia is important since respiratory complications are a major cause of mortality in both the acute and chronic stages (Silver & Gibbon, 1968). This can be attributed to paralysis of some respiratory muscles which results in a marked decrease in vital capacity, rib cage mobility, ability to cough, and a marked reduction in the maximum inspiratory and expiratory forces as measured by mouth pressure (Fugl-Meyer, 1971).

At the Rehabilitation Institute of Montreal a long-term research project has been established to: 1) evaluate inspiratory muscle function in quadriplegic patients; and 2) improve respiratory function through resistive inspiratory training with these patients.

This paper is concerned specifically with: muscle patterning during non-resistive and resistive inspiration, before and after training; and the strength and endurance of the diaphragm, before and after training, as determined by localized muscle fatigue.

## METHODS

Twelve traumatic quadriplegics whose lesions were complete at levels ranging from C4 to T1 have participated in the study. All were at least one year post-injury to ensure relative stabilization of physiological and motor function. Before training it was established by laboratory tests and physical examination that the subjects were free from respiratory disease and were in good health.

Using surface electrodes in a bipolar configuration electromyographic (EMG) signals of the UT,ScM,Pect,IC (first and second intercostal spaces for females, second and third for males), and Diaph (sixth and seventh intercostal spaces next to the costal margin) were recorded unilaterally for each subject during a variety of respiratory manoeuvres. From this it was determined that the Diaph and the ScM were the two principal muscles involved in non-resistive inspiration. These two muscles were chosen, therefore, for analysis and monitoring during the first eight-week inspiratory training phase. Electrodes were placed on the right side of the subject to minimize the cardiac signal which is unavoidably picked up by surface electrodes placed in the thoracic region.

Prior to the first training phase each subject was tested at three different inspiratory resistances for ten minutes each to determine which would be selected for training. The resistance chosen for each subject was the lowest resistance at which: 1) the subject demonstrated localized neuromuscular fatigue as determined by the analysis of the EMG frequency spectrum(Kadefors, Kaiser, and Petersén, 1968); 2) the expired $CO_2$ stayed within normal limits; and 3) the subject could breathe without distress.

The first resistive inspiratory training was carried out over an eight-week period. Subjects inspired against the same resistance for two fifteen-minute periods daily, six days per week. Twice weekly this training was done in the laboratory. During these laboratory sessions the EMG signals of the Diaph and the ScM were presented to the subject on an oscilloscope and were recorded on tape for subsequent analyses.

After training, the pre-training test with the same three inspiratory resistances was repeated and the EMG signals of the five muscles originally monitored were again analyzed for the presence of fatigue. Based on this analysis a decision was made concerning the next resistance at which the subject would train. After a further eight weeks training at the chosen resistance, laboratory testing was repeated. This program of training, testing, and analyzing continues with eight subjects. Four subjects opted to terminate at some point after the first training period.

## RESULTS

As aforementioned, in pre-training non-resistive inspiration the Diaph and ScM were the two principal muscles involved. However, for one subject only, the UT was involved also in inspiration. For all subjects, in pre-training resistive breathing there was a definite increase in the amplitude of the EMG signal for the Diaph and the ScM and there was a tendency for the duration of the electrophysiological response to increase in both muscles.

In the first post-training EMG evaluation it was observed that the electrophysiological response of the ScM (plus UT in the one subject) had decreased to near zero during non-resistive and resistive inspiration and that there was a possible decrease in amplitude within the Diaph. Subsequent post-training EMG evaluations at increased resistances have shown similar results.

Seven subjects have progressed markedly beyond their first training resistance and have continued to increased resistances after an average of 12.7 weeks training. The mean percentage increases between the first training resistance and subsequent resistances are 56.7, 98.7, 146.9, and 198.2%.

## DISCUSSION

Studies have shown that in skeletal muscle tissue fatigue can be detected by observing changes in the frequency spectrum of the myoelectric potentials (Kadefors et al., 1968; Lindstrom, Magnusson, and Petersén, 1974). These studies showed that when muscle sustains a fatiguing load the amplitude of the high frequency component (H) of the EMG decreases with time and the amplitude of the low frequency component (L) gradually increases resulting in a decrease in a H/L ratio. Gross, Grassino, Scott, Ladd, and Macklem (1978) studied the EMG pattern of the diaphragm in quadriplegic patients during inspiratory loading and also found a decrease in the H/L ratio over time. The drop was interpreted as indicating the presence of localized fatigue. This analysis served as the basis for choosing the lowest level of resistance at which fatigue occurred.

The EMG data from this study clearly show a change in the muscle patterning during both non-resistive and resistive inspiration. All subjects relied heavily on the ScM before training which resulted in paradoxial breathing. This muscle was not essentially employed during post-training non-resistive and resistive inspiration, thus inspiration was performed by the Diaph only; there was also a significant progression to increased resistances. These results could be interpreted as indicating an improvement in the strength/dynamic response and endurance of the diaphragm.

This research is supported by the Medical Research Council of Canada grant #MA-5855.

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# STUDIES OF THE POSTURAL MUSCLE ACTIVITIES AND THE OXYGEN CONSUMPTION IN JUVENILE AND ADULT PATIENTS WITH CEREBRAL PALSY.

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## ABSTRACT

A study of electromyography and oxygen consumption were performed in patients with cerebral palsy. Considerable amount of research has been reported on the well correlation between the postrual muscle activity and oxygen consumption. It would be interesting in patients with cerebral palsy to know further latent function potentials of the lower extremities.

Materials were consisted of 60 cases with cerebral palsy whose lower extremities motor age (L.M.A.) were over 12 months, and of 25 cases without cerebral palsy as a control, in total 72 cases. The electromyographic and oxygen consumption studies were done at easy standing and supine position.

The results were obtained as follows:

1) According to the electrical response using the surface electrodes, the amplitudes of lower than 100 micro-volts were shown on the postural muscles at both easy standing and supine position in control group.

On the other hand, majority of cases showed the electrical responses of higher than 100 micro-volts, in cerebral palsy group.

2) The marks were made in proportion to the electrical responses. The mean values of these total marks of electrical responses in each person were resulted in 1.6 at supine, 4.3 at easy standing in control group, on the other hand, these were 5.1 at supine, 12.7 at easy standing in cerebral palsy group. Available evidence indicates that discharges from the postural muscles were more active with cerebral palsy group than with normal.

3) As to the mean values of oxygen consumption with position changes from supine to standing position, these were increased by 11.83% in control group, and 32.22% in cerebral palsy group.

4) It was revealed to have a tendency to decrease the values of marks in the cases with cerebral palsy who were treated operatively and gained some functional improvement.

NOTES

ELECTRICAL STIMULATION FOR SCOLIOTIC PATIENTS

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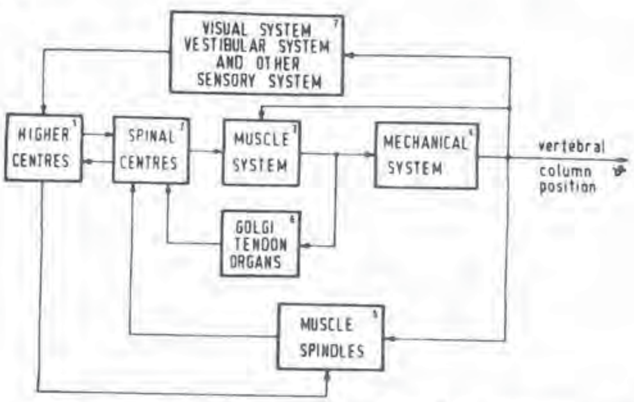
## INTRODUCTION

The basic idea of using electrical stimulation to treat scoliosis was put forward many years ago /1/ . These studies are based on the analysis of the posture control system and the statement that scoliosis is due to an improper functioning of the posture control system (Fig. 1).




The analysis of the system and the control of various therapies have been carried on by means of a mathematical model of the system implemented on a digital computer and by means of EMG and thermographic analysis.

## METHODS

Various techniques can be used to stimulate patient's muscles both with surface and implanted electrodes, using different wave forms (pulses, train of pulses, sinusoids) at various frequencies. In our experience, after having abandoned the trains of pulses, we have found it better to use sinusoidal currents of 2,5 KHz modulated by a 50Hz square wave as adopted by Kots to increase muscle force (and power) of athletes /2/. Every application consists of ten second stimulation followed by a rest interval of 50 seconds, repeated for 20 minutes. Kots obtained with 15 applications an increase of 30% of force using tetanic contractions in isometric conditions. We have obtained analogous results repeating the same experiment without reaching the tetanic contraction and not blocking the movements that are the conditions used to stimulate the muscles of the back of the scoliotic patient. Our methodology is :

- -EMG analysis of the spine muscles.
- -Surface stimulation of the part where EMG activity is lower. Each cycle of stimulation is characterized by daily applications of 20 minutes for 15 days. This cycle is repeated every three months.
- -Control of theraphy with X-ray, EMG and thermographic analysis.

## RESULTS

We report now the results obtained with 24 patients treated only with electrical stimulation from a minimum of 18 months to a maximum of 30 months (with an average of 22 months). These patients represent a total of 52 curves. According to the classification adopted by Bobechko a decrease of $5^{O}$ or more in the curve is rated "better"while $5^{O}$ or more increase is "worse" and a change of less than $5^{O}$ is "unchanged"

| | |
|---|---|
| $\Delta \leq -5^{O}$ | BETTER (B) |
| $-4^{O} \leq \Delta \leq 4^{O}$ | UNCHANGED (U) |
| $\Delta \geq 5^{O}$ | WORSE (W) |

The treated group shows 21% of the curves better, 73% of the curves unchanged, 6% worse. A total of 94% of the treated curves have successfully been arrested or partially corrected. If we compare the results here achieved with those achieved with the Milwaukee brace, we can see that the advantage lies very much with electrical stimulation. Namely the successfull results (%Better + % Unchanged) with Milwaukee brace reach at most 74%.

## RESULTS OF ELECTROSTIMULATION

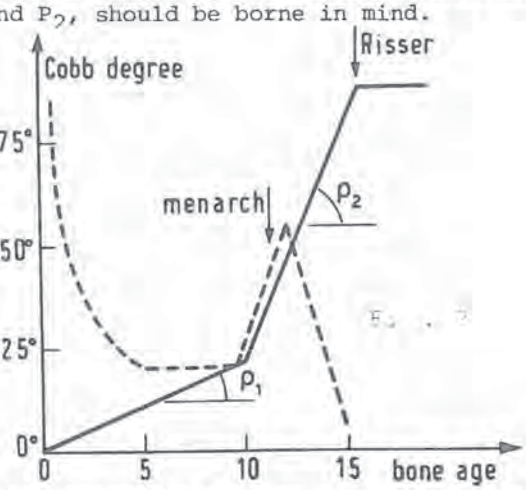
Stimulations for periods from 18 - 30 months
For 24 patients, average age: 13-15 years,
risser 0-3 + for a total of: 52 curves.

BETTER: 11 CURVES = 21% average improvement: $-8^{O}$
UNCHANGED:38CURVES= 73% av. stabilization: $-0.2^{O}$
WORSE : 3 CURVES = 6% av. worsening: $+5.3^{O}$

If we consider also patients under treatment for less than 18 months we have the following results:

| | |
|---|---|
| 36 PATIENTS | B = 40% |
| | U = 53% |
| 73 CURVES | W = 7% |

The results agree with the results obtained by other researchers. We have observed that electrical stimulation causes a decrease of the curve with the first cycles and then tends to mantain the results achieved. In order to follow the evolution of each patient and provide up to date statistics, a program for automatic files management was implemented. The curve which represents the progression of scoliosis, given by Duval Beaupere/3/, the typical form of which is shown in fig.2 by two straight lines of slopes $P_1$ and $P_2$, should be borne in mind.




The point which joins the two straight lines corresponds to the beginning of puberty. The bone growth stops with completion of the covering of the iliac crests (Risser's test). It is to be noted that in 60% of cases $P_2 \cong 5P_1$. After an orthopedic treatment the scoliosis progression mantains the same slope. The graphs are produced by the program for automatic file management; all data concerning the patient, controls and theraphies are represented on the top of the graph, and the curves (expressed in Cobb degree) as a function of the time are shown below. The letter S indicates a cycle of stimulation.

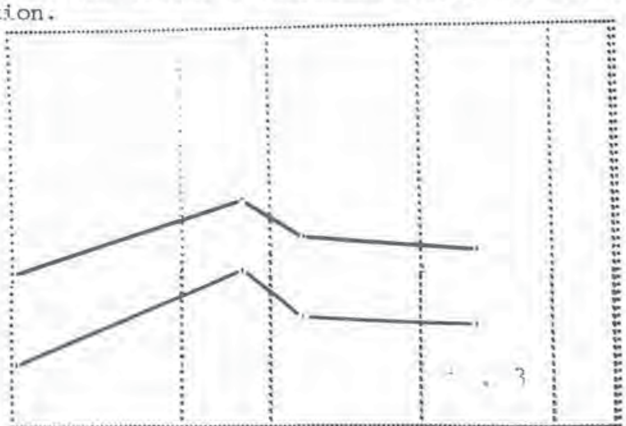


Fig.3 shows the case of a joung girl. The first X-ray control (at the age of twelve) indicated a right dorsal scoliosis (8 Cobb) and left lumbar (20 Cobb). After a year it was increased to 20° and 29°. After the first stimulation cycle, scoliosis decreased to 14° and 24° and after a second cycle it was at 13° and 22°. The girl is going on with stimulation.

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# SCOLIOSIS PROGNOSIS BY MEANS OF THERMOGRAPHIC ANALYSIS

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## INTRODUCTION

One of the most important needs in the study of scoliosis is,clearly, the early diagnosis and prognosis of the disease. Namely, it is extremely important to know between two scoliotic curves which look the same, the one which will develop becoming an evolutive scoliosis and the one which will remain unaltered or will manifest itself only in postural attitudes. Nowadays it is practically impossible to do a reliable prognosis in a short observation time. X-rays can not be repeated frequently and certainly they do not give a prediction of the future evolution of the curve even if they give a step by step picture of the development of the disease. Thermography can successfully be employed to this end. It starts from the consideration that in normal people the thermogram of the back is symmetrical with the respect to the vertebral column. In the case of scoliosis there is an asymmetry certainly due in the great majority of the cases to a different activity of the paravertebral muscular masses at the same level. In revealing such muscular asymmetries the technique enables us to spot the disease in its early stages, well before the vertebral column deviates from its normal straight position. The thermogram of the back is extremely useful also in prognosis of the disease. From the level and extent of thermographic asymmetry one can assess the gravity of the muscular unbalance and make a prognosis with the aid of a mathematical model of the spine implemented on a digital computer.

## MATHEMATICAL MODEL

The posture control system of the spine can be schematized as shown in fig.1. The aim of this system, which is sketched here in its essential blocks, is to permit the various movements of the vertebral column and to maintain the upright standing normal position with the column straight. The part of the above mentioned system which has been simulated is the "mechanical part" represented in fig. 1. Inputs to this system are: "muscular forces" and "external disturbances" (due particularly to theraphy, i.e. braces,casts, Harrington operation etc.). Outputs are the angles $\phi_i$ which represent the relative rotations in the frontalplane of each vertebra with respect to the inferior contiguous one.The program gives the final configuration of a spine subjected to a set of forces acting on it. The muscular masses with asymmetric behaviour are determined via thermographic and electromiographic examinations. The collected data can be utilized to simulate the state of the disease and its probable evolution.

## CLINICAL FINDINGS

What above stated is confirmed by the following experimental results: in the case of non scoliotic subjects, the thermogram of the back appears symmetric with respect to the spinal column, due to the fact that there is no asymmetry in the muscular activities of a subject standing in orthostatic posture. It is to be noted that such a result is confirmed also by E.M.G. measurements. In the case of scoliotic patients different thermograms may result according to the stage of evolution of the pathological form. a) Non Evolving scoliosis; b) Evolutive scoliosis; c) Stabilized scoliosis.

a) Non Evolving scoliosis: the thermogram is practically symmetric with respect to the spine. This means a corresponding balanced activity of muscular masses. The scoliosis is not worsening.
b) Evolutive scoliosis: the thermogram is asymmetric with respect to the deformed spine. The activity of muscular masses is not balanced, the deformity will develop and worsen in the near future; the degree of scoliosis is strictly dependent on the asymmetry magnitude fig. 2.
c) Stabilized scoliosis; the thermogram is almost symmetric with respect to the deformed spine. The subject is at the end of growth. The curvature is about 50° Cobb. (fig. 3). The results show a clear association between the degree of asymmetry with the scoliosis evolution. Therefore thermography can be used for the prognosis of desease and for early mass screening. (It is possible to submit the subject to repeated examinations with no damage in a very short time).

## AUTOMATIC ELABORATION OF THERMOGRAPHIC IMAGES

In order to quantify the asymmetries revealed by the thermocamera an automatic acquisition system is used. In this system the image is subdivided in a net of 128x128 points and to each point is associated a variable level of grey from 0 to 255, that is 256 levels are available which go from white to black. After having digitized and stored the images it is easy to elaborate the data by means of software. At the present time the available programs have been studied in order to permit in future a wide mass screening for which automatic procedures are needed. The available software may be divided in two main parts: the first one concerns the management of the acquisition, visualization and storage of the thermographic images; the second one permits the computation and the printing of the graphics and characteristic indices. The storage is done by means of the computer tape unit. The data stored are: the anagraphic data, anamnestic data, data concerning the present thermogram and the thermogram itself. The management of the files is very simple because the data are supplied in "conversational" by means of video terminal. At the present time the elaboration programs allow the visualization of the axis of the vertebral column superimposed on the thermographic image and the computation of the thermic asymmetry indices with respect to the axis itself. The thermographic image is subdivided in 128 lines (j) with 128 points each (i). $y_{ij}$ is the level of grey (quantized in 256 levels ) of the $i^{th}$ point on the $j^{th}$ line. $K_j$ is the point of the line j (with 0 $K_j$ 128) corresponding to the intersection of line j with the axis of the vertebral column and N is the number of points taken into account on the right and on the left of $K_j$ while computing the asymmetry index given by

$$A_j = \sum_{i=K_j+1}^{K_j+N} y_{ij} - \sum_{i=K_j-N}^{K_j-1} y_{ij}$$

For a given N the programs supply the graphic of $A_j$ for j varying from 0 to 128.

## CONCLUSIONS AND FUTURE DEVELOPMENTS

The method and the elaboration here reported provide the necessary basis for a correct prognosis of the scoliosis evaluation. This methodology can be applied as a routine for a large number of patients because it takes a reasonably short time for each of them and this allows a mass screening for children of school-age. The thermographic analysis will permit also the optimization of the therapeutic procedures controlling their efficacy for each subject.

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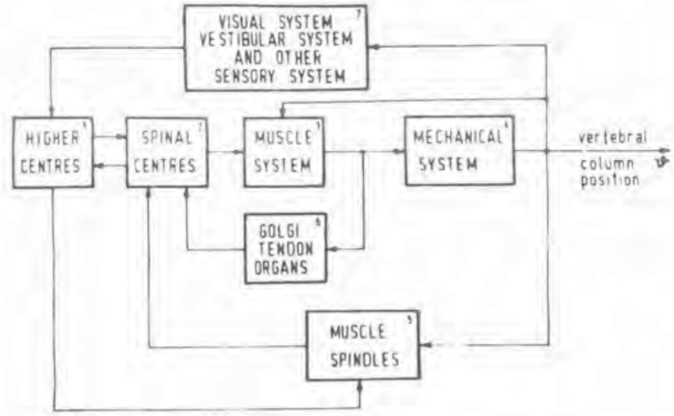


Fig. 1 - Mathematical Model of the System.

Fig. 2 - Thermogram of scoliotic patient.

Fig. 3 - Thermogram of scoliotic patient at the end of growth.

NOTES

# FATIGUE II

## Session 5A

A SPECTRAL MOMENT ANALYZER FOR QUANTIFICATION OF ELECTROMYOGRAMS

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## INTRODUCTION

Localized muscle fatigue is conveniently studied in the frequency domain. Kadefors et al (1968) showed that prolonged muscle contractions result in a decrease in the amount of high frequency content and an increase in the low frequency content of the EMG. These spectral findings are explained by a shift of the power spectrum of the EMG towards lower frequencies caused by a decrease in conduction velocity of the action potentials (Lindström 1970).

Studies of localized muscle fatigue by means of spectral changes of the EMG have become an important tool in ergonomic research. A drawback of earlier methods has been the difficulty of comparing fatigue effects between individuals and even in one individual in different work situations and different electrode placements. This drawback has been resolved by the introduction of the so called "fatigue index" (Lindström et al 1977).

## THEORY

Under some simplifying assumptions, the power spectrum W of the EMG can be written as

$$W(\omega) = v^{-2}\, G(\omega/v). \qquad (1)$$

Here, ω is angular frequency, v is the action potential conduction velocity and G is a function accounting for aspects such as electrode geometry, distance between electrode and active muscle fibers, and summation of the activity of active fibers. The spectral moment of order n is

$$M_n = \int_0^\infty \omega^n\, W(\omega)\, d\omega. \qquad (2)$$

The conduction velocity can be expressed as functions of these moments. As an example, consider the moments of orders zero and one:

$$v = \int_0^\infty G(x)dx \Big/ \int_0^\infty x\, G(x)dx \cdot M_1/M_0 \qquad (3)$$

Thus, in a measurement situation where the function G keeps constant, the ratio $M_1/M_0$ will track the time course of the conduction velocity v. As the function G is mainly dependent upon electrode geometry and anatomical factors, which are essentially constant in a given experimental situation, it is possible to evaluate the relative amount of decrease of the conduction velocity. This example shows that by using ratios between spectral moments it is possible to construct parameters of the EMG allowing comparison of fatigue effects between individuals,

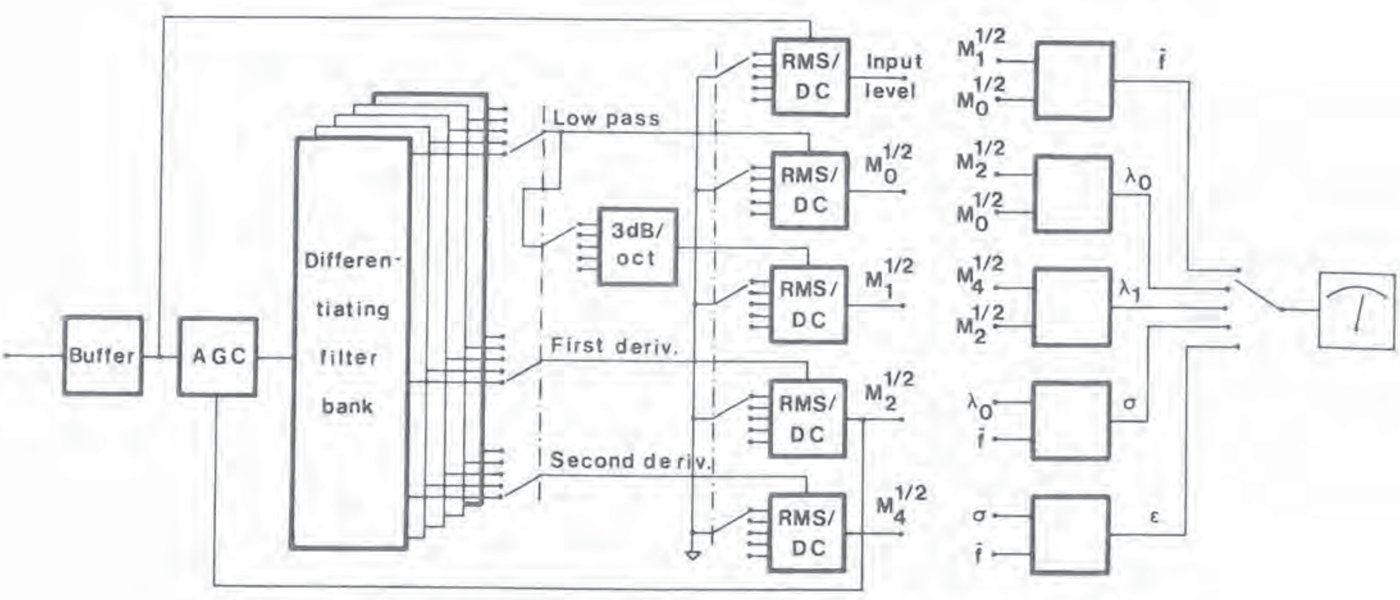


Figure 1. Block diagram of the spectral moment analyzer.

between different work loads etc.

## CONSTRUCTION

The spectral moment analyzer evaluates several basic parameters characterizing the power spectrum of a noisy signal. The square roots of the spectral moments $M_n$ as defined by eq.(2) are estimated and presented as intermediate results. The following set of variables is then calculated: The mean frequency $\bar{f}$ - the center of gravity of the power spectrum - which equals $M_1/M_0$; the density of zero crossings $\lambda_0$ (Rice 1945), which equals $(M_2/M_0)^{1/2}$; the density of turning points $\lambda_1$, which equals $(M_4/M_2)^{1/2}$; the bandwidth $\sigma$, defined as $(\lambda_0{}^2-\bar{f}^2)^{1/2}$, and finally the relative bandwidth $\varepsilon$, defined as $\sigma/\bar{f}$. The bandwidth definition is in analogy with the definition of the standard deviation of a stochastic variable. The variables are evaluated in the analog electronic computing blocks shown in Fig. 1. After passing through an input buffer amplifier, the signal is fed to two blocks. The first is an RMS/DC converter which gives the root mean square value (RMS-value) of the input. The second - in the main stream of calculations - is an automatic gain control circuit (AGC). From the AGC, the signal is fed to a set of filters with different lowpass 3-dB frequencies. Thus a filter suiting the frequency content of the input can be chosen. These filters all have three outputs: a lowpass, the first and second derivatives of the lowpass. After RMS/DC conversion the square roots of the unnormalized spectral moments $M_0$, $M_2$ and $M_4$ are obtained. To keep the signal levels high at this point of the calculations, the square root of $M_2$ controls the AGC. The square root of $M_1$ is estimated by feeding the lowpass filtered signal to a filter with a +3 dB/octave amplitude characteristic, followed by an RMS/DC conversion. The time constant of all RMS/DC conversions can be chosen, compromising between accuracy of estimation and speed of response. Finally, all further computations are carried out in multifunction circuits which perform multiplication, division and square-root calculation as demanded. The details of the analyzer are described by Broman (1979).

## APPLICATION

The instrument has been applied to a study of localized muscle fatigue in the shoulder region. This is the theme of another report at this conference. The full printout from another experiment is shown in Fig. 2. The subject was positioned in a chair with support of the elbow, the upper arm vertical and the forearm horizontal and in a supine position. A pair of surface electrodes was applied to the belly of the lateral head of the biceps brachii. The two electrodes were positioned along the muscle fibers and approximately two cm apart. The EMG was amplified, bandpass filtered - 3 dB frequencies 10 Hz and 1500 Hz - and fed to the analyzer. The lowpass filter of the analyzer had a 3 dB frequency of 500 Hz, the time constant of the RMS/DC conversions was one second. Fig. 2 shows the result of a 34 s run with 11 kg load in the hand. As can be seen, the mean frequency, the intensities of zero crossings and turn points, and the bandwidth all follow approximately the same time course, in accordance with the theory. The relative bandwidth keeps almost constant - a slight decrease towards the end of the run - indicating that the shape of the power spectrum is essentially unchanged. The decrease of the mean frequency from 95 Hz to 48 Hz shows that the average conduction velocity of the action potentials decreases to some 50% of the initial value during this 34 s heavy contraction.

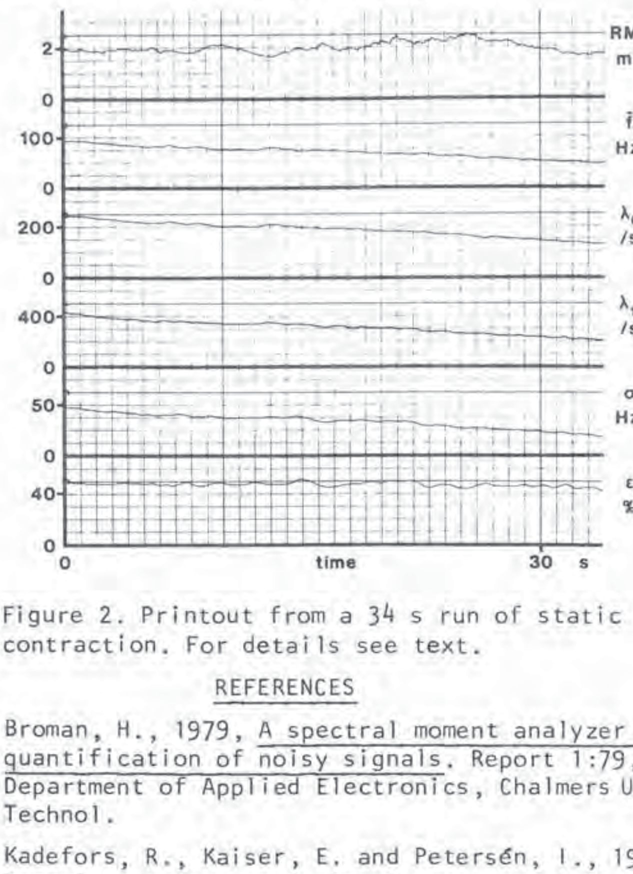


Figure 2. Printout from a 34 s run of static contraction. For details see text.

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MEDIAN FREQUENCY OF THE MYOELECTRIC SIGNAL AS A MEASURE OF LOCALIZED MUSCULAR FATIGUE

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## INTRODUCTION

Since the work of Piper (1) in 1912, the frequencies of the myoelectric (ME) signal have been known to "slow" as a contraction is sustained. Lindström et al. (2) have presented a mathematical model for the power density spectrum of the ME signal which explicitly depends on the conduction velocity of muscle fibers. The resulting expresssion can account for the slowing of the ME signal. Specifically, a decrease in the conduction velocity causes a compression of the spectrum into lower frequencies. A current hypothesis for this change in the spectrum is: 1) acidic metabolic by-products of the contraction accumulate 2) pH decreases 3) conduction velocity decreases and 4) the spectrum is compressed. There is much supporting evidence in the literature for this hypothesis; however, there is no direct evidence which demonstrates the causality or even correlation between pH and conduction velocity, or conduction velocity and ME spectral compression. Yet, the spectral compression is a consistent repeatable occurrence which accompanies sustained exhaustive contractions. Hence, a measure of the spectral compression may be useful to monitor localized muscular fatigue associated with sustained contractions.

## TECHNIQUE AND DEVICE

The spectral compression may be monitored by continuously calculating a representative frequency, such as the mode, median or mean frequency or zero-crossings. In our laboratory we have developed and used a technique which calculates the median frequency on-line and in real-time (3,4,5,6). A prototype is shown in Figure 1 and it is referred to as the Muscle Fatigue Monitor, or MFM.

The technique is very simple to implement in analog hardware. The ME signal is passed through parametric filters to divide the spectrum into two regions. The cutoff frequencies of the filters are set by a control voltage which forces the signal power output of the filters to be equal. Hence, the control voltage uniquely corresponds to the median frequency. A modified version of the above technique is also being constructed which will be compact (hand-held) and suitable for routine clinical use.

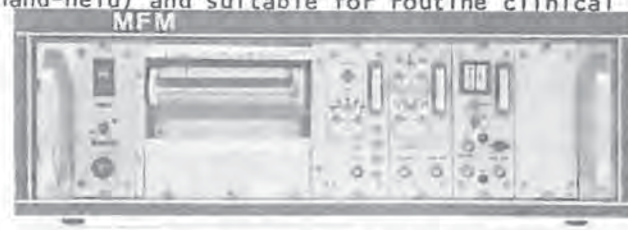


Figure 1. The Muscle Fatigue Monitor, MFM.

## EXPERIMENTS AND DATA ANALYSIS

Experiments were conducted to obtain surface ME signals from normal human subjects as they performed sustained constant-force, isometric contractions of the deltoid and first dorsal interosseous muscles at 20%, 50% and 80% of their maximal voluntary contractions (MVC's). The electrode consisted of two parallel silver bars 10 mm long and whose outside edges were 10 mm apart. The reference was a DISA ground strap placed around the wrist. The electrode was connected to a differential amplifier whose single-ended output was connected to the MFM via an opto-isolator amplifier.

Normally the data would be analyzed in real-time and on-line. However, in order to perform a rigorous analysis of the behavior of the median frequency the ME signal was recorded on an FM tape recorder and was later played backwards. This maneuver was necessary because the compression of the ME spectrum is most rapid at the beginning of the contraction. Since the MFM uses an analog technique the initial fall-off may be "smeared" by the lags in the circuitry. The recorded ME signal was then passed through the MFM and the median frequency was digitized. Thus, the median frequency was backward in time but the estimate was more accurate at the beginning of time. Once digitized, the median frequency curve was "flipped" back. A median frequency curve is shown in Figure 2. This curve was obtained from an ME signal recorded from the first dorsal interosseous muscle during a sustained isometric contraction at 50% MVC.

A curve-fitting routine was developed to globally fit the median frequency from the beginning to the end of a sustained contraction. The parameters of the curves were used as estimates of the initial and final value of the median frequency, and the rate of the inital fall-off.

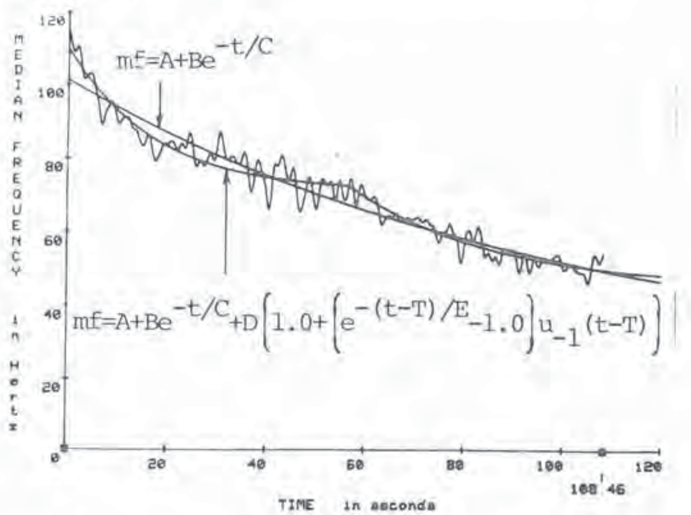


Figure 2. Median Frequency of the ME.

## RESULTS AND DISCUSSION

Previously, Lindström et al. (2) reported that an exponential decay to zero well described the behavior of their mean frequency parameter. However, none of the median frequency curves decreased to zero. Initial attempts were made to fit the median frequency curves with an exponential decay to a final non-zero value. However, as seen in Figure 2, there is a tendency for the median frequency to level-off in the middle of the contraction. This region is followed by another decrease to a final value. Therefore, a more complicated equation was required to globally fit the median frequency, mf , as a function of time, t. The form of the equation is:

$$mf=A+Be^{-t/C}+D\left[1.0+\left(e^{-(t-T)/E}-1.0\right)u_{-1}(t-T)\right]$$

It represents an initial exponential decay followed by a delayed exponential decay. The parameters of the curve are A,B,C,D,E, and T which are to be calculated from the curve-fitting routine. The more complicated curve provides a more significant fit than the single decaying exponential (see Fig. 2). Similar results were obtained for 20% and 80% MVC.

Many other functions could be used to fit the median frequency, however, the proposed equation has an appeal in terms of muscle fiber-type physiology. Conceivably the initial decay may represent the effect of "fast-fatiguing" white fibers; and the second delayed decay may represent the effect of the "fatigue resistant" red fibers.

The results of all the contractions will be presented.

## CONCLUSION

A technique has been developed to calculate the median frequency of the ME signal on-line and in real-time. The median frequency of the ME signal obtained from a sustained constant-force isometric contraction is well described as a function of time by an initial exponential decay followed by a delayed exponential decay.

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EFFECT OF ISCHEMIA, COOLING & LOCAL ANESTHESIA ON THE MEDIAN FREQUENCY OF THE MYOELECTRIC SIGNALS.

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## INTRODUCTION

It has long been held that conduction velocity of muscle fibers is affected by variations in the blood supply or temperature. Stalberg (1966), using single fiber EMG, reported a reduction in the propagation velocity in the muscle fibers either during ischemia or during a lowering of the muscle temperature. Also, procaine anesthesia has been reported to reduce the sodium transport in the muscle fiber membrane (Inoue & Frank, 1962). This sodium immobility may result in conduction velocity changes. However, it has been difficult to record such changes accurately from myoelectric signals during a normal muscular contraction. In this study a non-invasive technique has been used to measure changes in the median frequency of the myoelectric signals during ischemia, local muscle cooling and procaine anesthesia infiltrated into the muscle. Changes in the median frequency may be used to measure changes in the frequency spectrum of the myoelectric signal during sustained muscle contractions which Lindstrom et al (1970) have shown to be related to the conduction velocity of muscle fibers.

## MATERIALS AND METHODS

Thirteen normal adult male volunteers, whose age ranged from 18-53 years, were tested using the method of Stulen and De Luca (1978). Each subject was asked to abduct the index finger and elicit an isometric contraction equivalent to 20% of his maximal voluntary contraction (MVC). He was requested to maintain the constant force output for 5 sec. and this was repeated at intervals. The procedure was repeated for 80% MVC. The force was measured with a force gauge whose output was displayed on an oscilloscope. The myoelectric signal was recorded by two silver bars, spaced 10 mm. apart, and located on the skin overlying the first dorsal interosseous muscle. The myoelectric signal was analyzed with a muscle fatigue monitor (MFM). This is a device developed in our laboratory which is capable of tracking the median frequency automatically, in real-time, and non-invasively. In this investigation the initial median frequency, i.e. the median frequency at the beginning of the contraction, was calculated for every muscle contraction. The procedure was repeated separately with the muscle in ischemic condition, cooled, and anesthetized. Ischemia was induced by a pneumatic cuff placed on the wrist for 10 min. while inflated at 220 mmHg. The muscle was cooled by placing ice bags directly on the skin overlying the muscle for 10 min. and the intramuscular temperature was monitored via a needle thermocouple. Local anesthesia was administered by infiltrating 0.5 cc of 1% procaine directly into the first dorsal interosseous muscle. The myoelectrid and force signals were stored for further analysis.

## RESULTS

In some subjects the value of the initial median frequency of the 80% MVC was higher than that of the 20% MVC (fig 1), whereas, in other subjects the opposite was noticed. It is speculated that this may have been due to the different shapes of the action potentials of the higher-threshold motor unit types amongst subjects. Ischemia, cooling and local anesthesia showed significant decreases in the initial median frequency (fig 1). However, during ischemia, the 80% MVC showed a more prominant reduction in initial median frequency than did the 20% MVC (fig 2). A gradual reduction in the initial median frequency with time of ischemia and with a decrease in muscle temperature was noticed. The motor unit action potentials significantly increased in duration and it was more difficult for the subject to maintain the 80% MVC level. The median frequency took 10 min. for full recovery after ischemia and cooling. Moreover, the initial median frequency of the 80% MVC recovered faster than that of the 20% MVC after ischemia. Both contraction levels overshot the pretested value with a larger increase in the median frequency for 10 min. (fig 2).

During anesthesia, a gradual reduction in the initial median frequency was noticed with time up to 30 min. The myoelectric signal (EMG) of the muscle was significantly smaller than the pretested value (Fig. 1.). Some subjects were unable to track for the 80% MVC 5 min. after the anesthesia, but recovered in 5-10 min. afterward. In such cases, measurements were discarded during this period.

## DISCUSSION

In ischemia the conduction velocity change are believed to be related to the accumulation of metabolic biproducts, which in turn may cause the shift of the frequency spectrum to a lower value (Mortimer et al., 1970). The difference noticed between the 20% and 80% MVC during ischemia demonstrated the significant effect of metabolic biproducts and blood supply on the high threshold units. It seems probable that the different proportions of the fast and slow muscle fiber types of the 80% and 20% MVC respectively, have a distinguishing different effect on the median frequency during ischemia. Moreover, during cooling the action potential has been reported to increase in duration (Sabbahi Awadalla 1976). This phenomenon, which was verified in this study, may also be attributed to the reduction in the intramuscular conduction velocity due to changes in membrane permeability of the muscle fibers. The significant reduction in the initial median frequency, with temperature decrease, may be due to the continuous increase of the action potential duration being reflected as a compression of the frequency spectrum. In this study a direct relationship has been noticed between the action potential duration and the median frequency of the myoelectric signal. However, during anesthesia, the intramuscular conduction velocity has been reported to be reduced in conjunction with a decrease of sodium ion transport (Inoue et al., 1962). In our study the initial median frequency also decreased during anesthesia. The reduction in the amplitude of the myoelectric signal during anesthesia noticed in this experiment is most likely due to a decrease in the amplitude of the motor unit action potential.

## CONCLUSION

It is proposed from this study that changes in the conduction velocity, though they may be caused by different mechanisms, can be related to the median frequency measured by the muscle fatigue monitor.

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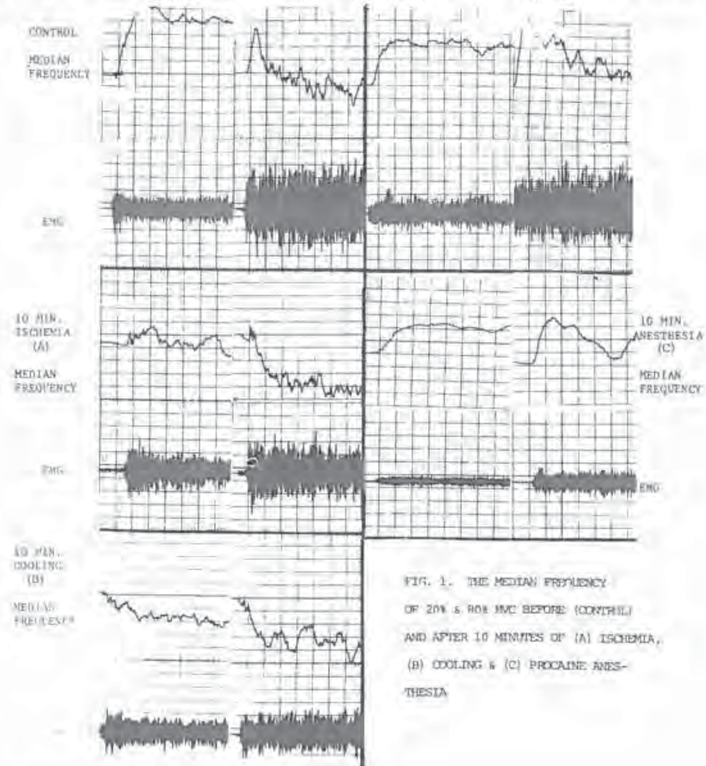
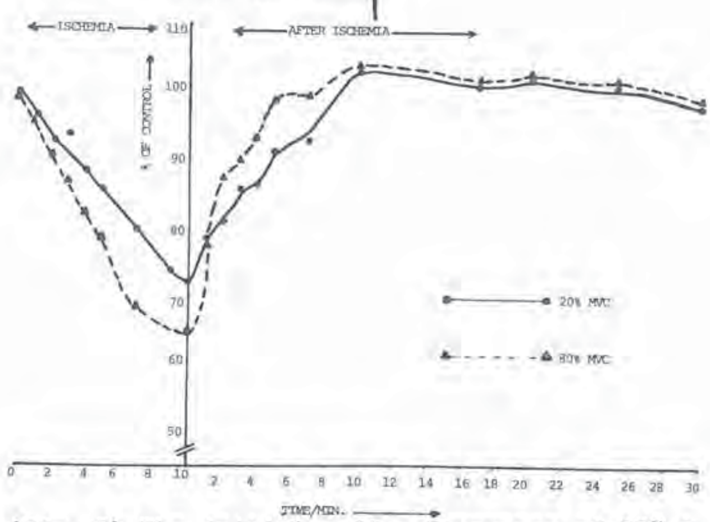



FIG. 1. THE MEDIAN FREQUENCY OF 20% & 80% MVC BEFORE (CONTROL) AND AFTER 10 MINUTES OF (A) ISCHEMIA, (B) COOLING & (C) PROCAINE ANESTHESIA




(Fig. 2) The initial median frequency of 20% & 80% MVC during & after Ischemia.

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# ELECTROMYOGRAPHIC EVALUATION AND MODELLING OF LOCALIZED FATIGUE DURING PERIODIC MUSCULAR WORK

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## Introduction

During work with small or medium sized muscles or groups of muscles, localized muscle fatigue can occur without essential effect on pulsrate or rate of breathing. To quantify the load on the body under such circumstances, one is referred to electromyographic methods. Localized muscle fatigue can be studied with spectrum analysis of myoelectric signals by means of the fatigue index method (Lindström et al, 1977). The method involves power spectrum analysis and yields relative values of action potential propagation velocity along the muscle fibers, decreasing velocity being associated with increasing fatigue.

The myoelectric signal response during static, isometric contraction is well known: the action potential propagation velocity decreases approximately exponentially with time. Under varying muscle loading, the propagation velocity changes are irregular (Örtengren, 1972). During periods of forceful contractions, the velocity decreases approximately exponentially with a characteristic time-constant which depends on the load level; during periods of rest, or contraction of negligible force, the velocity increases approximately exponentially with a recovery time-constant which may be considered as independent of the load level.

The purpuse of the present investigation is to study localized muscle fatigue during periodic muscular work and to incorporate the results into a mathematical model.

## Methods

Five healthy volunters participated in the experiments, each in several sessions. Myoelectric signals from the right brachial biceps were picked up by means of bipolar surface electrodes, low pass filtered, and fed to a digital computer in which the analysis was performed. The loading was accomplished by means of weights lifted in a handle. To indicate when the load was lifted, a load cell was introduced between the weight and the handle. A signal from the load cell was also fed to the computer. Because of the large amount of data stored in the computer's disc memory, the analysis was performed in several steps. The first step included fast Fourier transformation and calculation of relative conduction velocity via spectral moments. During the continued analysis theoretically calculated relationships for the velocity development were fitted to the experimental findings. The parameters of the theoretical functions were estimated by means of parameter identification algorithms. For each subject, further, the possibilities were studied to predict the obtained values by means of the duty cycle and the values obtained for the same subject during isometric, isotonic contraction.

## Results

For most experiments the variance of the velocity along an average line was moderate and it was possible to fit exponential functions to the fatigue and recovery courses and to determine the parameters of the exponential curves. The most striking effect in the results is the increasing endurance time when the duty cycle is decreased from one, i. e. continuous loading, to lower values. When the load was 80 newton, the duty cycle 0.5, and the period time 15 s, no signs of fatigue was found for any of the subjects within eight to ten minutes. When the same load was continuous, none of the subjects reached an endurance time above 3 minutes.

The fatigue curve approaches a certain final value, a plateau, in almost all experiments. Both the level of this plateau and the time-constant for the fatigue curve could well be predicted by the mathematical model.

An interesting finding during the investigations is that the time-constant of the recovery is not constant but shortens when the fatigue depth is considerable.

## Discussion

The physiological basis for the theoretical model is as follows: When the muscle develops a force of sufficient strength, the blood flow through the muscle is reduced, totally or partially (Barcroft & Millen, 1939), resulting in an almost immediate transition to anaerobic metabolism with lactate as the main end product

This transition pertains to all types of muscle fibers - red fibers also change their oxidative metabolism during anoxia. The metabolites diffuse easily across the muscle fiber membranes, thus, all fibers will be affected by the lactate even if the production is confined to certain fibers only. The decrease in intracellular pH reduces the excitability of the membranes (Tasaki et al, 1965) and causes a diminishing of the propagation velocity of the action potentials. Relaxation of the muscle restores the oxygen supply and enables a wash-out of the acid metabolites. This wash-out is mainly limited by diffusion from the cells to the capillaries.

When deriving the model, we assume that the propagation velocity changes exponentially, both during work and relaxation. A more detailed analysis in which the production rate of acid metabolites is considered to be constant during work, and the wash-out is assumed to be diffusion limited, yields a slight modification of the derived equations. These new equations also explain the experimental finding that the recovery time-constant depends on the fatigue depth when the depth is considerable.

The present model describes localized fatigue during repetitive muscular loading. It can easily be adapted to describe fatigue development in aperiodic cases. Apart from applications to basic muscle physiology, the model is of interest in applied ergonomics: Most jobs in modern production contain repetitive elements, often with a static basic component.

## Acknowledgement

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# ELECTROMYOGRAPHIC FATIGUE EFFECTS AND RECOVERY OF ENDURANCE IN FOREARM MUSCLES

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## INTRODUCTION

Most manual tasks in vocational situations as well as in activities of daily living appear to require more force from the forearm flexor muscles than from the extensor muscles. Nevertheless symptoms of forearm extensor muscle overstrain are common. The explanation to this apparent divergence may be that stabilisation of the radiocarpal joint require almost the same extensor force as the coincident flexor force. Thus, a sustained forceful power grip of the hand can be expected to cause fatigue not only of forearm flexor muscles but also of forearm extensor muscles.

Stull and Kearny (1978) recently studied the recovery of muscular endurance following submaximal, isometric exercise of the power grip. The recovery of muscular endurance time was used as a measure of recovery from muscular fatigue. The pattern of submaximal isometric endurance recovery after different interbout rest periods followed a three-component exponential curve. The first component was assumed to be related to CP repletion. The second component, which accounts for almost 50 % of the recovery process, could possibly be related to hyperemic removal of metabolites. The physiological process underlying the third component may be the repletion of cellular energy stores.

The aim of the present study was to reduplicate the study by Stull and Kearny with the addition of electromyographic examination of some forearm muscles. The purpose was to follow the recovery of the myoelectric signs of fatigue in relation to the recovery of the endurance time.

## EXPERIMENT

The flexor digitorum superficicalis, the extensor digitorum and the extensor carpi radialis longus muscles of the right forearm were examined electromyographically in five healthy subjects, one male and four females age 25-35 years. Bipolar surface electrodes were used. The myoelectric signals were recorded on magnetic tape for subsequent computer analysis.

The task to be performed consisted in squeezing a hand-gripping device for as long as possible at a tension of 50 % of maximum voluntary contraction (MVC). When the prescribed tension could no longer be maintained, the subject was given a predetermined rest of 5, 10, 20, 40, 80, 160, 320, 640, 1280, or 2560 seconds. After that rest the subject had to again squeeze the hand-gripping device at a tension of 50 % of MVC for as long as possible. The endurance time in both contractions were measured. Each subject was examined at ten occasions experiencing each rest period at random order.

## DATA ANALYSIS

The percentage of recovery of muscular endurance was calculated by dividing the holding time of the second contraction bout by the time of the first bout.

The myoelectric signals were subject to frequency analysis using FFT-techniques on 1.024 second samples every 10 seconds. From each power spectrum the mean power frequency (MPF) was calculated. The change in MPF over time was estimated (exponential fit) for each contraction period.

The myoelectric signals were also analysed in the amplitude domain. The EMG amplitude levels were determined as root-mean square (RMS) values for consecutive 10 second samples. The RMS versus time regression (linear fit) was estimated for each contraction period.

The typical fatigue effects in the myoelectric signal is a decrease in MPF and an increase in RMS during a constant-force isometric contraction. Persistent fatigue effects from one contraction period to the next is likely to be a lower "start" frequency and a higher "start" amplitude in the second contraction period as compared to the first period. The "start" MPF and RMS values were calculated for each contraction period from the exponential and linear, respectively, regression curves.

## RESULTS

The recovery of muscular endurance proved to be exponential in nature, 50 % of the recovery

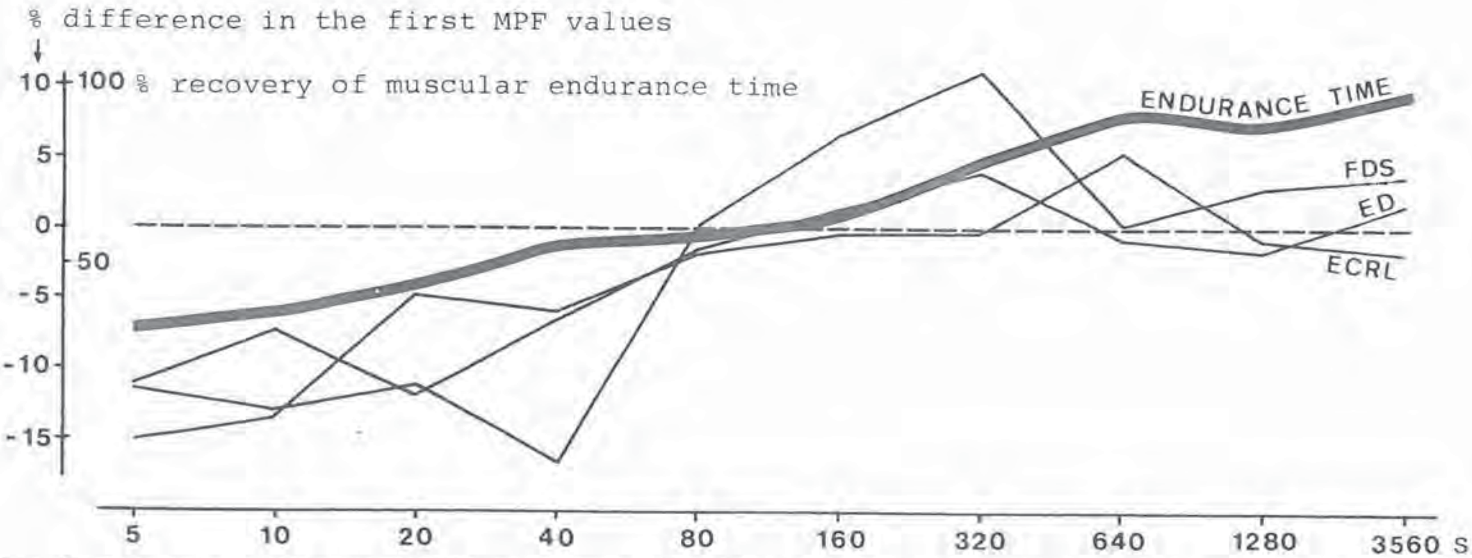


Fig. 1 Percentage of recovery of endurance time (solid line) and persistent fatigue effects in MPF (thin line) as a function of interbout rest time. The "myoelectric recovery" is expressed as the percentage difference in "start" MPF between the first and second contraction, negative values indicating lower "start" values after the interbout rest. FDS = flexor digitorum superficialis muscle, ED = extensor digitorum muscle, ECRL = extensor carpi radialis longus muscle.

taking place within less than 40 seconds. The recovery was almost complete a good 42 minutes (2560 s) after the first contraction bout.

As expected there was a marked decrease in MPF over time in all three muscles during the contractions. Persistent fatigue effects from one contraction to the next appeared in MPF when the interbout period was shorter than 80 seconds (fig. 1) corresponding to approximately 60 % recovery of muscular endurance.

The amplitude of the myoelectric signal showed an increase over time during the contractions, and persistent fatigue effects from one contraction to the next appeared when the interbout period was less than about 80 seconds.

## DISCUSSION

There was a difference between the recovery of muscular endurance time and the myoelectric signs of recovery from fatigue after sustained constant-force isometric contractions. The "myoelectric recovery" was complete long before the recovery of the muscular endurance time of contraction. This indicates that absence of myoelectric changes does not prove the absence of muscular fatigue. The myoelectric changes are probably, as suggested by Lindström et al (1970), due to the effect of accumulation of metabolites in the muscle fibres. The persistent and long lasting fatigue effects which may be due to incomplete repletion of cellular energy stores obviously does not effect the myoelectric signal characteristics.

The presence of fatigue effects in both the flexors and the extensors of the forearm shows that the power grip of the hand loads both muscle groups and may explain why symptoms of muscular overstrain are common in the forearm extensor muscles after heavy manual work.

## CONCLUSION

Recovery of myoelectric signs of fatigue occur more rapidly than the recovery of muscular endurance time. Sustained power grip will cause fatigue of forearm flexor as well as extensor muscles.

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## ACKNOWLEDGEMENT

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UNITS, TERMS AND STANDARDS IN THE REPORTING OF ELECTROMYOGRAPHICAL RESEARCH
(First Interim Report of ISEK Committee on EMG Terminology)

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## THE PROBLEM

One of the signs of maturity in a scientific area is the ability of its researchers to communicate with clarity the results of their work. In EMG research there has been serious communications problems because of the lack of completeness, correctness and consistency of reporting.

In a recent review of a dozen reputable articles by one of our graduate classes it was agreed that we could replicate only two of them because of lack of details and confusion over the protocol, the recording equipment or the processing technique. At a high level international congress two years ago some of us attended a session on EMG and found the term "integrated EMG" to be used 3 different ways, and only one of the 6 papers was using it correctly. No wonder there are conflicts and misunderstandings.

To correct this problem, our ISEK President, Herbert Ladd, has asked that an ad hoc committee be formed to make recommendations. This paper represents the first interim report of the Committee. The purpose of the report is not to tell researchers how to do their work (e.g., which processing technique to use), but rather how to report it technically correct. In taking on this task the Committee felt there was no need to define new terms, but rather use those which were already well defined and accepted in engineering, mathematics and the biological sciences. The present report concentrates on the correct use of terms and units, with some suggestions being made concerning desirable standards.

Four sub-areas of EMG recording and processing are dealt with:

1. Electrode and amplifier characteristics
2. Temporal processing techniques
3. Frequency processing
4. General experimental protocol: anatomical, physiological and kinesiological.

## 1. Electrodes, Amplifiers and Recording Equipment

The purpose of the electrodes and preamplifier is to provide an undistorted and amplified EMG signal in its "raw" form for further processing. Because each researcher has his own particular recording system it is not mandatory to report the amplifier gain; rather, it is adequate to report the signal as seen at the electrodes (in mv or µv). The following checklist will serve as a basis of reporting:

(i) Electrode type (give details of surface and indwelling) and lead (shielded or unshielded).

(ii) Skin preparation, including skin impedance, if possible.

(iii) Amplifier bandwidth; as well as reporting lower and upper cut-off frequencies the order of the filter should be specified. It is not desirable to specify the time constant of the low pass filter, because such a description is only accurate when a first order filter is used. For surface electrodes a B.W. of 10 Hz to 1000 Hz is suggested, and 20 Hz to 2000 Hz for indwelling is adequate.

(iv) Amplifier input impedance, which should be at least 100 times the electrode/skin impedance; for surface electrodes 1MΩ is usually adequate, and sometimes higher for indwelling electrodes.

(v) Amplifier common mode rejection ratio (CMRR). It is desirable that this should be above 80 dB, preferably 90 dB.

(vi) Recording equipment bandwidth: for pen recorders specify high frequency cut-off; for instrumentation tape recorders specify bandwidth.

(vii) Sampling rate of analog-to-digital converters. Remember, the rate, in theory, should be twice that of the amplifier high frequency cut-off, and in practice it should be about 4 times as high.

## 2. Temporal Processing

The "raw" EMG signal is rarely used as a quantitative measure, although pen or oscilloscope recordings are frequently used as semi-quantitative evidence. Temporal processing, either digital or analog, is normally used to arrive at an average or total measure of electrical activity during a muscle contraction. It is in the temporal processing that there is the greatest confusion. How can we reconcile integrated EMG (iEMG) when it is reported in mv, mv/s and mv.s? Or, how can we possibly replicate an experiment that has all EMG measurements in "arbitrary?" units?

To assist us to sort out this confusion let us examine how the same EMG signal can be processed a number of common ways, as presented in Figure 1. The more common temporal processing is based on the full wave rectified EMG, as shown on trace 2. Its units are in mv, the same as the "raw" signal. Integrated EMG was a term introduced by Inman (1952); unfortunately, he was not referring to a true electronic integration, but erroneously to a linear envelope detector (full-wave rectifier + low pass filter). In trace 3 we see the linear envelope detector with units in mv; all that needs to be specified are the low pass filter details (cut-off frequency, order and type of filter).

True mathematical integration of the EMG is often done in 3 basic ways, but in all cases the units to be employed are mv.s. The iEMG during the time of a total contraction is merely the time integral of the full-wave rectified signal from the start to the end of the electrical activity (trace 4). A second method is to integrate over a short period of time, reset, then repeat. Therefore it is necessary (trace 5) to specify the integration period (ms) as well as the amplitude (μv.s). A third and popular method is to integrate to a preset amplitude, then reset and repeat. Each reset produces a pulse not unlike a neural pulse. The pulse rate increases and decreases with the level of contraction. Since each pulse represents a finite value of iEMG it must be specified. In traces 6 and 7 we see the integrations and pulses, each representing 38 μv.s.

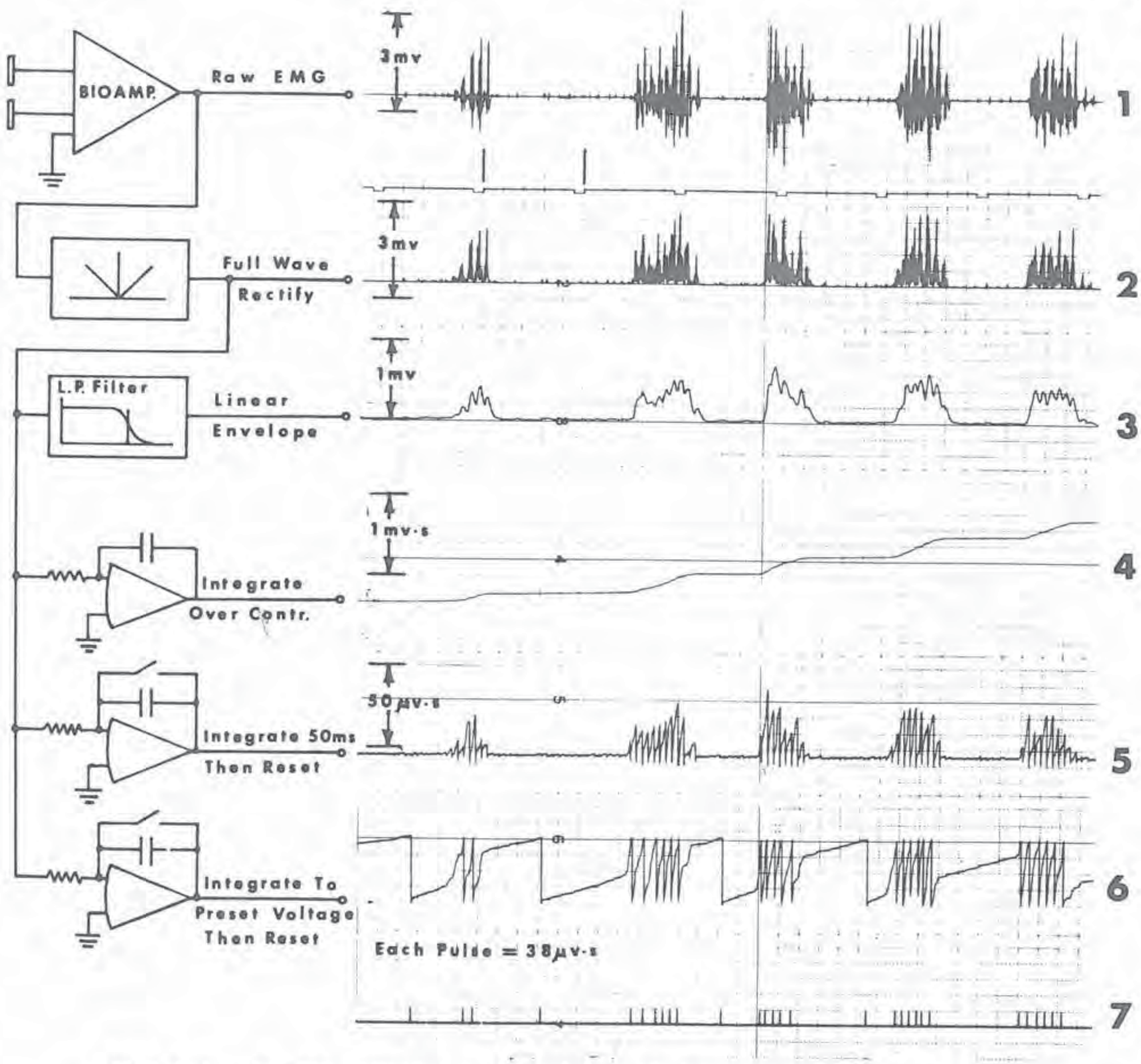


Figure 1 - Common temporal processing techniques; see text for details

The average EMG, expressed in mv, requires a mathematical integration over the time of the contraction. For many research projects the contraction is isometric-isotonic, thus the average can be taken over variable lengths of time. Therefore the researcher should specify the averaging period.

Many other temporal processing schemes have been reported or proposed. However, there are far too many to report and until there is consistency of reporting within the more common techniques we feel it would be ill advised to cloud the issue with too much detail.

3. Frequency Analyses

The harmonic content of the EMG is higher for indwelling electrodes than for surface, and can vary with fatigue (Kadefors, et al., 1968), the level of contraction, the type of fibres present in the muscle, and with certain myopathies and neuropathies. Two spectral density plots can be used; an amplitude density function (μv/Hz vs Hz) or a power spectral density function, PSDF, ($\mu v^2$/Hz vs Hz). The latter function is more common, and if the units are correctly specified the area under the curve yields the power of the EMG in $\mu v^2$. Often researchers are only interested in reporting the relative power at each frequency and therefore normalize the peak power = 1.0. The peak usually occurs around 100 Hz and care must be taken that a false peak does not occur because of a small amount of hum in the signal. Such an occurrence is usually obvious when the PSDF shows a spike in the curve at 50 or 60 Hz (Kwatny, et al., 1970).

Analog filtering of the EMG is a convenient on-line method to observe the spectral content. Tuned filters that are used should be specified by their B.W. and centre frequency. The output of each filter is usually reported in μv. Because each filter's B.W. is usually proportional to its centre frequency it is often preferable that the output is divided by the B.W. to get an amplitude or power spectral density plot.

The mean spectral frequency (MSF) has been used as a single measure to indicate overall shifts in the spectrum as a result of fatigue, recruitment, and certain pathologies, and is given by the following expression:

$$MSF = \frac{\int_{f_1}^{f_2} fG(f)df}{\int_{f_1}^{f_2} G(f)df}$$

where G(f) is the PSDF.

MSF is in Hz and shows net shifts towards the high or low ends, but may not show a change if there has been cancelling frequency shifts at both ends of the power spectrum.

4. General Experimental Variables

Bouisset (1973) made a strong plea for more details from all researchers when they report their experimental protocol. At this time we consider the following checklist may be valuable to ensure that the study may be replicated or compared:

(i) Age, sex (and, if pertinent, height and weight) of all subjects
(ii) State of subject's fatigue, training, pathology
(iii) Anatomical position of electrodes, including all agonist and antogonist muscle sites
(iv) Type of contraction: isometric (constant length), isokinetic (constant velocity), isotonic (constant tension); range of tensions, torques, joint angles, velocities, etc.

Conclusions

The more common errors in reporting of EMG research have been noted, and proper units and necessary variables have been detailed. It is hoped that all researchers will use these items as a checklist for reporting their work.

Future steps by the Committee will include development of desirable standards based on inputs from many researchers. Also, steps will be taken by ISEK to encourage the proper use of terms and units at ISEK conferences and in research journals.

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NOTES

NOTES

# CLINICAL II

## Session 5B

TIBIALIS ANTERIOR REINNERVATION BY COLLATERAL BRANCHING WITH AND WITHOUT ELECTROTHERAPY.

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## INTRODUCTION

The value of electrotherapy of denervated muscles has yet to be proved with respect to reinnervation ; in fact some experimental research points to an adverse effect on it (see Ref).The present study deals with the rate of reinnervation by collateral branching (r.c.b.) in a group of patients (8) treated with muscle electrical stimulation (m.e.s.)and in a group (6)assisted only with physiotherapeutic treatment (ph.tr.).

## PATIENTS

All the patients were affected by unilateral axonothmesis of the sciaticus popliteus externus (S.P.E. or lateral popliteal) nerve,mainly of traumatic origin; those included in this study did not show any sign of direct reinnervation six months after the injury (see method: 4). Group A:6 (mean age 47 y.) out of 21 patients treated three times weekly with m.e.s.of tibialis anterior (T.A.)by means of 50 msec.trapezoidal impulses, 22-31 mV, skin impedance reduced to < 20 m.ohms 1/sec, for a period of 20'-30'. Group B: 6 (mean age:32 y.) out of 18 patients treated with ph.tr.four times weekly.
Supplementary Group (S): 2 (16 y.; 24 y.) out of 9 patients were treated six times weekly with m.e.s. of T.A.by means of 10 msec.impulses, 30-35 mA. at < 20 m.ohms,5/sec, for a total period of 60'and with ph.tr.

## METHODOLOGY

1) I/t curves have been measured for T.A.
2) The degree of fibrillation potentials (f.) recorded in a 10' period after mechanical stimulation of the muscle was classified on a 5 points rating scale according to the duration and intensity of the trains of f. (*)
3) Both S.P.E. and sciaticus popliteus internus (S.P.I. or medial popliteal) nerves were electrically stimulated through surface electrodes and the M response was recorded from T.A. with surface electrodes. If a response was elicited :
4) Selective electrical stimulation of the single SPE and SPI with needle electrodes was carried out to discriminate between direct reinnervation ( from SPE) and r.c.b. from SPI; in fact this nerve can normally contribute to the innervation of a small back portion of T.A. and it was not damaged in these patients.

## RESULTS

1) The degree of f.in Group A decreased from 5/4 at the 1st month after injury to 1/0 at the 4th - 5th month; on the contrary in the Group B it remained at 4/3 until the 5th month. In the two S-Group-patients it was 2 and 1 at the 5th month.
2) The maximal M response amplitude from T.A. to SPI excitation was expressed as % of the amplitude of the maximal M response of the controlateral T.A. to SPE+SPI excitation (=% M). As it is shown in the fig. the % M values were slightly different in the two groups ; the pre-treatment value was higher in Group A, while the increase at the 6th month was higher for Group B ( 28% - 39% ) than for A ( 4% - 18% ). In the S-Group-patients the initial % M.s were zero and 15% and reached 28% and 42% at the 6th month.
Since the 6th month M-values seem to reach the same maximum in all groups, the concept of a limit top value in absolute amount of r.c.b. in T.A. should be maintained ; consequently the higher M increase in Group could be explained on the basis of the lower previous innervation from SPI in these cases. However it must be pointed out that the patient of Group B recognizable in the fig. where he is represented by the lowest % M at the 3rd month, showed a 32 % M increase at the 6th month , whereas a similar patient in Group A showed only 18 % increase .

## DISCUSSION

The f.rate is thought to be related to an increased number of Ach receptors on the muscle

fibre membrane and it is known that this increase enhances r.c.b. The finding of a progressive decrease of f. in patients treated with m.e.t. ( type B and S )leads to the expectation of a negative effect of m.e.t. on r.c.b.
In fact our M study shows that the T.A.muscles not treated with m.e.t.were provided with a higher amount of r.c.b.
These results seem to support the opinion that r.c.b. could be hindered in some cases by m. e.t.

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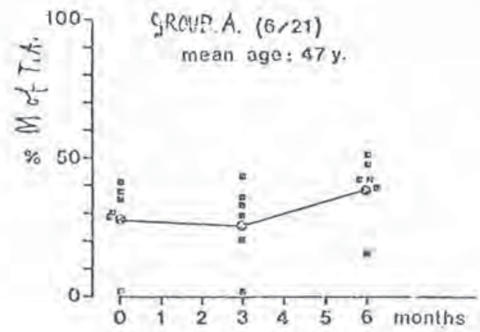
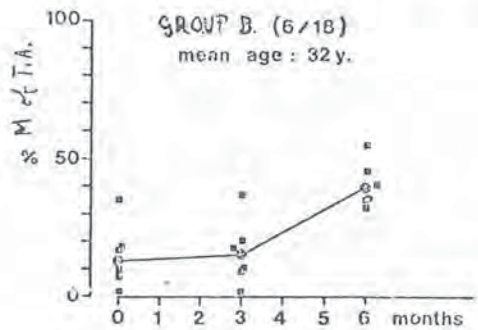
(*) Classification of f. activity

Interference pattern of f. followed by a more scattered pattern present for the all period of 10' with silent interval no longer than 1 sec was classified as 5 ; initial short f. interference pattern with >10% < 20% silent trace in 10' was classified as 4 ; no consistent f. interference pattern and 15% - 30% silent trace, as 3 ; when the silent trace reached 40% : Point 2 ; when it was >40% < 75% : point 1 ; when f. was not higher than in normal subjects : 0.

Mx. M response to SPI excitation

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# AN ELECTROMYOGRAPHIC PROCEDURE FOR THE STUDY OF LATERALIZATION IN THE LOWER EXTREMITIES

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## INTRODUCTION

Little attention has been given so far to the functional asymmetries of the lower extremities. This can partly be explained by the greater practical and socio-cultural interest for the manual activities. Functional asymmetries of the lower extremities are involved in a wide variety of motor activities in daily life, industrial work, dance and sports. They play a most important role in many athletic performances, as well as in the rehabilitation of unilaterally physically handicapped individuals.

The lower extremities have postural and locomotor functions, but can also be used for operant tasks such as kicking a ball or operating a pedal while driving a car or playing piano. The various motor activities of the lower extremities have different mechanical and neuromotor characteristics leading to different kinds of "differentiated complementary bilateralizations" (Vanden-Abeele, 1979). These are the overt expressions of several underlying processes. The investigation of the involved covert neuromotor mechanisms requires the development of new testing procedures, because of the inadequacy of the traditional questionnaire and performance methods.

## ELECTROMYOGRAPHIC APPROACHES

Electromyographic methods have already successfully been used in the study of the functional asymmetries of the distal motor functions of the upper extremities, through the recording of the contralateral motor irradiation (Černáček, 1961; Hopf, Schlegel & Lowitzsch, 1974; Podivinský, 1964).

Our attempts to apply a similar procedure to distal and proximal muscles of the lower extremities were not satisfactory because of concomitant posturodynamic components induced by the relatively high force levels required by the procedure. The study of a frequency-oriented characteristic at lower tension levels appeared thus more appropriate. The functional mobility of muscle, which Vvedenski called "lability" almost a century ago (see Mateev, 1960; Zimkin, 1960) is such a frequency characteristic, quantified by the number of contraction cycles per unit of time.

The present paper reports our efforts (a) to establish a methodology for electromyographic recording of the lability of lower extremity muscles, and (b) to compare the labilities of homonymous muscles in both legs.

## PROCEDURE

Closely spaced (1 cm) surface electrodes are placed above and below the motor points of the M. rectus femoris of each leg. Rectified signals are passed through high-input impedance differential amplifiers. The interelectrode resistance is always kept below 10 kOhms. Subjects are in a comfortable, relaxed, supine position. They are asked to perform repetitive contractions of the M. quadriceps femoris, with full relaxation between successive contractions. Recordings are made in sets of four sequences of fifteen contractions each, according to a right-left-left-right pattern for the subjects with a right leg preference for kicking. A left-right-right-left pattern is used for the subjects with a left leg preference for kicking.

A typical recording session occurs as follows: (a) 3 maximum contractions in each leg. (b) 1 set of self-paced contractions. (c) successive set of metronome-paced contractions, starting at 45 beats per minute, with successive increases of 15 beats per set. (d) 3 maximum contractions in each leg, to verify if any fatigue has been induced.

The performance of a subject is considered as satisfactory when (a) the deflection of the recording pen returns to the baseline after each contraction, indicating full relaxation between contractions, and (b) the required frequency is respected.

## WORK HYPOTHESES

1. In most subjects the muscular lability of the two legs are not equal.
2. Trained individuals have higher lability levels than untrained individuals.

## PRELIMINARY RESULTS

Preliminary case studies indicate that the proposed testing methodology is quite satisfactory, and provide data which are congruent with the hypotheses.

The muscular lability seems to be mainly related to the relaxation properties. The higher lability levels we recorded from trained individuals are then consistent with other studies reporting an improved relaxation ability resulting from training and measured by voluntary relaxation (Ratov & Fedorov, 1959) as well as by reflexogenic methods (Petajan & Eagan, 1968).

## CONCLUSION

The proposed procedure appears to be quite satisfactory as an electromyographic procedure for the study of the functional asymmetries of the lower extremities. Further work is currently in progress.

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ELECTROPHYSIOLOGICAL EVIDENCE OF CROSS-OVER FROM MEDIAN TO ULNAR NERVE IN ULNAR NEUROPATHY WITH CONCOMMITANT URINARY TRACT MALFORMATION. REPORT OF CASE.

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## INTRODUCTION

Anomalous innervation of extremity muscles has been known since the later part of the 19. century. In 1886 Brooks dissected the innervation of intrinsic hand muscles in 30 cadavers. He found aberrations from the normal in ten cases. Rowntree (1949) reported 20 % of 226 patients with nerve lesions had atypical innervations of the intrinsic muscles of the hand. More recently other anatomic variants of innervation have been described by Lambert (1969), Infante and Kennedy (1970) and Gutmann (1970). Schafer and Thane (1897) drew attention to strands between median and ulnar nerves in the forearm. According to Gassel (1964) a branch from the median nerve in the upper forearm joins the ulnar nerve at the medial aspect of mid-forearm. Murphey et al (1946) demonstrated the anastomosis by dissection in one of their cases and mentioned the fact of its illustration in the anatomical text by Poirier and Charpy (1899 - 1907). To exclude possible error caused by spread of exitation due to volume conduction Highet (1942), Rowntree (1949) and Bowden (1954) suggested employment of nerve block technique.

Since anomalous muscle innervation may be the source of considerable error in nerve conduction velocity determination the present case is reported.

## REPORT OF CASE

A 40 year old right handed clerk was referred for electrophysiological studies to roule out left ulnar neuropathy. Four months prior to referral patient incurred a severe contusion of his left elbow. Ever since he had been complaining of "pin and needles" sensation and numbness of left ring and little finger with transitory weakness. Physical examination of both upper extremities was unremarkable. There was definitely no weakness or wasting of intrinsic muscles of either hand. Electromyographic examination (EMG) and motor nerve conduction velocity (MNCV) determinations were performed.

## TECHNIQUE

Cathode oscilloscope with bipolar concentric needle electrodes, visual display and auditory monitoring (EMG); supramaximal percutaneous stimulation of ulnar and median nerves, bilaterally, recording with beckman surface electrodes from appropriate muscles (MNCV). Infiltration of left antecubital fossa with 1 % procaine solution to achieve nerve block of left median nerve.

Muscles tested: Representative muscles of both upper extremities including intrinsic muscles of both hands.

## RESULTS

I. EMG: In all ulnar nerve supplied muscles of left hand there were, at rest, positive sharp waves and fibrillation potentials. During voluntary contraction motor units were discrete, of normal duration and amplitude. There was fairly good recruitment upon moderate volition; a dense interference pattern could be achieved on maximal volition. All median nerve supplied muscles of left hand and all intrinsic muscles of right hand showed normal electromyographic findings.

II. MNCV: Supramaximal stimulation of median nerve at standard sites, bilaterally yielded normal values including terminal latencies (3.5 msec left; 3.1 msec right). Right ulnar nerve showed also normal conduction velocities. Conduction velocity of left ulnar nerve from axilla to elbow was slowed down to 40 m/sec (normal 63.4 (SD5.3) m/sec) compared to 56 m/sec on the right. An attempt at supramaximal stimulation of the ulnar nerve at the ulnar groove of left elbow gave no response. However, stimulation of median nerve in left antecubital fossa evoked a strong muscle action potential in abductor digiti quinti. Stimulation of site of left ulnar nerve at wrist brought about an evoked action potential in the abductor digiti quinti with terminal latency of 9.4 msec.

Stimulation of left median nerve at wrist did not evoke any response in the abductor digiti quinti.

## DISCUSSION

Electromyographic and electroneurographic findings in this patient are consistent with axonotmesis due to ulnar neuropathy as well as with anomalous cross-over from median to ulnar nerve. One might speculate over the cause of the terminal latency of 9.5 msec: The terminal portion of the anomalous branch might be considerably longer than the skin distance mesured or the nerve might have assumed an irregular course. There might be a delay of conduction through the myoneural junction.

To ascertain that we really deal with innervation of the left abductor digiti quinti by the median, the left antecubital fossa was infiltrated with 1 % solution of procaine: Supramaximal stimulation of the median at this site as well as at the medial aspect of left wrist did now not evoke any response.

Review of patient's record revealed a double ureter on the right, diagnosed by I.V. Pyelogram. This additional malformation within another system is reported for its interest. An interrelationship between these two concommitant anomalies is, at best, highly hypothetical.

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# MULTITERRITORIAL ELECTROMYOGRAPHY IN DUCHENNE MUSCULAR DYSTROPHY : A KINESIOLOGICAL STUDY

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## INTRODUCTION

The mechanism of muscular degeneration in Duchenne muscular dystrophy (D.M.D.) is not yet understood. In order to clarify the process of degeneration, the present study aims to supplement pathologic electromyographic criteria and to specify clinical sequellae at the non ambultory stage of D.M.D. patients. Various investigations have emphasized the signs of anatomical deterioration of muscle by the appearance of the limbs and by histologic observations with light, electron and scanning microscopy. Efforts are being undertaken to relate observations to the genetic history and stages of each specific type of muscular dystrophy as described by Walton[1] and Serratrice[2]. Basic observations reported by Dubowitz and his group[3] pointed out the need for further muscle biopsy to search for pathognomic morphologic changes in each type of myopathy and in each stage of myopathy. A muscle cell surface membrane abnormality[4], defects of the sarcolemma (even in nonnecrotic fibers)[5] and focal disruptions of the plasmalemma can be associated with subjacent sarcoplasmic abnormalities[6,7]. Using electromyography, observations of low potentials during rest[8] and twitch tensions abnormally prolonged[9], gave evidence of the morphophysiological anomaly of muscle fibers in D.M.D. Reduced amplitude of muscle action potential in maximal contraction, high frequency and brief duration of isolated motor unit action potentials have been reported to be the consistent clinical electromyographic signs of myopathies [10 11 12]. A rich electromyographic interferential pattern of low amplitude, usually seen in myopathy, is quickly obtained even if no resistance is applied [13]. This interferential pattern of muscle activity is usually serrated and superimposed on brief peak potentials [13]. In other studies, reduction of Ca + + accumulation by the sarcoplasmic reticulum has been seen in D.M.D. [14]. In those patients, a considerable fibre to fibre variation in response to caffeine was reported. This may indicate that the abnormality of the sarcoplasmic reticulum, like that of the contractile proteins, is not expressed uniformly within the muscle [15]. In D.M.D. there are "weak" muscle fibers showing inability to develop control level tension and this may be related to specific extensive alterations in the muscle contractile proteins [15]. Circulatory defects [16] as well as nervous defects [9] were also demonstrated in D.M.D. and must be taken into consideration in the interpretation of signs and sequellae of this disease. One aspect of this study has been the application of a biofeedback electromyographic method developed [17] and utilized by the author (Simard) for more than a decade to aid the evaluation of fine motor control over the voluntary activity of isolated single motor units in patients. With this method, the isolated discharge characteristics can be further analysed by using a newly developed technique based on multiterritorial muscle detection [17]. This can be achieved with the use of a non painful quadribipolar electrode [17]. The necessity of multiterritorial evaluation recommended earlier [18] was followed in this study. With muscular dystrophy, not only qualitative but also quantitative evaluations are necessary to establish critaria of this specific type of myopathic lesion [19].

## SUMMARY

This work reports degenerative characteristics seen in patients at the non ambulant stage of Duchenne muscular dystrophy. A multiterritorial electromyographic evaluation of the finest motor response of the neuromuscular system was carried out in conjunction with an individual clinical rehabilitation program. Severe anomalies were observed in action potentials of isolated motor unit, in potentials during maximal effort of muscle contraction, as well as during deep inspiration. Degree of muscle activity evoked by the agonist function of deltoid, abductor pollicis brevis and sternocleidomastoid was related to the value of vital capacity and range of pertinent segmental movements. Serrated action potentials, fluctuation of internal deflexions and high frequency of isolated discharges were frequently seen. Unusual synchronic action potentials

and variation of their duration were demonstrated in a few cases. No mode of pattern of territorial muscular degeneration was found. However, a marked reduction in the rectified and filtered EMG intensity could be related in some cases to the limit of pertinent segmental movements and to the individual degree of reduced vital capacity. A follow-up investigation of earlier well classified clinical stages of this disease is required to understand more clearly the mechanism of degeneration and to find a reliable method of evaluating the individual prognosis.

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# AN ELECTROMYOGRAPHICAL STUDY ON LOOSE SHOULDER

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## INTRODUCTION

Among adolescents and young adults one can find, not so infrequently, those cases in which inferior subluxation of the shoulder joint can be easily resulted by passive downward traction of the arm. This is so-called "loose shoulder" which is not caused by paralysis of the muscles. Some of these cases complain of dull pain along the arm and shoulder, pain at motion, instability and recurrent and/or voluntary dislocation of the shoulder joint.

Our successful treatment, which was performed on twenty-six joints of twenty-three cases, is transfer of the pectoralis major muscle to the inferior angle of the scapula. The purpose of this paper is to discuss on pathomechanism of the loose shoulder and usefulness of this treatment from the viewpoint of EMG and motion analysis.

## CLINICAL MATERIALS AND EXPERIMENTAL METHODS

Using ten patients with loose shoulder motion analysis of the arm and the scapula, electromyography and roentgenography were performed before and after surgery.

With bipolar fine-wire electrodes action potentials were picked up from the pectoralis major, supraspinatus, infraspinatus, trapezius (upper fibers), rhomboideus and deltoideus (middle fibers) muscles.

On the other hand, degree of elevation and depression of the arm and spinohumeral angle were measured by electrogoniometers.

EMG and change of these angles were simultaneously recorded on ink-writing oscillograph papers.

Tilting of the glenoid was measured on roentgen pictures taken in several positions of arm elevation. On the stress roentgenogram of both shoulders taken with a five kilogram weight suspended from each wrist, the degree of laxity was evaluated. The above pictures ere taken before and after surgery.

## RESULTS

1) Inferior subluxation of the glenohumeral joint by passive traction and voluntary anterior dislocation could be prevented by the patient's own effort to elevate the scapula or by passively holding the scapula in abduction position.
2) By passive downward traction of the arm the supraspinatus and the trapezius were markedly activated. On active elevation of the scapula the trapezius was more active than the supraspinatus.
3) In most cases intensive action potential appeared in the pectoralis major muscle at the moment of voluntary dislocation. On the contrary, at the moment of voluntary reduction its EMG discharge reduced. The trapezius muscle showed a completely reverse attitude to that of the pectoralis major.
4) After surgery recurrent and voluntary dislocation disappeared completely and degree of loosening of the glenohumeral joint decreased markedly.
5) During arm elevation abduction movement of the scapula on the non-operated side was preceded by that on the operated side.
6) In most cases the transferred pectoralis major muscle was active during elevation and depression of the arm.
7) In spite of absence of paralysis of the serratus anterior muscle, in some cases, posterior displacement of the scapula during depression of the arm was observed before surgery. However, this phenomenon disappeared completely after surgery.

## DISCUSSION

As Basmajian mentioned, the supraspinatus muscle is considered to prevent inferior subluxation or dislocation of the glenohumeral joint in the normal subject.

In the patient with loose shoulder sometimes shows downward tilting of the glenoid. In the erect position, therefore, the humerus takes a relative abduction position to the scapula. In this position the suspension mechanism of the supraspinatus muscle cannot operate sufficiently. In such cases the glenoid is made to turn somewhat upwards by transfer of the pectoralis major muscle to the inferior angle of the scapula. Consequently, the humerus is brought to the adduction position and the suspension mechanism can operate well.

On the other hand, the pectoralis major muscle seems to play an important role to produce voluntary anterior dislocation. Which is based on loose shoulder. Therefore, transfer of this muscle is reasonable, because dislocation force becomes weaker and abduction force of the scapula against dislocation is increased by this prodecure.

## CONCLUSION

By electromyography and motion analysis of the scapula pathomechanism of loose shoulder without paralysis was studied. It was confirmed that adduction position of the scapula commonly observed in this disease obstructed arm suspension mechanism of the supraspinatus muscle, and the pectoralis major muscle played an important role to produce voluntary anterior dislocation of the shoulder joint which was based on the loose shoulder.

Therefore, it was demonstrated that transfer of the pectoralis muscle to the inferior angle of the scapula, which was performed on twenty-six joints of twenty-three patients, was reasonable.

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NOTES

# LOCOMOTION I

## Session 6A

STATE ESTIMATION FROM MULTI-CAMERA LANDMARK OBSERVATIONS IN 3-D RECONSTRUCTION OF LOCOMOTION PATTERNS

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## INTRODUCTION

The investigation of human locomotion in its full 3-D complexity suffers from a number of methodological and practical drawbacks, including: quantization noise, data unreliability caused by shadowing effects, unstability or even unidentifiability of corporal landmarks with respect to the corporal structure, and the sensitivity of estimated limb segment translations and rotations to such errors. By consequence, estimated forces, turning moments, and energy flows in subsequent kinetic analysis suffer not only from limited knowledge of mass distribution parameters in the human body, but also from errors in the kinematic input data. Moreover, commonly used filtering, smoothing, and differentiation algorithms do either not allow temporary data loss or employ heuristic interpolation techniques in order to create suitably contiguous data sequences: especially frequency-domain methods exhibit these properties, and the complex influence of camera positioning and observation distances on reconstructed rigid-body kinematics confounds the precision issue even more (Woltring, 1979).

The purpose of the present paper is to present an outline of time-domain modeling in the estimation of 3-D kinematics; here, a jointly deterministic and stochastic model of the movement pattern is recursively updated from new, incoming information, and this in a well-defined, optimal fashion. At the expense of a rather heavy computational burden, such modeling allows to exploit all incoming information on, e.g., observed landmarks for direct estimation of translational and rotational variables, their temporal derivatives -- all of which are required for subsequent kinetic analysis -- and the covariance matrices of all variables for precision evaluation.

## TIME DOMAIN DESCRIPTION OF A MOVEMENT PATTERN

Such time-domain modeling evolves around the notion of the *state* of the process under study, i.e., the behaviour of a combination of those current aspects of the process which, together with future process inputs, completely specify future behaviour of the process, or are assumed to do so. An example of a *state vector* are position and velocity at any time instant, with acceleration or force acting as input variables.

For a sufficiently differentiable signal $x(t)$, a value $\bar{x}(t>t_0)$ may be predicted from estimates $\hat{x}(t_0)$, $\hat{x}^{(1)}(t_0)$, $\hat{x}^{(2)}(t_0)$, ... , $\hat{x}^{(k)}(t_0)$ of the signal and its derivatives up to the k-th order at time instant $t_0$,

$$\bar{x}(t) = \hat{x}(t_0) + \hat{x}^{(1)}(t_0).(t-t_0) + \ldots + \hat{x}^{(k)}(t_0).(t-t_0)^k$$

By analytically differentiating this series, and adding an as yet unspecified truncation error term to this model, the following *state-space* description is obtained:

$$\bar{\underline{x}}(t) = \Phi_k(t-t_0).\hat{\underline{x}}(t_0) + \underline{v}$$

Here, $\underline{x}$ denotes the vector-valued state comprised of the signal and its derivatives up to the k-th order, $\Phi$ the *transition matrix* with elements $\phi_{ij} = 0$ for $i>j$, $\phi_{ii} = 1$, and $\phi_{ij} = (t-t_0)^{j-i}/(j-i)!$ for $j>i$, and $\underline{v}$ serves as the *input* to the model. In the case of locomotion studies from camera observations, this input is unknown, and one might model it as a noise term with known or postulated statistics, typically, mean value $E(\underline{v})$ and covariance matrix $V$. Assuming that the initial value $\hat{\underline{x}}(t_0)$ is known, with known covariance matrix $\hat{P}(t_0)$, the predicted state and its covariance matrix follow as:

$$\bar{\underline{x}}(t) = \Phi_k(t-t_0).\hat{\underline{x}}(t_0) + E(\underline{v})$$

$$\bar{P}(t) = \Phi_k(t-t_0).\hat{P}(t_0).\Phi_k^T(t-t_0) + V$$

with $\hat{x}$ and $\underline{v}$ assumed uncorrelated. This implies that, for time instants between observations on the movement process, the state may be predicted from the past, which incorporates (part of) the interpolation problem referred to above. In the more complex case of 3-D rigid body kinematics, the state vector may be defined as a second order model in all 6 degrees of freedom, so that $\underline{x}$ becomes an 18-elements vector, and $\Phi$ an 18x18 upper triangular matrix. Usually, one will be interested in angular velocities and accelerations, rather than in derivatives of Euler angles; this may be incorporated in the state model, and will result in a transition matrix which is also a function of $\hat{\underline{x}}(t_0)$.

## KALMAN FILTERING, ADAPTIVE FILTERING, SMOOTHING

In updating the state vector by new incoming information, the *Kalman Filter* (Gelb, 1974) assumes that measurements are made at known time instants, of linear combinations of the state vector elements, contaminated with additive noise with known mean value (usually $\underline{0}$) and covariance matrix R:

$$\underline{z}(t) = H.\underline{x}(t) + \underline{w}$$

Assuming that $\underline{v}$ and $\underline{w}$ are uncorrelated, these measurements may be optimally combined in a minimum variance sense by means of the following Kalman Filter equations:

$\underline{y}(t) = \underline{z}(t) - h.\bar{\underline{x}}(t)$, the *innovations* process

$K(t) = \bar{P}(t).H^T.(H.\bar{P}(t).H^T + R)^{-1}$ the *Kalman Gain*

$\hat{\underline{x}}(t) = \bar{\underline{x}}(t) + K(t).\underline{y}(t)$ the new, *filtered* state

$\hat{P}(t) = \bar{P}(t) - K(t).H.\bar{P}(t)$ the covariance matrix of $\hat{\underline{x}}(t)$

The filtered values $\hat{\underline{x}}(t)$ and $\hat{P}(t)$ than serve as new starting values for prediction to the next observation moment, until all data have been processed.

For example, if $z(t)$ is a noisy observation of a position, and $\underline{x}(t)$ a state vector containing position, velocity, and acceleration, the off-diagonal elements in $\bar{P}(t)$ will be nonzero by virtue of the triangular form of $\Phi_2$; this correlation between the predicted state elements will result in a filtering correction not only to the position term in the state vector, but also to the two other components. For nonlinear measurement equations, such as camera-observations contaminated with additive quantization noise, these equations may be linearized about the predicted state vector $\bar{\underline{x}}(t)$: this is the so-called *Extended Kalman Filter* (Gelb, 1974). The only requirement would be that the observed landmarks have known coordinates with respect to a frame of reference defined in terms of the modeled limb segment, and even these coordinates might be carried as estimable parameters in the filtering procedure.

The initial values required for starting a Kalman Filter may be obtained in a number of ways; for example, one might use a zero state vector in combination with a large covariance matrix; after a few filtering iterations, both the state and its covariance matrix will converge to realistic values. The properties of the measurement noise may be experimentally assessed, for instance, by observing a stationary process, or by considering the type of error sources most likely involved. Quantization noise, for example, may often be modeled as uncorrelated noise with standard deviation of $1/\sqrt{12}$ L.S.B. However, the properties of the process noise $\underline{v}$ are not so easy to establish.

In *adaptive filtering*, the mean and covariance matrix of $\underline{v}$ may themselves be continuously estimated by reference to the innovations sequence $\underline{y}(t)$; if the statistical properties of $\underline{y}(t)$ appear inconsistent with the known properties of the measurement noise $\underline{w}$, this discrepancy may be used to update estimates of $E(\underline{v})$ and V.

Up to this point, current estimates followed from current observations and past estimates; in a non-real-time context, also observations at later time-instants may be used. For such *smoothing* procedures, one might have two Kalman filters operating, one forward, and one in reverse time, and combine the two filter outputs in an optimal sense (Gelb, 1974). Other, numerically more efficient techniques are available.

## DISCUSSION

Contemporary, recursive optimal state estimation procedures based on time-domain modeling offer a universal means to process multisensor data in locomotion studies. Until the present time, they have been most extensively applied in aerospace navigation studies; these processes and the measurement systems employed in their investigation have a formal structure similar to human locomotion as measured by cameras, gioniometers, or accelerometers. Major problems are the heavy numerical burden, and the possibility of filter divergence if data loss gaps become too large or if the model is too simple an approximation of the real world.

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## ACKNOWLEDGEMENTS

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# EMG ACTIVITY IN TREADMILL AND OVERGROUND WALKING

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## INTRODUCTION

The treadmill has been utilized as a convenient means of training and evaluation. The efficacy of direct data extrapolation from treadmill to overground walking conditions is of the utmost concern for the development of exercise prescription for rehabilitation patients and as a stress device in muscle metabolic research.

The existing literature reveals anomalous conclusions regarding the accuracy of data extrapolation from treadmill to overground conditions. In terms of metabolic cost, temporal factors and mechanical energies, a number of authors (Daniels et al., 1953; Custance et al., 1970; Nelson et al., 1972) have obtained differences in the two experimental conditions. On the contrary, Ralston, 1960 and McMiken, 1976 have observed no metabolic differences.

## Procedures and Data Analysis

Six male university students with no known pathological disorders were chosen for investigation. The subjects underwent six 10-minute orientation sessions to familiarize them with treadmill walking. During actual data collection the surface EMG activity of five major muscle groups was examined: tibialis anterior, soleus, quadriceps and biceps femoris.

Each subject performed a maximal isometric voluntary contraction for each of the muscle groups examined. Following this the subjects were randomly assigned to either the level ground walking or the level treadmill condition. The raw EMG activity of the muscles was recorded via a multichannel biotelemetry system over a period of ten strides, while the subjects walked at their natural cadence.

The full wave rectified EMG was passed through a low pass filter with a cut-off frequency of 6 Hz to obtain the linear envelope. This was digitally processed to obtain the average EMG phasic activity over 10 strides. The average EMG per stride, expressed as a percentage of the EMG of the maximum voluntary contraction, was calculated for ten strides. The mean of these EMG percentages and their standard deviations are tabulated in Table 1. A t-test of significance was completed on each muscle for the two conditions and the significance level is shown in Table 1.

## Results and Discussion

For three of the four muscles tested (tibialis anterior, soleus and biceps femoris) the EMG activity was significantly different ($p<.001$) for all six subjects in the two walking conditions. For the quadriceps muscle, four of the six subjects showed significantly higher activity ($p<.001$) on the treadmill, for one subject it was lower ($p<.05$) and for the sixth subject there was no significant difference. Figures 1 through 4 indicate the phasic activity of the tibialis anterior and the quadriceps femoris muscles under each of the experimental conditions, for two of the subjects. The plots show the average of ten strides beginning and ending with heel contact; the vertical lines indicate one standard deviation either side of the mean.

In Table 1 it is shown that the EMG activity during treadmill gait was greater than for overground. A look at the EMG patterns gives us some insight about the differences. The activity of the rectus femoris during treadmill gait (Fig. 1) has larger and greater activity over the entire stride than the overground trial (Fig. 2). The increased activity during the first half of stance suggests that he locked his knee and let the treadmill carry his leg beneath him. During late stance and early swing the extra hip flexion activity appears to compensate for the need to accelerate the thigh forward from its reverse velocity condition. Figures 3 and 4 indicate that the tibialis anterior is more active and for a longer period of time following heel contact on the treadmill. This may be a result of the muscle pulling the shank forward over the foot. The increased activity during the swing may be an overcompensation to clear the ground.

## Conclusions

It is evident neurologically that treadmill gait is quite different from overground walking. These differences may result from energy transfers between the subject and the treadmill motor; this was reinforced by the observed variations of treadmill motor current during the stride, which was about 8%. In addition, the velocity of the treadmill belt was found to fluctuate 10% over the stride.

Table 1

| Subject | Trial | $M_1$ | $M_2$ | $M_3$ | $M_4$ |
|---|---|---|---|---|---|
| 1 | T | 48.1 | 116.7 | 149.2 | 44.0 |
| | | (2.9) | (24.9) | (10.1) | (2.8) |
| | | *** | ** | * | *** |
| | O | 20.2 | 141.7 | 181.6 | 18.0 |
| | | (2.2) | (23.6) | (18.4) | (2.9) |
| 2 | T | 33.0 | 46.5 | 16.1 | 90.5 |
| | | (3.1) | (4.5) | (2.1) | (10.0) |
| | | *** | *** | *** | *** |
| | O | 18.1 | 19.4 | 6.9 | 29.0 |
| | | (3.1) | (3.0) | (.79) | (1.5) |
| 3 | T | 62.6 | 133.4 | 48.2 | 59.4 |
| | | (6.4) | (8.0) | (7.2) | (3.6) |
| | | *** | *** | *** | *** |
| | O | 35.2 | 60.9 | 23.5 | 16.6 |
| | | (10.9) | (16.9) | (10.3) | (4.0) |
| 4 | T | 15.9 | 15.1 | 3.1 | 13.6 |
| | | (4.3) | (1.2) | (.5) | (1.7) |
| | | NS | *** | NS | *** |
| | O | 16.3 | 19.7 | 3.1 | 6.6 |
| | | (5.4) | (2.6) | (.7) | (2.7) |
| 5 | T | 242.4 | 49.1 | 23.3 | 38.3 |
| | | (20.5) | (2.4) | (1.2) | (1.7) |
| | | *** | *** | *** | *** |
| | O | 133.1 | 19.6 | 6.9 | 12.6 |
| | | (23.7) | (3.9) | (.9) | (1.7) |
| 6 | T | 68.4 | 33.6 | 28.6 | 59.8 |
| | | (6.2) | (4.2) | (2.5) | (5.6) |
| | | *** | *** | *** | *** |
| | O | 38.9 | 19.1 | 13.1 | 41.7 |
| | | (2.6) | (4.8) | (1.2) | (3.3) |

*** $p<.001$
** $p<.025$
* $p<.05$
NS - not significant

O - Overground Trial
T - Treadmill Trial
$M_1$ - Tibialis Anterior
$M_2$ - Soleus
$M_3$ - Quadriceps
$M_4$ - Biceps Femoris

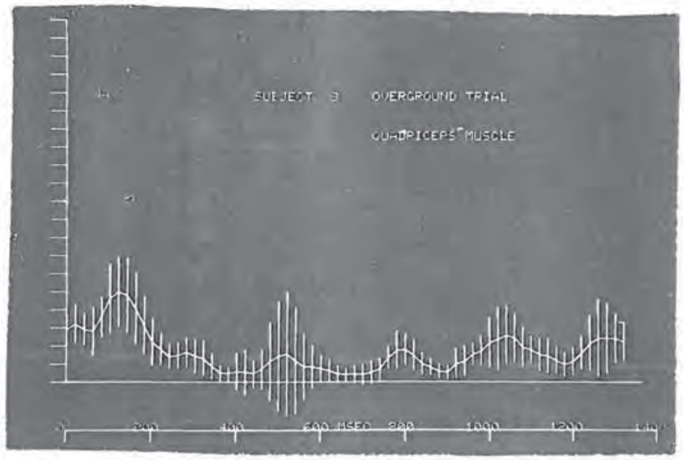


Figure 2

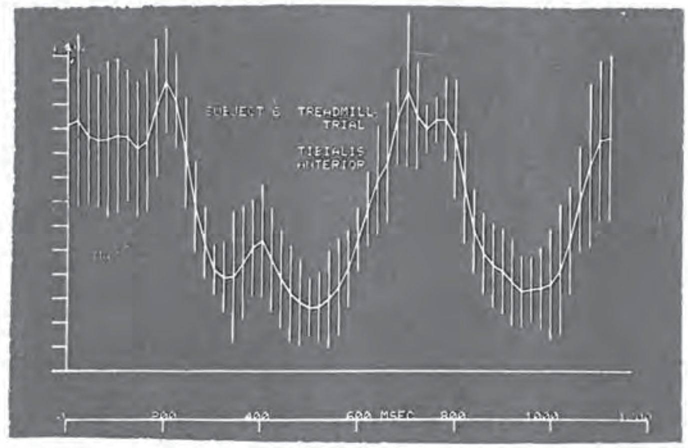


Figure 3

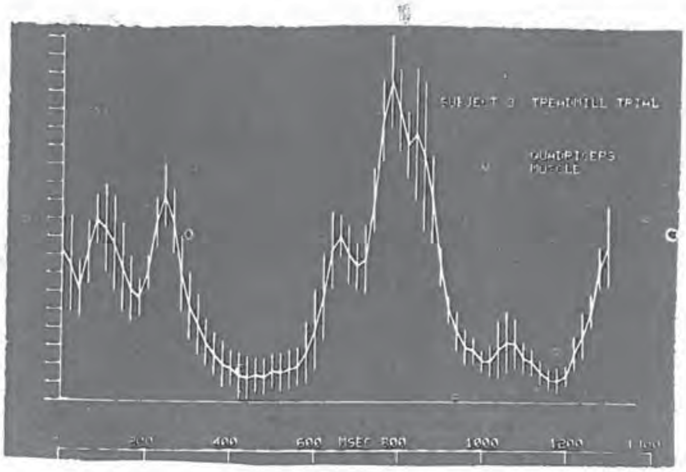


Figure 1

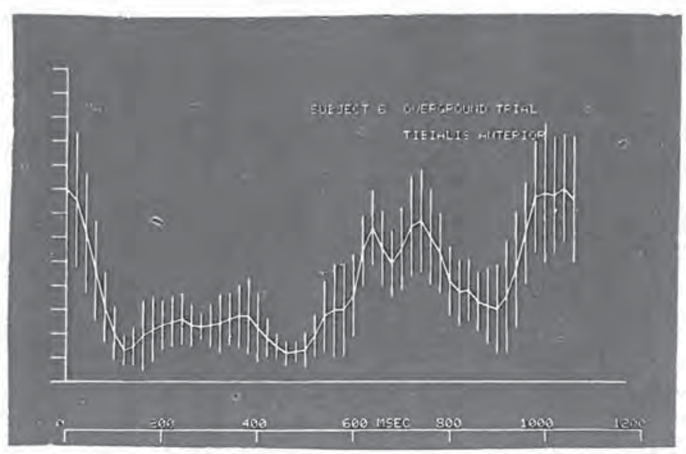


Figure 4

AUTOMATIC 3-D GAIT ANALYSIS USING A SELSPOT BASED SYSTEM

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## INTRODUCTION

A system has recently been developed to automatically acquire kinematic data on moving humans and process this to yield position, orientation, force and moment information for up to ten body segments of interest. The raw data acquisition is handled by a Selspot system marketed by Selcom AB of Sweden. This uses a lateral-photo-effect diode sensitive to infrared light as a sensor of position of the image of an infrared source in the image plane of a camera. The Selspot system has the capability of monitoring thirty infrared LEDs by time multiplexing their duty cycles at 315 Hz each. Two cameras are positioned accurately, and their image position data are manipulated to yield three-dimensional points by a PDP 11/40. The 3-D points associated with a rigid array are identified, and if three or more points are in an array, the system computes that body segment's orientation. These data are available in real-time at 12 Hz or in off-line storage at up to 315 Hz. In the off-line, post processing mode low-pass data filtering are available, and several kinematic data display and plotting routines can be exercised. Currently the dynamic resolution and accuracy of the system are in the range of 3 to 5 mm in position and 1 degree in orientation at 2.5 m range. The IR LEDs are roughly 1 by 2 by 3 cm in size, and an array of 3 on a plexiglas frame is not obtrusive to a subject, and is easily attached with elastic straps. The quality of the data is not affected by ambient visible light; thus subjects can walk in a room illuminated normally. No umbilical cords are needed; the LEDs can be powered with a battery pack. At present the viewing volume is approximately a 2 m cube. There are plans to enlarge this.

After an off-line data experiment has been conducted and the data are fully manipulated, it can then be sent to a dynamics routine. This incorporates the kinematics, the inertial properties of each link and Newton's equations to produce estimates of the inertial forces and moments that must have impinged on the link for the observed motion to have occurred. The first step is to obtain velocities and accelerations from the position and orientation data. This is done using a 5-point Lagragian interpolation formula for equally spaced absicissas. Each derivative is low-pass filtered at the same frequency as the kinematic data using a sixth-order Butterworth filter. The geometry is the last data needed before the force estimator can run. This information, essentially the vectors from the center of mass of a limb segment (the origin of the body coordinate system) to the ends of the segment, is generated using a scheme to find the instant axes of rotation between two segments. A point on the axis inside the joint is chosen as the segment end. This eliminates the subjective selection of joint centers by utilizing calculated instant axes of rotation. These points are taken to be the points of application of the resultant inertial forces and moments to the link, determined by the force estimator. The technique is general enough to accept linkages of any complexity and connectivity; however, only ten links can be monitored. For an open assemblage with ten or fewer links, the system provides complete dynamics. In assemblages of more than ten, the human body for example, a reference force and moment must be available somewhere along the linkage, most easily obtained in gait analysis with a force plate.

The force estimator has been demonstated to perform correctly with ideal data and has shown very favourable results with actual kinematic data. Experiments were performed using a 2-degree-of-freedom pendulum of known inertial properties. Newton's laws were applied with an initial offset to obtain equations of motion and resulting forces. These were compared to the actual kinematic data with the same initial conditions and the results from the force estimator. The Newtonian model yielded oscillating horizontal forces in the plane of motion with an amplitude of 0.25 newtons. The force estimator produced forces in the same direction with an amplitude of 0.27 nt. In the vertical direction Newton predicted a 0.05 oscillation around 2.706 nt. 2.706 nt is the weight of the pendulum. The force estimator yielded a similar 2.7 nt mean, and a 0.05 nt oscillation.

The estimated forces from the actual kinematic data taken from the physical pendulum are not as smooth as the sinusoids generated from the Newtonian equations of motion. The discrepancy is due to noise interjected by the Selspot system. However, the magnitude of the noise is small compared to the amplitude of the forces. Digital filtering attenuates this but does not elimate all of the spurious signals. Position noise is amplified when the data are differentiated. Comparison of the quations of motion and the actual data show this to be the sole source of error. For human-sized gait frequency phenomena, this error is a high frequency component superimposed on the resultant forces with an amplitude of no more than 10 per cent of the true value. This requires intelligent selection of low-pass cut-off frequencies for the data filtering. Processing time for a single moving segment sampled at 62 Hz for 5 seconds is roughly 15 minutes--half due to the kinematics, and half to the dynamics.

This linkage analysis technique represents a considerable improvement over conventional gait analysis schemes in speed, objectivity and accuracy. Movie techniques require digitization of the data frame by frame, a laborious process taking days to complete. Often a physical therapist does the digitization in order to identify bony prominences as indicators of joint centers. These difficulties are eliminated by a scheme that takes no more than 15 minutes per body segment, and automatically finds the instant axes of joint rotation. The absolute accuracy of the linkage analysis system has been assessed, and the errors have been found to be due to sampling noise. This noise is small, and its effect known. There will be additional errors in gait analysis due to movement of the LED arrays with respect to the skeleton and small changes in the inertial properties of the limb segments during gait. But these are inescapable for noninvasive gait analysis, and this new technique eliminates all of the other sources of error common to conventional gait analysis.

## ACKNOWLEDGMENTS

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MUSCLE FORCES DURING WALKING: A MINIMUM ENERGY APPROACH

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## INTRODUCTION

Mechanical analysis is applied to human gait at several levels ranging from kinematic description of the lower limbs to detailled descriptions of skeletal force distributions. The precision with which the later can be accomplished is limited, however, by inexact knowledge of individual muscle forces during walking cycles. Mechanical redundancy, which precludes direct solution for these forces, is responsible for this uncertainty. This redundancy arises because the number of anatomically (and therefore neurophysiologically) distinct muscles exceeds the degrees of freedom of the skeletal links by more than three times.

## MINIMUM ENERGY FORMULATION

This redundancy can be resolved by using mathematical optimization, where a unique set of muscle forces necessary for dynamic equilibrium (out of an infinity of possibilities) is chosen on the basis of a performance or cost criterion. Thus the biomechanical problem becomes one of selecting an appropriate optimal criterion. Taking the view that the natural system eventually learns to use this redundant system in an optimal manner itself, a physiologically rationalized criterion should be chosen. This leads to the concept of minimizing the total energy consumption of the muscles during walking.

For this investigation the system under consideration consisted of four skeletal segments (the pelvis, femur, tibia, and foot) with 7 degrees of freedom (3 at the hip, 1 at the knee and 3 at the ankle) and 31 muscles spanning these joints. Thus equilibrium at the joints can be described by

$$R\,\underline{f} = \underline{m} \qquad (1)$$

where $R$= 31x7 matrix of muscle moments arm

$\underline{f}$= vector of 31 muscle forces

$\underline{m}$= vector of 7 joint moments

and the minimum energy criterion can be expressed

$$\min_{T} \int \sum_i e_i \, dt \qquad (2)$$

$e_i$= instantaneous power requirement of muscle i

T= time for one walking cycle

A model for chemical energy consumption rate of chemical power was developed[1] by combining empirical thermostatic studies of muscle with recently developed sarcomere simulation models to obtain an expression for the chemical power:

$$e = \ell f \, g(v/v_{max}) \qquad (3)$$

$\ell$= rest length of muscle

f= muscle force

and $g(v/v_{max})$ is shown in Figure 1.

The absence of any dynamic terms in this expression comes from considering the low frequency use of muscle during walking and also from considering the tendons as infinitely stiff.

Noting that (3) is linear in force and has no dynamic terms, linear programming can be applied as an optimization method to solve equations (1) and (2). This was accomplished by breaking the walking cycle up into discrete intervals and treating each time step as a separate linear program. Thus a complete solution can be had by combining these individual solutions in sequence.

## EXPERIMENTS

Essential to the solution of equation (1) are: 1) the muscle moment arems (matrix R), and 2) the dynamic joint moments (vector $\underline{m}$). The former were obtained by first taking approximate muscle measurements on preserved skeletons. Then, using appropriate transformations, the resulting three dimensional moment arms were modified according to the measured kinematic state of the limbs.

These kinematic measurements were performed with the TRACK system (Telemetered Real - time Acquisition and Computation of Kinematics) developed by Conati[2] at M.I.T. This system combines the SELSPOT (trademark of SELCOM A.B. Sweden) point location device with a PDP 11/40 minicomputer to provide complete three dimensional kinematic information about the limbs. This kinematic information was also used in a dynamic model of the lower limbs[3] that calculates the net joint moments given the mass properties of the limbs.

In addition to these measurements, myoelectric signals (MES's) were recorded from ten surface muscles for comparison with the force optimization prediction.

## RESULTS

The above experiments were performed on two subjects walking at normal speeds. The minimum muscle energy optimization was applied to the resulting data and the force predictions compared to the MES's. Typical findings are shown in Figure 2. The two force results shown were chosen as the best and worst cases (for a single individual) when compared to the MES's. In general the predicted force - MES agreement was good for hip muscles and deteriorated as more distal muscles were considered. Also evident was a complete lack of antagonist activity in the predicted forces at the knee (which was considered to have only one degree of freedom) despite indication of this activity in the MES's.

The sensitivity of the result to the optimal criterion was also investigated by varying parameters within the minimum energy formulation and by using a minimum force criterion. The results indicate that the temporal aspects of the solution are almost unchanged regardless of the criterion choice. However, the muscle force magnitudes are a strong function of the criterion. Thus it appears that the basic (temporal) pattern of muscle use in most strongly determined by the moment requirments and the choice of muscles to provide those moments is quite limited. This suggests that the system is much less redundant than anticipated.

## CONCLUSIONS

The minimum muscle energy optimization approach has been shown to predict muscle forces in reasonable correspondence with myoelectric measurements. The choice of this criterion, however, appears to have minimal effect on the temporal sequencing of the muscles but does have a strong effect on the relative muscle force magnitudes within this pattern. It is clear, therefore, that temporal inaccuracies (which can be detected with myoelectric signals) will not be resolved by modification of the criterion of optimality, but rather by incorporation of addition physiologically based information in the constraints

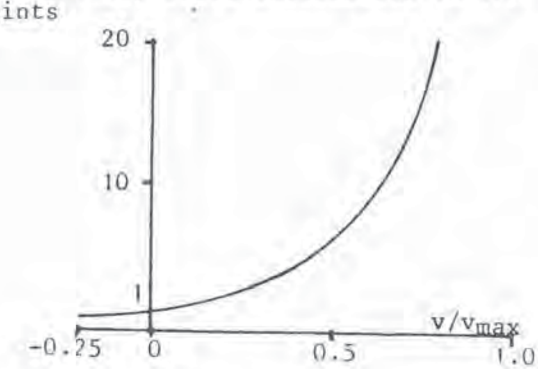


Figure 1 The function $g(v/v_{max})$

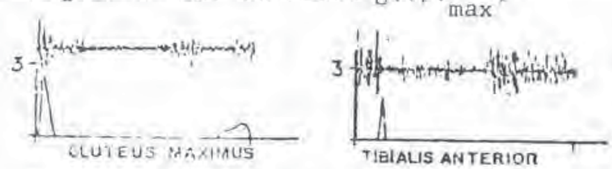


Figure 2 Typical force - MES results

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NOTES

# CLINICAL III

## Session 6B

EFFECT OF ECG CONTAMINATION ON EMG-BASED INDICES OF MUSCLE FATIGUE

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Use of the centroid of the power spectrum (2,3) or the "HI/LO" ratio (1) of the EMG as indices of muscle fatigue is complicated by ECG contamination in the case of the respiratory muscles. This has usually been dealt with by gating out portions of the EMG around the QRS complexes. High pass filtering has also been suggested, but has potential drawbacks (3), since fatigue does affect the low frequency portion of the EMG spectrum (2), which overlaps the ECG spectrum. Estimation of the centroid and the HI/LO ratio is also complicated by respiratory variation in the EMG. This study is an attempt to assess quantitatively the effects of ECG contamination by artificially adding ECG signals of known properties to EMG signals, free of ECG, obtained during periodic, fatiguing contractions of a skeletal muscle.

## Methods

Surface EMG was recorded from Biceps Brachii in a subject seated with upper arm vertical, elbow supported and forearm horizontal. The subject repeatedly lifted a 10 kg mass 10 cm above its support for 2 sec, then rested for 2 sec, and continued until unable to lift the mass. A standard precordial ECG was recorded from another subject at rest.

Analog magnetic tape records were subsequently filtered with 4-pole high and low pass filters at 5 and 500 Hz respectively, sampled at 1000 Hz, and stored digitally. Combined EMG/ECG records were created by scaling and adding in the computer. The total power and centroid of ECG, EMG and EMG/ECG signals were computed from power spectra obtained by averaging modified periodograms (4), using segments of 256 points. ECG spectra were computed from 25 sec records; EMG and EMG/ECG spectra were computed for each 2 sec biceps contraction. HI/LO ratio for each contraction was computed from the spectrum as the ratio of power in the 125-350 Hz band to power in the 20-40 Hz band.

## Results

The effect of ECG contamination on centroid and HI/LO ratio of the biceps EMG are shown in Figs. 1 and 2 for a typical case in which ECG to EMG power ratio was 1/4. Results are plotted for the first five (no fatigue) and the last five (fatigued) contractions. Addition of ECG results in a reduction of the centroid and of the HI/LO ratio in nearly all contractions. The average percent reduction is similar, both for centroid and HI/LO ratio, and for fatigued and non-fatigued contractions (7-12%).

## Discussion

The centroid can be regarded as the center of gravity of the power spectrum. The centroid of a signal which is the sum of two signals is simply related to the centroids of the component signals, if these are statistically stationary and uncorrelated. The problem is the same as that of finding the balance point of a

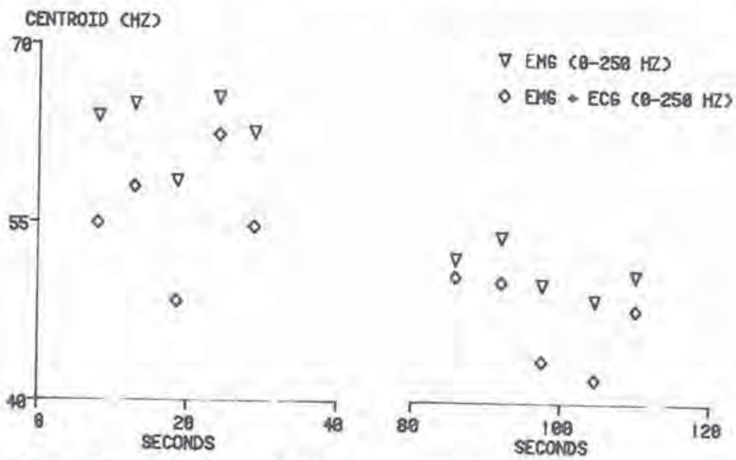


Fig. 1. Effect of ECG on EMG centroid

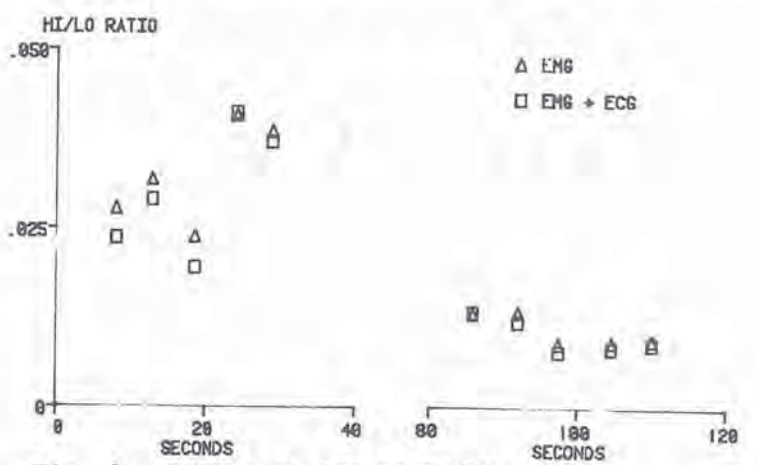


Fig. 2. Effect of ECG on HI/LO ratio

thin beam upon which rest two objects whose individual centers of gravity are known (Fig. 3). Thus

$$Ct = (CmVm + ChVh)/(Vm + Vh)$$
$$= (Cm + RCh)/(1 + R) \qquad \text{Eq. 1}$$

where Ct, Vt, Ch, Vh, Cm and Vm represent the centroids and variances (total power) of the combined signal, the ECG and the EMG respectively, and R = Vh/Vm. Note however that, due to statistical errors, this relation is only approximate unless long records are used to estimate the parameters.

Ct should increase with increase in Ch, but decrease with increase in R. The effect of a change in R should be small if R is small initially. Fortunately Ch varies little with heart rate (Table 1).

Eq. 1 suggests two methods of reducing the error introduced by ECG contamination. (1) If Vh and Ch can be estimated (e.g., during breath holding), then Cm can be calculated exactly when Eq. 1 is true. (2) Since the error is very sensitive to the ratio R, a significant reduction in error may be achieved by even partial elimination of the ECG through high pass filtering.

Fig. 4 shows the effect of the two procedures. Centroids for each contraction are shown for the raw EMG, the combined EMG/ECG signal high pass filtered at 20 Hz, and the combined signal corrected according to Eq. 1.

In this case the results obtained by using the correction formula are poor, a fact which is attributable to statistical errors in estimating Ct and Vt owing to the shortness of the contractions. It seems clear that if this method is to be used successfully then estimates must be made from a spectrum obtained by averaging over several consecutive contractions.

## Conclusion

EMG centroid and HI/LO ratio generally fall in the presence of ECG contamination. The error can be reduced substantially by high pass filtering. A further correction based on theoretical analysis appears feasible but may require averaging over longer records.

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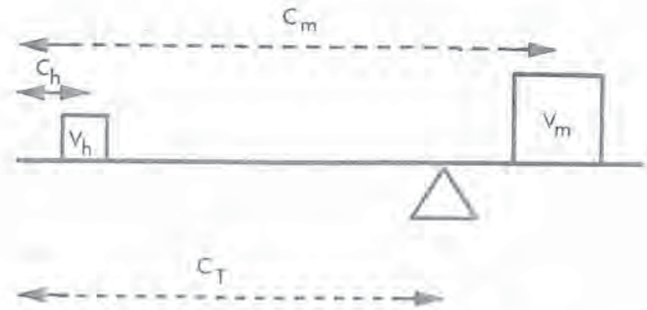


Fig. 3. Analog of centroid relationship

| Heart Rate | ECG Centroid |
|---|---|
| 65 → 130 | 17.4 → 19.4 |
| 72 → 120 | 19.4 → 20.3 |
| 74 → 92 | 16.3 → 16.7 |
| 82 → 90 | 17.4 → 17.4 |
| 60 → 88 | 22.4 → 24.0 |
| 60 → 122 | 18.3 → 20.1 |
| 70 → 130 | 19.2 → 21.1 |

Table I. Change in Ch with heart rate. The heart rate was increased with exercise (first six) and with intravenous atropin (last two).

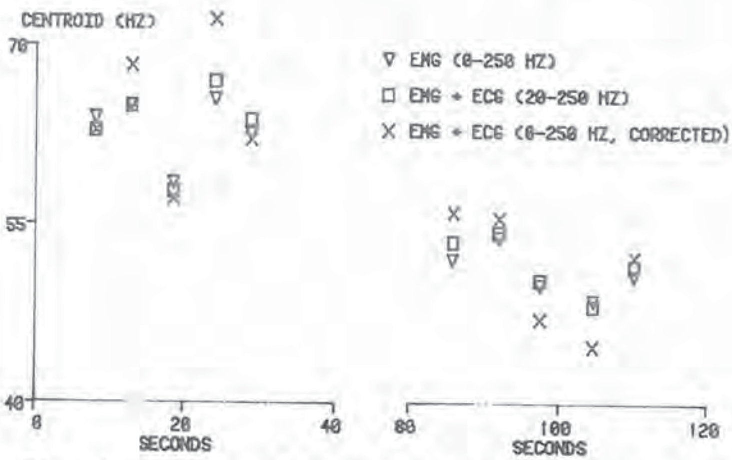


Fig. 4. Effect of correcting the centroid

# H/M RATIOS OF GASTROCNEMIUS LATERALIS AND SOLEUS MUSCLES OF MEN AND WOMEN.

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## INTRODUCTION

In many laboratory animals as well as in humans, there is a clear structural and functional differentiation between slow and fast twitch muscle fibers (Taylor, 1979). In these animals, the soleus muscle is considered as a red (slow twitch) muscle whereas the gastrocnemius muscle is considered as a pale or white (fast twitch) muscle (Levy, 1963). The soleus muscle takes a greater part in the stretch reflex contraction (Levy, 1963). In humans, the soleus muscle have a higher proportion of slow twitch fibers than its synergist, the gastrocnemius (Edgerton et al., 1975). The experiments on the soleus muscle of human subjects have demonstrated that only the slower fibres within the muscle contribute to the reflex twitch (Buchthal and Schmalbruch, 1970).

There are many reports concerning the fraction of the soleus motoneurone pool activated in the monosynaptic H-reflex. There are, however, very few reports dealing with the H reflex of the gastrocnemius muscle and none of them have specifically reported data on women. The purpose of this study was to evaluate the H/M ratio of soleus and gastrocnemius lateralis following one single shock and to compare these ratios between men and women.

## METHODOLOGY

Eight male and eight female kinanthropology students, aged 19-22 years, volunteered for the study. Subjects were made to lie prone, with the feet hanging beyond the edge of the examination table. Their take-off leg was studied.

Muscle action potentials of the soleus and gastrocnemius lateralis muscles were each picked-up by a pair of surface Ag-AgCl electrodes. They were positioned, at one centimeter from each other, over the motor point, predetermined by electrodiagnosis. The muscle action potentials were amplified by differential amplifiers. The peak-to-peak amplitude of these potentials were evaluated from the memoscope, when maximum response occurred.

Monopolar stimulation of the posterior tibial nerve in the popliteal fossa with square waves pulses of 1-ms duration was used at a rate of 0,3 cps. The cathode was positioned in the popliteal fossa where stimulation revealed a H-response in both muscles.

A 2 x 2 analysis of variance was performed to test for statistical differences.

## RESULTS

The means and standard deviations of the H/M ratios of gastrocnemius lateralis and soleus muscles of the students appear in table 1. In the soleus, this ratio was three times higher than in the gastrocnemius for both groups of subjects. It can be noted that the variability of these ratios is much greater in women than in men. This is partly due to one subject whose ratios were twice as high as the reported means. Grouping the males and females together brought the standard deviations to 0,07 for gastrocnemius muscle and to 0,15 for soleus muscle, thus increasing the homogeneity of variances. Statistical differences was found only between the means of soleus and gastrocnemius lateralis muscles ($p < .001$).

Table 1. H/M ratios of soleus and gastrocnemius lateralis muscles.

| | | Gastrocnemius | Soleus |
|---|---|---|---|
| Women | $\overline{X}$ | 0,17 | 0,42 |
| | $S_X$ | 0,09 | 0,21 |
| Men | $\overline{X}$ | 0,14 | 0,42 |
| | $S_X$ | 0,03 | 0,10 |

## DISCUSSION

The H/M ratios of the gastrocnemius lateralis reported here is similar to the one reported by Novikova (1970) for the gastrocnemius medialis ($\overline{X} = 0,123$) but somewhat lower than the one reported for the same muscle by Mongia (1972) ($\overline{X} = 0,25$).

Ginet et al. (1975) reported that the H/M ratio of the soleus muscle was somewhat lower in sedentary people ($\bar{X}$ = 0,58) than in trained people ($\bar{X}$ = 0,69). While Le Bars and Paulet (1976) showed that this ratio was not modified by a seven-month swimming program, Le Bars and Toulouse (1974) showed that this ratio was statistically different between sprinters ($\bar{X}$ = 0,34) and long distance runners ($\bar{X}$ = 0,50). The data reported here ($\bar{X}$ = 0,42) fall within this range.

As in Novikova's study (1970), the H/M ratio of the soleus was higher than the same ratio of the gastrocnemius lateralis muscle. In parallel, the proportion of slow twitch fibers is higher in the soleus than in the gastrocnemius (Edgerton et al., 1975). Reminding that Buchthal and Schmalbruch (1970) have demonstrated that, at least for the human soleus muscle, the H-reflex recruited only slower muscle fibers, one might be tempted to infer that the H/M ratio gives an indication of the proportion of slow twitch muscle fibers in a mixed squelettal muscle.

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EXCITABILITY OF THE MONOSYNAPTIC REFLEXES WITH TOPICAL ANESTHESIA

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## INTRODUCTION

Skin receptors have been shown to affect the motoneuron discharge in the flexor reflexes (Hagbarth and Finer, 1963). However, electrical nociceptive stimuli are not selective for a specific receptor. Sabbahi and Sedgwick (1976) demonstrated inhibition of the H-reflex during natural stimulation of the mechanoreceptors and thermoreceptors. Other receptors should have an effect on the motoneuron discharges and this effect would be related to the skin areas (Hagbarth, 1952) or dermatomes stimulated. In this work H-reflex and Achilles tendon reflexes (ATR) were studied for the effect of topical anesthesia applied to different skin areas and dermatomes.

## MATERIALS AND METHODS

Twenty five subjects volunteered, 13 males and 12 females, whose ages ranged from 18 - 53 years. They were tested for the changes in the H-reflex and ATR from prone position. The H-reflex was recorded using the method of Sabbahi (1976). The posterior tibial nerve in the popliteal fossa was stimulated unifocally with one millisecond pulses every five seconds and the soleus muscle action potentials were recorded with superficial disc electrodes. For the recovery curves, two identical stimuli with varied interstimulus intervals were used. The ATR was elicited by an electrically activated solenoid plunger which produced strikes of equal force to the tendon while the reflex response was recorded every 10 seconds via the same electrodes. Averages of not less than 10 consecutive sweeps were calculated.

Topical anesthetic (20% Benzocaine) was sprayed on the appropriate skin areas or dermatomes for 10-20 seconds and recordings were taken at time intervals for a total of 40 minutes. Anesthetic was applied to the skin overlying the calf, tibial, quadriceps and hamstrings in 12 subjects and to $L_2$, $L_3$, $L_4$, $L_5$, $S_1$, and $S_2$ dermatomes in 10 subjects. Skin of the whole lower limb was sprayed with anesthetic in other three subjects. The H-reflex (amplitude and recovery curve) and the ATR were recorded before (control) and after the application of the anesthetic. Peak-to-peak measurement of the reflexes amplitude and H-reflex recovery up to 100 msec. were recorded.

## RESULTS

The H-reflex was significantly increased in amplitude after application of the anesthetic to the skin overlying the calf, hamstrings, and quadriceps muscles (fig. (1)). This reflex facilitation continued to increase for the duration of the experiment. Thirty minutes after the anesthetic was applied to the calf skin the amplitude of the H-reflex increased by 76%; when applied over the hamstring muscles, by 200%; and when applied over the quadriceps muscles, by 167%.

After application of the anesthetic to the skin overlying the anterior tibial muscles, inhibition of the reflex was recorded, measuring 60% of the control after 30 minutes. This reflex inhibition was less significant in the first 15 minutes post anesthesia.

When the anesthetic was applied to the dermatomes of the lower limb, reflex facilitation was recorded in most dermatomes. Thirty minutes after the anesthetic was applied to $L_2$ dermatome the H-reflex increased by 126%; when applied to $L_3$ dermatome, by 50%; when applied to $L_4$, by 114%; when applied to $L_5$, by 177%; when applied to $S_1$, by 22% and when applied to $S_2$, by 97%.

When the whole lower limb was sprayed with anesthetic, the reflex increased by 111% after 30 minutes. The degree of reflex facilitation varied from one subject to another, but was consistent in the same subject.

When the anesthesia was applied and the H-reflex recovery curve was plotted significant earlier and faster recovery was noticed. In all cases where the H-reflex was facilitated, earlier recovery was noticed as well. This earlier recovery continued to increase for the duration of the experiment. Thirty minutes after the application of the anesthesia recovery was 10 - 55 msec. earlier (fig. 2). Faster recovery was noticed up to 100 msec. interstimulus intervals, which indicate a higher excitability of the motoneuron pool of the calf muscle.

After anesthesia the ATR showed fluctuation betwen mild inhibition and control value. No significant changes were noticed in the M-response.

## DISCUSSION

Hagbarth and Finer (1963) noticed that the extensor muscles are excited from skin area localized on the muscle itself and inhibited from other skin areas. Our results do not support their findings, but are consistent with our previous work (Sabbahi and Sedgwick 1976). In our previous work mechanoreceptors and thermoreceptors of dermatomes, innervated by the same roots as the soleus muscle, that is $L_5S_1$, were found to have inhibitory effect on the soleus H-reflex. We could not assume complete blocking of all skin receptors by topical anesthesia. However, relative blocking of those receptors in $S_1$ dermatome and skin over the anterior tibial muscle i.e. $L_4$, $S_1$ showed either no significant changes or reflex inhibition respectively. This suggests that the excitatory effect of these receptor areas is segmental. Other skin areas and dermatomes have an inhibitory effect on the soleus motoneurons, possibly via segmental and suprasegmental mechanisms (Gassel and Ott, 1973). This is confirmed by the changes in the H-reflex recovery after anesthesia.

Hagbarth and Finer (1963) as well as other workers used noxious stimuli, while in our studies natural stimuli and mild skin anesthesia were used.

The fact that skin anesthesia changes the H-reflex, that is the $\alpha$ motoneurone excitability, indicates an on-going inhibition and excitation for the soleus motoneurons during resting state.

The results of the ATR appear to indicate that some damping of the $\gamma$ system occurs when the skin area of the lower limb is anesthetized. This agrees with the observations of Urbscheit and Bishop (1970). However, more experiments are needed for a more definitive statement.

## CONCLUSION

Skin receptors were shown to have an ongoing excitation and inhibition effect on the soleus motoneurons. Skin areas and dermatomes supplied by the same root as the soleus muscle are excitatory, while other areas are inhibitory to the soleus muscle. This effect is possibly mediated via segmental and suprasegmental mechanisms.

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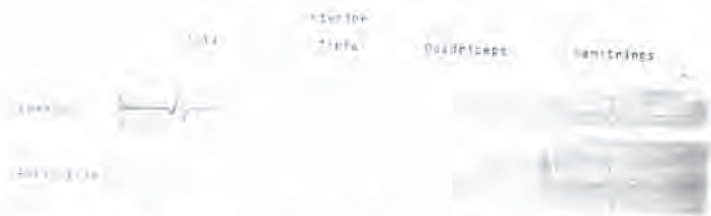

Fig. 1. H-reflex before (Control) and 30 min. after anesthesia (benzocaine) applied to different skin areas.

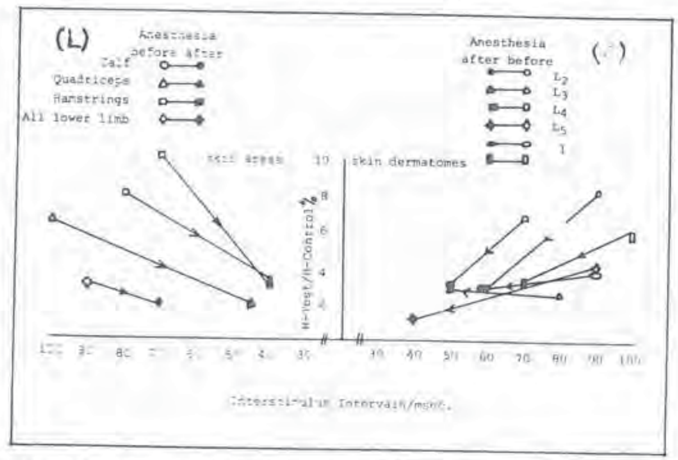


Fig. 2. Changes in the H-reflex recovery time (abcessa) and percent of recovery (Ordinate) 30 min. after anesthesia to different skin dermatomes (R) and skin areas (L).

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POST-TRAUMATIC NERVE REGENERATION ACCORDING TO TYPE OF CONSERVATIVE TREATMENT ADOPTED

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The Authors have carried out a triple blind study on patients affected by peripheral injuries of upper limb.

They have selected three homogeneous groups of patients affected by axonotmesis of the following nerves: Axillary, Radial, Median, Ulnar and Long Thoracic.

All patients received conservative treatment.

The first group underwent traditional physiokinesitherapy, i.c. electrostimulation of paralitic muscles, passive and active exercises, neuromuscular proprioceptive facilitations; the treatment of the second group consisted in the above mentioned therapy and in electromyographic biofeedback. (The patients begun activating unaffected muscle of the opposite limb, observing the potential evoked on the oscilloscope and hearing the noise produced by the loudspeaker. Then the patients were invited to try to reproduce the same phenomena, activating the recovering muscles).

The third group received the same treatment as the previous two and electrophoresis with drugs containing cerebral gangliosides applied on the limb at the presumed level of nerve injury.

The purpose of the present research is to control the time employed by the regeneration process, according to three different types of treatment adopted.

Early results prove that in case of axonotmesis injuries of limb, recovery times of patients treated with the traditional therapy, EMG biofeedback and ionophoresis are relatively shortened.

Each result has been assessed with electromyography detection and conduction velocity study of the affected nerves.

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# LOCOMOTION II

## Session 7A

# A STUDY OF THE QUADRICEPS ACTIVITY UNDER ISOMETRIC CONDITIONS IN THE PRESENCE OF PATHOLOGICAL KNEE JOINT EFFUSION.

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## INTRODUCTION

Re-education of the quadriceps muscle is a vital part of rehabilitation procedures in lesions affecting the knee joint. Many of these lesions result in effusion into the knee joint cavity. One object of the rehabilitation procedures is to reduce the amount of quadriceps atrophy and to increase the strength of the muscle. Atrophy of the quadriceps can be due to the knee joint effusion. Reports show that effusion of the joint leads to reflex inhibition of the quadriceps.

Some investigations have shown that effusion pressures vary with the position of the knee and that tension on the knee joint capsule results in reflex inhibition of the quadriceps muscle. Changes in effusion pressures will alter tension on the knee joint capsule. [De Anrade, J.R., *et al* (1965). Jayson, M.I.V. and Dixon, A. St. J. (1970)]

From a biomechanical viewpoint there is controversy in the literature as to the best angle at which to re-educate the quadriceps muscle. This present study was done to determine if there is a relationship between the degree of quadriceps inhibition in the presence of pathological effusion and the angle of the knee joint at which the quadriceps attempt contraction.

## EXPERIMENT

Fifteen normal volunteers whose ages ranged from 12 to 54 and who had no previous history of knee joint problems were enlisted for the study. Four pairs of surface electrodes were attached to the quadriceps muscle to record [myoelectric] activity from:
1. oblique fibres of the vastus medialis;
2. long fibres of vastus medialis; 3. Vastus lateralis; 4. Rectus Femoris.

The subject was seated on an exercise table 30 inches from the floor. An Orthotron was bolted to the end of the table. The lower part of the subject's right leg was strapped to the resisting pad on the Orthotron arm so that pressure was taken on the lower 1/3 of the anterior aspect of the tibia. The Orthotron arm was adjusted to the required angle and then made immovable. The subject was then asked to perform a maximum isometric contraction by pushing against the resisting pad of the Orthotron. This maximum isometric contraction of the quadriceps was maintained for 5 seconds. The subject was then allowed to relax for 30 seconds. This procedure was repeated 3 times at 4 different angles, 90°, 60° 30° and 0°. [parallel to the ground]. The constancy of knee joint angle was monitored by a parellelogram type electrogoniometer attached to the lateral aspect of the knee.

Five subjects with pathological effusion of the knee joint but with a pain free range of motion and no neurological disorder were also selected. The ages ranged from 23 to 73. These subjects followed the same experimental procedure as the normal group.

## DATA ANALYSIS

The myoelectric signals from the four pairs of surface electrodes were relayed to Grass RPS 107.B amplifiers. The raw signal was rectified and low pass filtered, producing an envelope wave form. All data was recorded on a Honeywell 8 channel ink recording oscillograph.

The experimental procedure produced 12 separate electromyographic linear envelopes for each component of the quadriceps. [4 angles x 3 attempts]. The measurement of the EMG wave form was performed by a digitizing process and a computer program. This was accomplished by tracing the outline of the curve with electronic recording of a series of points on the curve. A computer program calculated the area under the recorded points and the computed areas for each subject were stored on magnetic tape and displayed on a printout.

The figures obtained for each subject from the three contractions performed at one angle were averaged. This was done for each part of the quadriceps muscle tested. This average was then compared with the activity of the muscle component at the other angles in the same subject. Statistical analysis was performed using a two way analysis of variance (ANOVA) for each individual muscle component and the total quadriceps.

## RESULTS

When data from the normal group were analyzed only electromyographic activity at 60° angle showed a significant difference when compared with the activity at 0°, 30° and 90°. This difference occurred in the Vastus Medialis oblique fibres and the Vastus Lateralis ($P = <.005$). The Rectus Femoris and Vastus Medialis long fibres also showed a difference at this angle. This was not significant. Analysis of the total quadriceps showed a significant difference at 60° ($P = <.025$).

The highest values were recorded at 90° with 0° or 30° having the next highest.

Preliminary analysis of the effusion subjects shows the highest values at 90° and lowest values at 0° and 30°. Five more effusion subjects are to be tested.

## DISCUSSION

Lieb and Perry (1971), using 8 different angles, have demonstrated that emg output of the quadriceps is uniform at different angles of the knee under maximum isometric conditions. This present study demonstrated similar results apart from the data obtained at the 60° angle.

The subjects with knee effusion did not show the same uniformity of output at all angles. In the abnormal group the lowest emg values were obtained at 0° and 30°.

The differences identified between normal and effusion subjects indicate that the EMG values were highest when the knee was held in 90° flexion. Activity levels decreased as the knee moved to 60°, 30° and 0° [full extension]. These results support the concept of knee effusion, with increased intercapsular pressure, inhibiting quadriceps recruitment during contraction.

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# TECHNICAL PROVISIONS IN DYSFUNCTION OF THE LOWER LEG MUSCULATURE

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In dysfunctioning of the lower leg musculature, as seen in disturbances of the central or peripheral motor neuron, a number of provisions may be considered to be prescribed for the patients. Distinction should be made between active and passive provisions.

1. Active

In this instance the dysregulated musculature is stimulated indirectly and electrically. It will be clear that this is only possible in disorders of the central motor neuron, when the peripheral motor neuron is still intact. The stimulation is effected by means of a peroneal stimulator.

2. Passive

a. The traditional solution by means of a posterior bar ór spring.
b. Orthopaedic footwear with built-in arthrodesis socket.
c. An orthosis which is worn round the ankle.
d. The simple fitting of the foot in equinus position in longer existing peripheral pareses.

The peroneal stimulator (1)

This kind of electrical stimulation was introduced by Liberson in 1960. One of our studies (Zilvold, 1976) has shown that, as regards the therapeutic effect, the peroneal stimulator may at least be put on a par with standard conservative therapy. In addition, a number of advantages could be noticed in comparison with the common conservative therapies:
through electrical stimulation inactive muscles started again contracting systematically. As a result the condition of these muscles improved, as well as the circulation in the affected leg.

Another advantage is that the patient can perform the treatment himself and that he may continue to do so for many years.

Passive provisions (2)

a. The posterior bar or spring, appropriate for central and peripheral lesion, with the advantage that they can be readily supplied. Moreover this solution is suitable for lightweight summer-footwear that is usually more airy, which makes the application of appliances in or to these shoes more difficult.

b. The orthopaedic shoe with an arthrodesis socket round the malleoli, appropriate for peripheral and central pareses. Due to the fixation round the malleoli the foot cannot drop anymore.

   The socket may be built-in in a high-top shoe or boot. Particularly the last solution may provide female patients with a product that will meet their cosmetic demands.

   If the spasm is too strong an elastic bar will be preferred, because the spasm is often so strong that it could ruin the socket.
   On the other hand it has proved possible recently to position the ankle joint of patients with a strong equinus-varus deformity by means of phenolization in such a way that yet a shoe with a socket could be provided.

c. The same applies to c) the orthosis round the ankle. This is a fine cosmetic solution. The orthosis can be worn under stockings and in commercial footwear or in normallooking orthopaedic footwear.

d. In some patients with a peripheral peroneal paresis that has existed already quite a long time it is possible to adequately resist the paresis by means of footwear constructed for fixed equinus foot deformity.

The musculi peronei and tibialis anterior namely have become fibrotic in the course of time and have shortened. In fitting the foot in the fixed equinus deformity position the muscles contract maximally, thus automatically compensating the existing drop-foot.

# MUSCLE ACTIVATION DURING GAIT AND ISOKINETIC KNEE MOVEMENTS IN PATIENTS WITH RHEUMATOID ARTHRITIS.

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The purpose of this study was to evaluate neuromuscular function in patients with arthritic knee joints. Characteristics of torque, knee motion and muscle activation during isokinetic knee movements and gait were correlated to clinical information in 16 adult patients, 14 women and 2 men, with rheumatoid arthritis. In order to provide comparative data, a similar study was carried out in 19 healthy women subjects.

Surface electrodes were placed on the vastus medialis, vastus lateralis, medial hamstrings, lateral gastrocnemius and tibialis anterior muscles of the more affected lower extremity. Electrode leads were plugged into an electrode board and connected to the electrode selector panel of a Grass polygraph by means of a long shielded cable. The myosignals were fed into AC preamplifiers and recorded first as unintegrated EMG to check for movement artifacts or interference and later as integrated EMG for the analysis procedures.

Torque during knee extension and flexion movements of both lower extremities was measured with the Cybex II isokinetic apparatus at three speeds of angular rotation (30, 90 and 180 degrees per second). Subjects were seated and immobilized on a specially designed table which provided back and thigh support but allowed the lower legs to hang freely. The rotational axis of the torque measuring device was placed opposite the approximate center of rotation of the knee joint and the mechanical lever arm was attached to the lower leg.

The subjects performed the isokinetic movements in an arc from 90° of knee flexion to full extension under standardized conditions. The knee angular position was tracked by an electrogoniometer fixed to the rotational axis of the Cybex apparatus. Torque-angle curves were displayed on a storage oscilloscope and used for performance feedback and determination of the starting position for the knee movements, in addition to being recorded concomitantly with the electromyographic data on the polygraph.

Following completion of the torque measurements, the electrodes were left in place and pressure-sensitive electronic footswitches to demarcate heel, mid-foot and toe contacts were taped to the soles of both shoes. The footswitch signals and electromyographic activity from the five muscles were recorded simultaneously during walking first without and later with the addition of a specially designed electrogoniometer attached to the lower leg.

Computer programs were written to handle the numerical values obtained by manual measurement of the original recordings. Analysis procedures emphasized the description of muscle coordination and timing characteristics, quantitative evaluation of total myoelectric activity and relative contributions of the muscles to specific movements.

The results describe torque-angle relationships and the patterns of muscle activation in the five muscles during the different isokinetic movements and gait in the patients and in comparison to normal values. Correlations of the torque, angle and electromyographic data were used to define criteria descriptive of the neuromuscular function of the lower extremity. Similarities and differences observed in the neuromuscular function of the patients and normal subjects are discussed. The results of this study should contribute to a better understanding of neuromuscular function in normal and arthritic knee joints and to the evaluation and optimization of therapeutic approaches.

NOTES

LOCOMOTION DATA EXTRACTION USING A MICROPROGRAMMED PROCESSOR

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## Introduction

There are many methods being used presently for the extraction and collection of locomotion data (see [1] for an excellent review). The most popular is the use of some form of photography. The problem with these has been the turn around time envolved in the processing of the data and the lack of any on line computer representation. The work discussed here describes a system capable of real time data extraction.

## The Design Objective

Using present day technology, the objective was to design a data acquisition system capable of real time extraction of locamotion data. The data collection device had to be simple to operate with the method of marking limb coordinates being simple and unobstructive to body movement. It was desirable to have a system with the capability of monitoring many markers (body or reference) with a good degree of accuracy. The sample rate of the system should be high enough such that the spatial resolution in a frame as well as from frame to frame provided adequate results. Above all, the system should be relatively inexpensive and capable of storing a sufficient amount of data.

## The System

To meet the above objectives, the system was designed using salient features found in a few systems presently in use [3]. The processing of the raw extracted data is done by a microprogrammed processor.

The system is shown in figs. 1 and 2. It consists of basically 2 parts, a sampler which collects and stores a frame of sampled data, and the processor, which processes the sampled data to arrive at the marker coordinates.

A standard, 60 frames / sec, 525 line interlaced Sony video camera is used as an optical sensor to obtain the sampled data points. The body markers used are halfs of ping pong balls as described in [3]. The sampled data points are the elements of a matrix which a marker covers. The sample matrix (or sample window) is symmetric in that the horizontal and vertical resolution is equal. Two window sizes are available, 128 x 128, and 256 x 235.

The system is capable of storing data for up to 20 markers (body or reference). The x and y counters with their respective detection logic are used for the sample window and to form the matrix coordinate values. There are 2 sample memories, each 1k x 16 bits in size. The memories are used to store the sampled data points obtained from a single video frame. When the sampler is storing data in one of the two memories, the processor has control of the other. It is extracting the sample data points and calculating the actual marker coordinates. When a particular frame is finished, the memories are exchanged and the process continues.

To calculate the marker coordinates in real time, a bit sliced micro-programmed processor is used. Since the sampled data points stored by the sampler is in serial form, the processor must sort the points, cluster them with respect to the markers they belong to and calculate the final marker coordinate. To use an "off the shelf" processor is impossible due to timing limitations (reduction of this data in 1/60 of a sec.). Instead, a processor was designed with an architecture to optimize the reduction and clustering algorithm. The 2903 bit slice was used with the writeable register store expanded to 32 registers. The architecture has many special features which increase the throughput capability of the processor. The reduction and clustering algorithm is totally micro-programmed. The resultant marker coordinaters are then stored in a memory which can be accessed after a run is completed to provide on line displayed and do further calculations with this data.

## Conclusion

The work described here has taken advantage of present technology to provide a simple, compact system for the extraction of locomotion data. Many additional features could be added to the data acquisition system to provide such options as initial ordering of markers, on line display or preliminary calculation of other clinical information [3].


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## Acknowledgment

This work was supported in part by the National Research Council of Canada.


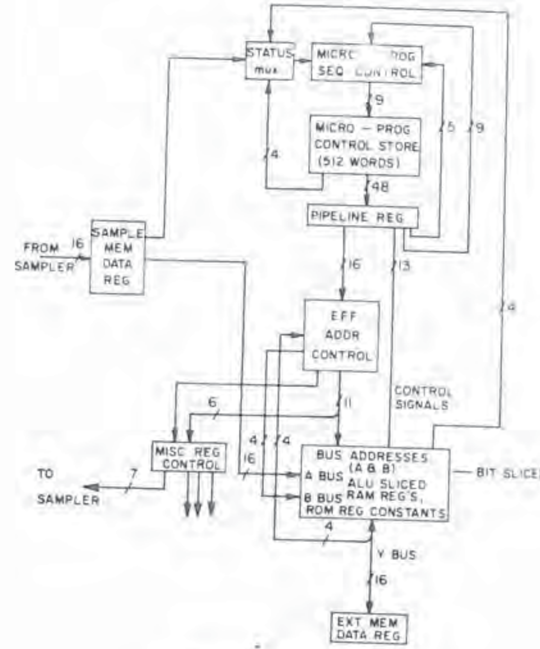


FIG 2 PROCESSOR

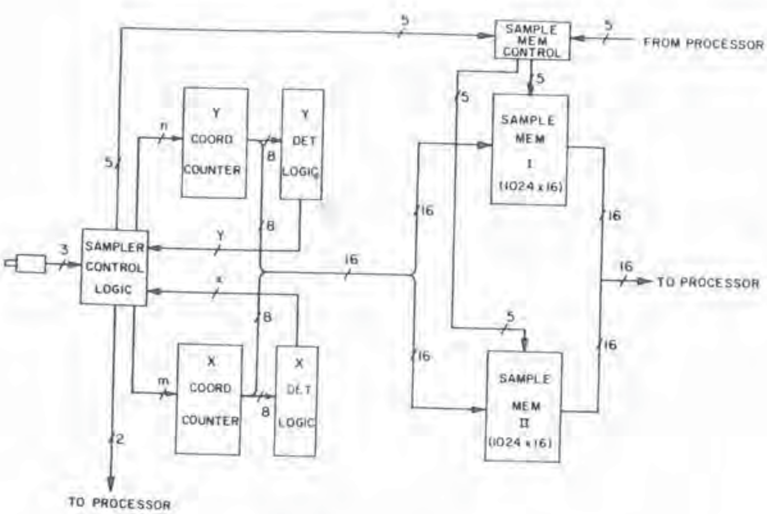


FIG 1 THE SAMPLER

# NOTES

# KINESIOLOGY III

## Session 7B

# THE UTILITY OF COMBINING EMG AND MECHANICAL WORK RATE DATA IN LOAD CARRIAGE STUDIES

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## INTRODUCTION

Ergonomic studies of load carriage have been of interest for many years. Most load carriage studies, however, have considered only metabolic cost in attempting to determine the most "efficient" load magnitude, distribution, mode of carriage etc. (cf. Cathcart et al.,1928; Soule et al.,1978). These studies have provided useful basic information on load carriage but, alone, the physiological measures of energy cost have not been sensitive enough to adequately detect differences in similar load carriage devices. Moreover, these single indices of performance have provided little insight into the interaction of the load and the carrier.

The purpose of this paper is to demonstrate the utility of combining EMG information with linked body segment mechanical energy and work analyses when assessing the effects of load magnitude on the carrier.

## METHODS AND PROCEDURES

Six physically active, healthy men in their twenties carried loads of 0.00, 15.16, 19.30, 22.65, 28.63 and 33.85 kg (20 to 40% of body mass) in a specially built back pack while walking on a horizontal treadmill at 5.64 km/h. Bipolar silver/silver chloride electrodes were placed about 3 cm apart over the bellies of the rectus femoris, gastrocnemius, lumbar erector spinae and spinal region of the trapezius muscles. The electrodes were hard wired to the amplifiers (BIOAMP model 2122, CMR 90dB, BW 10-1000 Hz) and the raw EMG signals, from 5 strides, were full wave rectified and integrated in an analog amplitude reset integrator. The subjects were filmed at 33 frames per second during minutes 7-8 and 10-11 of a 12 minute walk. The cartesian coordinates of selected joints were digitally filtered and mechanical energy time histories of each of the segments of a linked 15 member model of the subjects (including the back pack) were computed for one complete stride cycle. The instantaneous energy level of each segment was given by equation 1 and the internal mechanical work output by equation 2.

$$TE_{ij} = PE_{ij} + KE_{ij} + RE_{ij} \qquad (1)$$

$$WT_{wb} = \sum_{j=1}^{f} \left| \sum_{i=1}^{s} (\Delta TE_{ij}) \right| \qquad (2)$$

where $TE_{ij}$, $PE_{ij}$, $KE_{ij}$, $RE_{ij}$ are the total, potential, translatory and rotational kinetic energy of segment i at time j; $\Delta TE_{ij}$ is the change in energy level of segment i from film frame j-1 to j and $WT_{wb}$ is the total work output which accounts for mechanical energy transfers within and between body segments. Expired gas samples were taken while the subjects stood for 12 minutes with the load on their back and from minutes 5-8 and 8-11 during the walk, and analyzed according to standard procedures. "Muscular efficiencies" of the subjects at each load condition were calculated as the ratio of the mechanical work rate to net metabolic rate.

## RESULTS AND DISCUSSION

The integrated EMG activity (arbitrary units) for four muscles is plotted as a function of load in Fig. 1. Statistical analysis showed no significant differences in rectus femoris activity across loads. Gastrocnemius activity at 23, 29 and 34 kg was higher ($p<.05$) than at the zero load condition. The erector spinae EMG was higher ($p<.05$) at the 29 and 34 kg loads than at the no load condition and the activity at the 34 kg load was higher than at all other loads. Trapezius EMG at the 34 kg load was higher ($p<.01$) than at the zero and 15 kg loads. No other statistically significant differences were observed.

The EMGs indicated that excessive muscular stress was placed on the lower back at the 34 kg load as evidenced by the sudden rise in activity from the 29 kg load. Somewhat surprisingly, no significant elevations in knee extensor activity were observed at even the heaviest load, although ankle extensor activity increased from the no load to loaded conditions. This implies, that no appreciable gait pattern changes were invoked to accommodate heavier loads.

Discontinuities in the EMG activity levels of the trapezius, erector spinae and gastrocs appear at the 19 kg load and seem to be data processing anomalies at first glance. However, Fig. 2 also reveals a discontinuity in mechanical work rate, $\dot{W}T_{wb}$, at this load which the metabolic data were not sensitive enough to detect. Paradoxically, this step elevation in mechanical work rate in the absence of a corresponding increase in metabolic rate appears as an increase in muscular efficiency (Fig. 2). One might interpret this as a desirable load to carry if the EMG data were not available which showed elevated trapezius activity accompanied by slight reductions in gastroc and erector spinae activity at the 19 kg load. Study of the individual body segment mechanical data (not shown here) revealed slight elevations in the shoulder bar, head and trunk work rates at this load. All of these discontinuities may have been caused by an undesirable resonance between the back pack and carrier at this particular load and cadence.

The mechanical data corroborated the suggestion from the EMG information that major changes in gait pattern were not implemented to carry the heavy loads. No significant differences in thigh energy work rate with increased load emerged. The foot and shank work rates were higher at all loads than at the no load condition but none of the loaded conditions were different from each other.

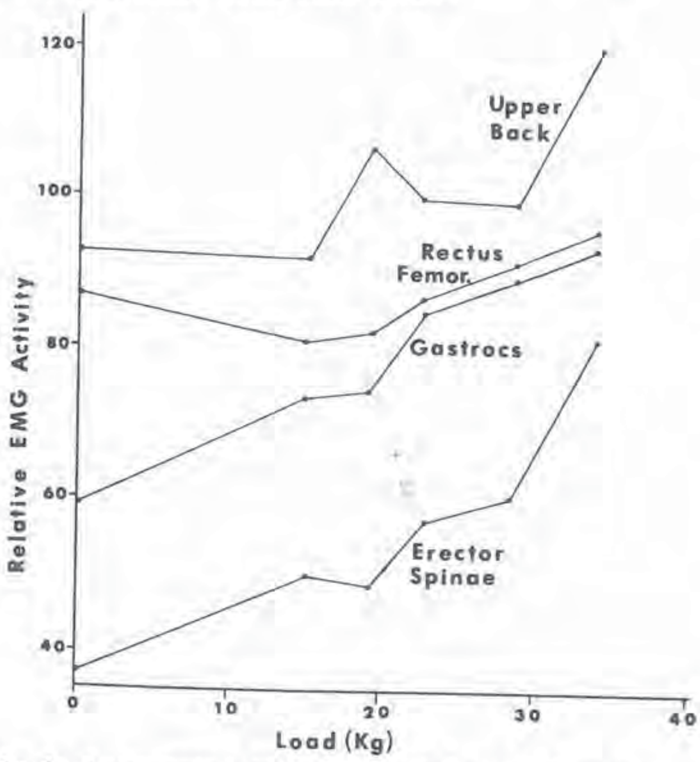


Fig.1. Integrated EMG Averaged Over 5 Strides
Upper Back-Trapezius

Fig.2. Work Rate, Efficiency and External Load
Nm - "Muscle Efficiency" Using Net Met. Rate
$\dot{W}T_{wb}$ - Mechanical Work Rate

## CONCLUSION

Neither EMG, body segment mechanical energy levels and work rates nor metabolic rate data by themselves provide adequate insight into the response of a carrie to increasingly heavy loads. By combining these analytical techniques, however, it may be possible not only to detect differences in load carriage devices but also to yield information useful as design criteria.

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## ACKNOWLEDGEMENT


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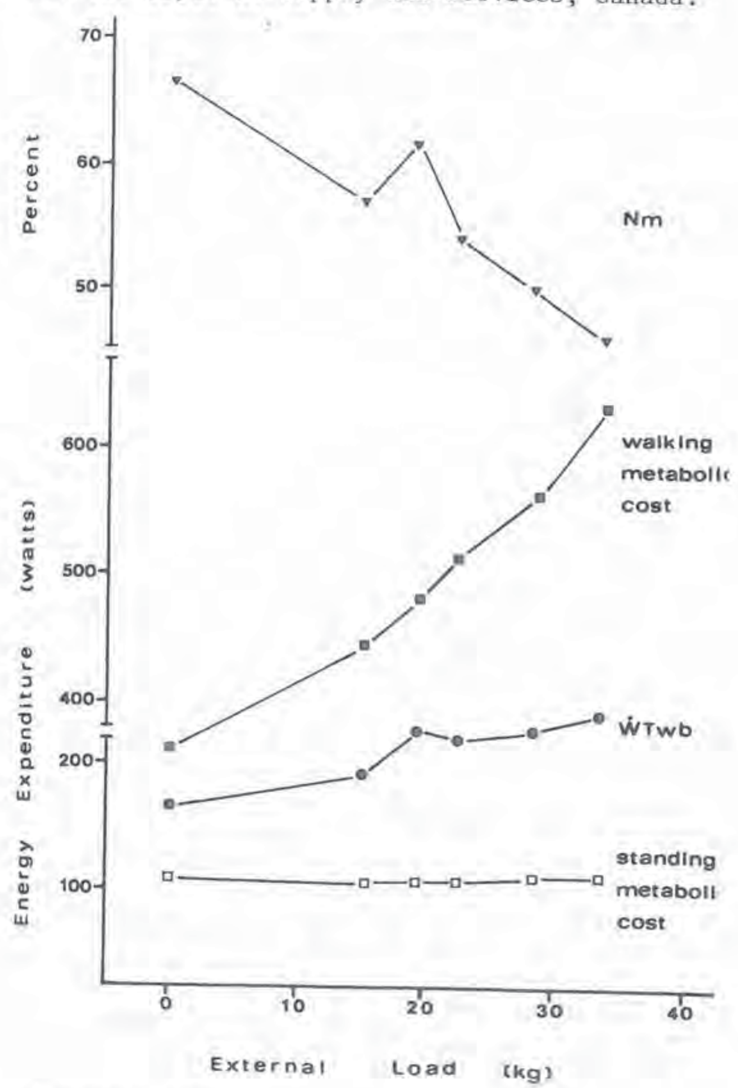

# ANALYSIS AND MEASUREMENT OF THE LOADS ON THE LUMBAR SPINE

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The loads on the lumbar spine have been studied during table work. Two methods were used; quantitative electromyography and biomechanical models.

## Methods

The myoelectric activity was picked up by bipolar surface electrodes at twelve locations; over the posterior back muscles at C 4, T 8 and L 3 levels, and over the external and internal oblique and rectus abdominus muscles. The electrodes were always placed symmetrically on both sides of the midline. The signals were rectified and low-pass filtered, and analyzed for total energy content.

The biomechanical model consisted of prediction of load from simple considerations of equilibrium. Mass center locations of the major body segments above the L 3 level, and of any load held were measured experimentally. From the appropriate weight and moment data, the net reaction at the L 3 level was computed. The major muscles crossing the L 3 level were represented by a few single equivalent muscles, and their contraction forces determined from the net reactions on a statically-determinate basis. Finally the compression on the spine was computed from the external weight and internal muscle contraction forces.

The study was performed on ten subjects in two separate series and included about 60 worktasks and bodypositions. Systematic changes were made in the weight of objects handled, in lever arms and in the direction of activity. The first series was completely controlled with respect to bodyposture, while in the second series the subjects were allowed to perform the tasks as they preferred. The same ten subjects participated in both series.

## Results

The myoelectric activities of the posterior muscles of the spine increased in parallel to increases in the external moments. The weight of the load and the distance between the load and the body are both important factors in this respect, as is the posture. When asymmetric positions were studied, high levels of lumbar muscle activity were found on the side contralateral to the load, while the activities were low on the ipsilateral side. The activity in the rectus abdominus muscle was always low. The oblique anterior muscles were active mainly in asymmetrically loaded positions. Good quantitative agreement was found between the myoelectric activity of the erector spinae muscles and the predicted muscle tensions both in symmetric and asymmetric positions.

## Conclusion

The myoelectric activity of the posterior muscles of the back is directly related to the moment acting on the spine. Simple biomechanical models can be used to predict the activity of these muscles. Myoelectric activity measurements and biomechanical models can be combined to help work place designers reduce the load on the spine during work.

## Acknowledgement

This work was supported by the Swedish Work Environment Found (Project No 77/100), USPHS Grant OH 00514 and Development award AM 00029.

NOTES

RESONANT CHARACTERISTICS OF THE HUMAN BODY DURING FORCED OSCILLATIONS ABOUT THE ANKLE JOINT

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## INTRODUCTION

Landing after a small jump with ankle extensors in a state of sustained contraction leads to a damped oscillation of landing force as measured by a force platform. Cavagna (2) compared this elastic bounce of the body to the movement of a damped harmonic oscillator and from characteristics of the damped sinusoid, ascribed properties of viscosity (B) and elasticity (K) to the human body. Bach (1) reexamined Cavagna's study and concluded that viscous and elastic properties were primarily a function of ankle extensor musculature and that the mass above the knee was adequately represented by a rigid mass. It was then argued that since viscoelastic characteristics were primarily a function of mechanical properties of ankle extensor musculature, any model of this movement should reflect the underlying muscular nature. Consequently,. a model reflecting current linear models of muscle was adopted (Figure 1).

This viscoelastic arrangement suggested the experimental hypothesis that the human locomotor system would resonate at certain frequencies of voluntary activation. The resonant frequency of a mechanical system is the frequency at which Gain, G, defined as the ratio of the output magnitude to the input magnitude, is a maximum. For the model shown in Figure 1, G is described by the function:

$$G = \frac{\omega/B}{\sqrt{\left(\frac{1}{M} - \frac{\omega^2}{K}\right)^2 + \left(\frac{\omega}{B}\right)^2}} \qquad (1)$$

where $\omega$ is the angular frequency, B is the viscous coefficient, K is the elastic coefficient, and M is the mass. Viscoelastic systems also exhibit a phase delay, $\phi$ , between input and output given by:

$$\phi = \frac{\pi}{2} - \tan^{-1}\left(\frac{\omega/B}{\frac{1}{M} - \frac{\omega^2}{K}}\right) \qquad (2)$$

## METHODS

Viscous and elastic coefficients for 5 subjects were obtained from damped sinusoids of landing force as described by Bach (1). Subjects oscillated vertically on a force platform such that output force approximated both magnitude and frequency of a sinusoid produced by a signal generator and displayed simultaneously with force on an oscilloscope. Magnitude of the generated sinusoid remained constant at $\pm.5$ times body weight and frequency was varried between 2.5 and 5.5 Hz. Rectified, Paynter filtered (10 ms time constant), lateral gastrocnemius EMG and force data were collected and digitized. Ensemble averages of force and EMG over 10 oscillations were subjected to Fourier analysis to determine gain and phase functions. These were matched by linear regression to theoretical functions obtained by substituting the subjects' values of B, K, and M into equations 1 and 2.

## RESULTS AND DISCUSSION

Results showed good agreement between theoretical and experimental gain ($r=.633$, $df=39$, $p<.01$) and phase ($r=.915$, $df=39$, $p<.01$) functions. The average experimental resonant frequency was 2.88 Hz. Theoretical and experimental gain and phase functions are shown in Figure 2.

Patla (5) has suggested that the rectified, lowpass filtered, ensemble average of EMG is a measure of instantaneous power output of a muscle. A realtionship with oxygen consumption rate is therefore implied. The study outlined here and that of Patla (5) in combination suggest that at certain frequencies of muscular contraction, energy cost is minimized relative to other frequencies.

The "human resonance" concept provides alternative explanations for some phenomena reported in the literature. Examination of Equation 1 reveals that for this system, G will be a maximum (ie. the system will reonate) when:

$$\omega = 2\pi f = \sqrt{K/M} \qquad (3)$$

Elastic coefficients for a variety of loads were determined by Cavagna (2). These can be used to predict resonant frequencies of the human body under a variety of conditions.

Examination of high speed cinematographic data from elite runners showed a disproportionate decrease in the frequency of ankle extension during the drive phase in the fatigue state. These changes were toward a frequency nearer the predicted resonant frequency for this situation. These results concur with Hatze's (3) suggestion that in fatigue, speed optimizing strategies are abandoned for energy optimizing strategies. Melville Jones and Watt (4) suggested that preferred frequencies of repetitive hopping on one leg were related to timing of otolith reflex facillitation of the subsequent hop. However, reexamination of their data by the author revealed that the frequency of the landing impulse at preferred hopping frequencies was very close to resonant frequencies predicted for this situation. It is an attractive possibility that humans choose movement patterns which take advantage of resonant characteristics of the system in order to minimize energy consumption.

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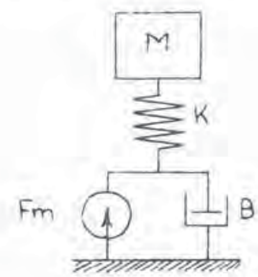


Figure 1. A viscoelastic model of the human body.

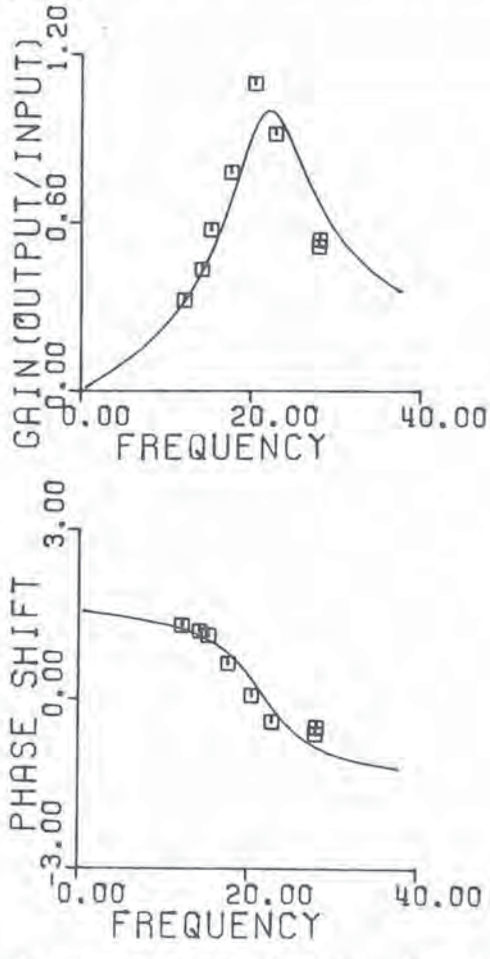


Figure 2. Theoretical (solid line) and experimental (squares) gain and phase functions for subject BC. Phase and frequency are in angular units.

ROTATION EFFECTS IN SCOLIOSIS

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## INTRODUCTION

The present work deals with the problem of the evaluation of torsion effects on the vertebral column's posture. The study is particularly devoted to the phenomenon of scoliosis.

## SIMULATION MODEL

By means of model simulation it can be assessed that the forces which are necessary for obtaining an appreciable deviation of the vertebral column in the frontal plane (if the column is compelled to move without torsion) are very far from those physiologically acceptable. The aethiology of scoliosis is then linked to the study of torsion effects and ultimately to the study of rotator muscle effects. It is also important to clarify the interactions with flexor muscle masses. Model simulation has been conducted by considering the actions of muscular masses. The following paragraph will illustrate the situation and computations with reference to a vertebral junction.

## STUDY OF A VERTEBRAL JUNCTION

A pair of vertebrae and the interposed intervertebral disc are now considered. A set of coordinate X,Y,Z are taken as reference axes with their origin in the center of the intervertebral disc, the YZ plane is taken to be coincident with the frontal plane, the plane XZ coincident with the sagittal plane.
The inferior vertebra is assumed fixed in the preceeding reference set of coordinate axes and the displacement of the upper vertebra with respect to the inferior contiguous one is assigned by giving the displacement of a set of coordinate axes fixed to the upper vertebra. The origin of the moving set of axes is the same as that of the fixed set, and given by the center of gravity of the intervertebral disc. Usually the position of a vertebra with respect to its contiguous one is determined by a measurement of the angle $\psi$ between the two adjacent vertebral plates in frontal projection and computation of the angle $\phi$ of rotation of the same vertebra around its vertical axis by using frontal and sagittal x-ray projections (see fig.1). Let X'Y'Z' be a set of coordinate axes rotated around X axis of the angle $\psi$ and i',j',k' the unit vectors of this set of axes and i,j,k the unit vectors of the fixed axes X,Y,Z, let also

$$\vec{r}=\vec{i}x+\vec{j}y+\vec{k}z$$

be the position vector of a point P(X,Y,Z) of the moving vertebra; it is possible to define a dyadic $\Psi=\vec{i}'\vec{i}+\vec{j}'\vec{j}+\vec{k}'\vec{k}$ such that $\vec{r}'=\Psi x\vec{r}=\vec{i}'x+\vec{j}'y+\vec{k}'z$ where $\vec{r}'$ is the position vector of the above mentioned point P(X,Y,Z) after rotation of the angle $\psi$ around X axis. In the same way let x",y" z" be a set of coordinate axes rotated of the angle $\phi$ around z' axis and by defining the dyadic

$$\phi=\vec{i}''\vec{i}'+\vec{j}''\vec{j}'+\vec{k}''\vec{k}'$$

results $r''=\Phi x\vec{r}'$

where r" is the position vector of the point P after the two successive rotations $\psi$ and $\phi$.
It is possible then to determine a dyadic M such that

$$M=\Phi x\Psi=\vec{i}''\vec{i}+\vec{j}''\vec{j}+\vec{k}''\vec{k}$$

and then $\vec{r}''=Mx\vec{r}$

It is possible to find the following relations (for notations see fig.3).

$$\vec{TR}'' = Mx\vec{TR};\vec{TL}''=Mx\vec{TL};\vec{S}''=Mx\vec{S};\qquad \vec{QR}''=\vec{QR};\vec{QL}''=\vec{QL}.$$

The force vectors are proportional to muscle lengths so:

$$\vec{F}_R=K(\vec{TR}''-\vec{QR}'')=K(Mx\vec{TR}-\vec{QR})$$
$$\vec{F}_L=K(\vec{TL}''-\vec{QL}'')=K(Mx\vec{TL}-\vec{QL})$$
$$\vec{R}_R=K(\vec{S}''-\vec{QR}'')=K(Mx\vec{S}-\vec{QR})$$
$$\vec{R}_L=K(\vec{S}''-\vec{QL}'')=K(Mx\vec{S}-\vec{QL})$$

where K is a constant and the inverted comma means that the vectors are evaluated after the successive rotations $\psi$ and $\phi$. The equilibrium condition is:

$$\vec{M}_{O\psi}+\vec{M}_{O\phi}+\Sigma\vec{M}_{OF_i}+\Sigma\vec{M}_{OR_i}+\vec{M}_L=0$$

where $\vec{M}_{O\psi}$ and $\vec{M}_{O\phi}$ are the reaction moments of the intervertebral disc $\vec{M}_L$, $\vec{M}_{OF_i}$, $\vec{M}_{OR_i}$ are the moments of the applied forces. The sign o indicates that the moments are taken with respect to the origin O of the axes which is the fixed point of the moving coordinate axes.

## FLEXOR AND ROTATOR MUSCLE EFFECTS ON SCOLIOSIS

A performance index has been determined for evaluating the influence of the various muscular masses on spine posture. The performance index gives a measure of spine deviation and thus of scoliosis. This index is given by:

$$P.I=2\Sigma i x_{B_i}^2/\Delta Y^2+\Sigma i\phi_i^2$$

where $x_B$ is the distance of the center of gravity of each vertebra from the Y vertical reference axis, $\Delta y$ is the distance on Y axis between first and last vertebrae involved, $\phi_i$ is the flexion angle in the frontal plane between two contiguous vertebrae. The constant 2 takes into account the assumption that for a scoliotic patient the displacement of the vertebral column from the vertical axis is more dangerous than the flexion.

By using this model the following studies have been performed:

1 - influence of rotator muscle unbalance on the spine configuration.
2 - influence of flexor muscle unbalance on the spine configuration.
3 - influence of flexor and rotator muscle unbalance on the spine configuration.
4 - as in 1 but with vertebral column previously subjected to clockwise rotation.
5 - as in 2 but with vertebral column previously subjected to clockwise rotation.
6 - as in 3 but with clockwise rotation
7 - as in 2 but with counterclockwise rotation.
8 - as in 3 but with counterclockwise rotation.

Two cases of muscular unbalance have been considered:

- muscular unbalance involving a restricted area of the muscular apparatus.
- muscular unbalance involving the entire muscular apparatus.

The results of the above mentioned studies are shown in Table I. The performance index has been normalized to maximum-value.

## CONCLUDING REMARKS

Rotator muscle unbalance is more dangerous than an equal unbalance on flexor muscles (see first column)

Spine torsion worsens scoliotic curve (see second column)

- The combined unbalance of rotator and flexor muscles worsen scoliotic curve in presence of spine torsion.
- Even with muscle balance spinal torsion is sufficient to cause scoliosis.

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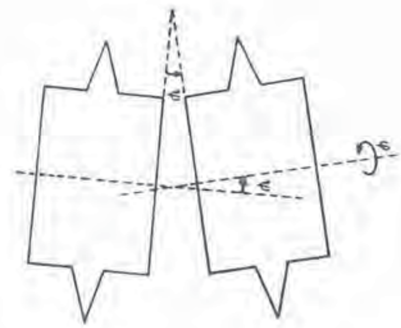

Fig. 1 - Definition of angles $\psi$ and $\phi$

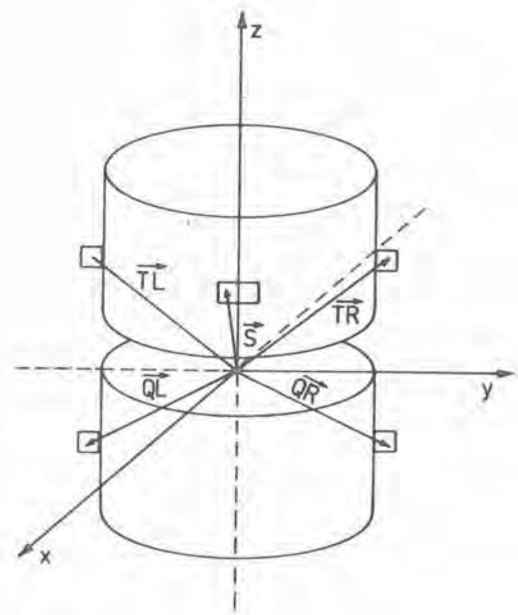


Fig. 2 - Position vectors of vertebral processes.

| | NO ROTATION | C W ROTATION | C C W ROTATION |
|---|---|---|---|
| MUSCULAR BALANCE | [illegible] | 0 31 R | 0 31 L |
| LOCAL ROTATOR MUSCLES UNBALANCE | 0 20 R | 0 32 R | |
| GENERALIZED ROTATOR MUSCLES UNBALANCE | 0 54 R | 0 52 R | |
| LOCAL FLEXOR MUSCLES UNBALANCE | 0 08 | 0 34 R | 0 19 L |
| GENERALIZED FLEXOR MUSCLES UNBALANCE | 0 29 R | 0 42 L | 1 L |
| LOCAL ROTATOR AND FLEXOR MUSCLES UNBALANCE | 0 11 R | 0 36 R | 0 14 L |
| GENERALIZED ROTATOR AND FLEXOR UNBALANCE | 0 76 R | 0 16 L | 0 33 L |

Table 1 - Model Simulation results.

NOTES

# LOCOMOTION III

## Session 8A

FOREARM MOVEMENT DURING HUMAN LOCOMOTION

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## INTRODUCTION

Movement of the upper limb which occurs predominantly in the sagittal plane, i.e. flexion and extension, is an integral component of normal human locomotion. However, this aspect of walking has been infrequently studied.

The muscles which could move the upper limb at the shoulder joint have been investigated and there is a striking absence of activity in the flexors except at the fastest cadences. It has been shown that the extensors of the upper limb at the shoulder joint are active and show a phasic burst of activity which controls and decelerates the flexion at the shoulder joint (Fernandez-Ballesteros et al., 1965; Hogue, 1969; Jackson et al., 1978). During flexion at the shoulder joint there is also flexion of the forearm at the elbow. This is not associated with muscle contraction of the flexors of the forearm at the elbow which are silent but with contraction of the triceps brachii, the main extensor of the forearm. The contraction was not seen at slower cadences and with increasing cadence it appears at an earlier phase of the cycle and with increasing force. Murray et al., (1967) made the most thorough survey of upper limb movement in walking. They showed that the upper limb flexed and extended at both the shoulder and elbow joints once in each complete cycle and that this movement was synchronized with the movement of the lower limb. They mentioned that a few of their subjects showed an "accessory flexion wave" of the forearm at the elbow joint. This observation was not elaborated or investigated.

## METHODS AND OBSERVATIONS

Jackson et al. (1978) described the movements of a mathematical model of the human upper limb during locomotion. For forward swings at the shoulder the model predicted a double wave of forearm flexion at the elbow in each step at low cadences and that this would gradually change to a single wave as the cadence increased. If the triceps brachii were active the model predicted that the change to a single phase could occur at lower cadences - that is, the triceps brachii would prevent the second wave.

Measurements of forearm flexion were made by means of a two dimensional radio frequency goniometer. The construction of this transducer has been described by Jackson (1977). The transducer is capable of measuring angular rotation by means of sensors attached to each side of the elbow joint. No physical attachment is made across the joint itself. Because of this and the lightweight construction of the transducer (25g) the movement being studied is not affected by the measurement process. This is specially important when the movement is controlled by small forces.

With this transducer attached to their right arm the subjects were asked to walk normally at different cadences dictated by a metronome. As many practice attempts as were necessary to familiarize the subject with the experimental procedures were allowed. The phase of the walking cycle was recorded by means of a small metal plate attached to the subject's heel making contact with the metal walkway surface.

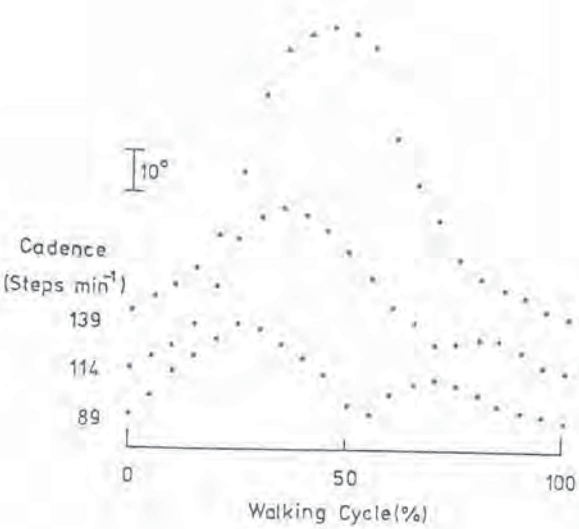


FIG. 1.

Figure 1 illustrates the type of result obtained in three of the seven subjects. The double flexion of the forearm was seen at low cadences. As the cadence was increased the double flexion movement of the forearm disappeared and became a simple flexion movement occurring once in each walking cycle. The amplitude of the flexion was small. The subjects who showed the double flexion movement of the forearm were those who showed a less than average flexion and extension movement of the upper limb at the shoulder.

Electromyographic recordings showed no muscular activity in either the flexors or extensors of the forearm at the elbow. With increasing cadences the triceps brachii became active and with increasing intensity and at an earlier phase of the cycle.

## DISCUSSION

It is interesting that the mathematical model can predict flexion of the forearm at the elbow as a result of purely passive forces. The fact that the two phases of flexion predicted by the model and occurring passively were seen in only three of the seven subjects may be explained as follows. Some subjects find it difficult to walk at slow cadences. They say that they do not feel as if they are walking but taking single steps so that the rhythmic co-ordinated movements of normal walking are fragmented. Other subjects may be affected psychologically by the presence of the sensors on their upper limb. They are aware that their arm movements are being investigated.

The change in the movement of the forearm at higher cadences associated with contraction of the triceps brachii may be an attempt to ensure that there is sufficient time for the forearm to return to its original position for the beginning of the next walking cycle. Limiting the range of flexion will also limit the speed of extension.

We would like to emphasize that the flexion of the forearm whether in the form of a single or a double phase, is not due to a "reflex contraction of the stretched elbow flexor muscles" as suggested by Murray et al. (1967) but is a normal consequence of the mechanics of human locomotion at low cadences.

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## ACKNOWLEDGEMENT

We wish to thank Guy's Hospital Endowment Fund for financing this work.

Fig. 1. Flexion of the forearm at the elbow joint for one subject at different cadences. The traces have been displaced vertically for clarity. Each trace is an average of 3 consecutive cycles produced as a series of 20 points by a computer program (see Jackson et al., 1978, for details of the program).

# POPULATION VARIABILITY OF GAIT PATTERNS AT NORMAL WALKING SPEED

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## INTRODUCTION

Electromyographic investigations have been undertaken to study muscular synergy patterns during level walking. For instance see Sutherland, 1966, and Dubo et al., 1976. Many of the results have been summarized by Basmajian (1973) and Paul (1974). A compilation of the results has demonstrated some significant disagreements on the phasings of particular major muscles including the quadriceps and hamstring groups. This leads to a major question. What are the ranges of muscle synergy patterns that can result in normal locomotion? The object of the investigation reported here is to demonstrate the variability of muscular synergy patterns in the normal population walking at free (normal) speed.

## METHODS

### Measurements

The muscular synergy and foot-fall patterns of twenty-five normal subjects ranging between twenty and forty years of age were measured and studied. The electromyograms (EMG) of either major muscles were measured using miniature Beckman surface electrodes and standard EMG differential amplifiers incorporated in a Biosentry Telemetry System. The foot-fall patterns were measured using footcovers and a B & L Engineerint Foot-Pattern Telemetry System.

The specific protocol was:

1. Electromyograms of the following muscles on the right lower extremity were measured: tibialis anterior (TA), peroneus longus (PL), gastrocnemius (GAS), soleus (SOL), rectus femoris (RF), vastus lateralis (VL), medial hamstring (MH), gluteus medius (GM).
2. Foot-fall patterns of both feet were measured simultaneously.
3. All measurements were appropriately filtered and recorded on an analog magnetic tape recorder and an oscillograph recorder.

The population averages were achieved by averaging the event times in all the individual gait patterns. For instance, the average time of onset of contact of the fifth metatarsal was calculated by averaging the normalized times of onset of contact of all subjects. The population EMG patterns were calculated in the same manner except they were first separated into different classes.

## RESULTS

### Kinematics

The population average of the foot-fall patterns are shown in Figure 1. The patterns of all the subjects had the same sequence of events except one subject had a digitigrade foot and another a valgus foot. The variability of an event time is depicted by the double-headed arrow centered round and located below the even occurrence. The length of the arrow represents two standard deviations.

The statistics of certain kinematic parameters were calculated. The average delocity (V) was 0.98 M/S with a standard deviation of 0.17 M/S. Similarly the group cycle time (CT) and stride length statistics were 1.14 ± 0.11 sec and 1.10 ± 0.17 M respectively.

### Muscle Synergy Patterns

The population averages for the muscle synergy patterns are shown in Figure 2. The main characteristics of each muscle's activity can be classified into primary and secondary patterns according to the proportion of subjects exhibiting them. The percentage of subjects having a particular EMG pattern is denoted by the number located to the right of the patterns. In a particular pattern, some phases of activity were not demonstrated by all subjects. This phase is represented by a dashed line and the number above the line denotes the percentage. The double-leg (DLS) and single-leg (SLS) support portions of stance phase are shown on the same line which designates percentage of the gait cycle.

## DISCUSSION

### Kinematics

The foot-fall patterns demonstrated a bilateral symmetry. The relative timing for most of the foot-events is fairly consistant and is indicated by the relatively small standard deviations. Only two foot-events, toe-contact and heel-off, have a large range of times. The statistics on the kinematic parameters show that there is a broad variability in the speed of progression and in stride length and times. These have been shown to be a function of stature (Grieve, 1968).

### Muscular Synergy Patterns

The primary muscular synergy patterns coincide fairly well with the classically accepted ones (Basmajian, 1972). However, the secondary patterns demonstrate the population variability and show that the different results found by investigators using a small number of subjects are really a reflection of individual differences. The patterns show that in the leg some individuals utilize the TA during mid-stance and/or the plantar flexors during loading (first DLS). The hamstrings are not always active for deceleration (late-swing) and the rectus femoris is sometimes active throughout SLS. Similar to the hamstrings, the gluteus medius is not always active during deceleration. In general, the primary patterns for the thigh muscles do not always have a second phase during the second DLS stage.

## CONCLUSION

Given the consistency in the time of most of the foot-events, the foot-fall pattern of normal individuals seems to be a stereotyped pattern. The population muscular synergy patterns demonstrate a great amount of inter-individual variability. This variation could just be a consequence of the different walking speeds. However, given that the lower extremities form indeterminate systems, several muscle patterns can feasibly produce the same motions.

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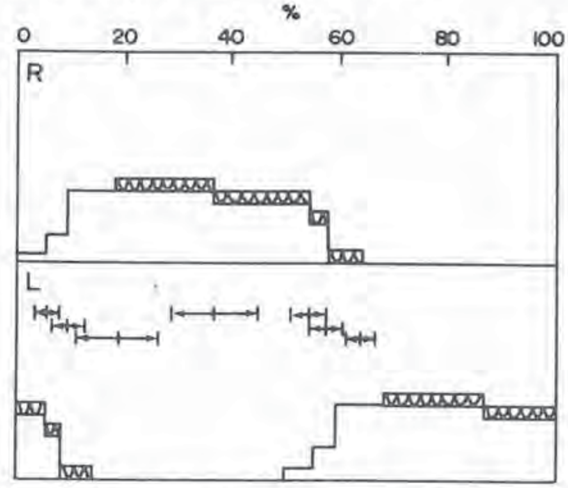


Fig. 1 Foot-Fall Patterns. Contact of the hee-, 5th metatarsal, 1st metatarsal and large toe are encoded as deflections of 1, 2, and 4 units, and oscillation of 1 unit respectively. Several parts in contact simultaneously result in a simple sum.

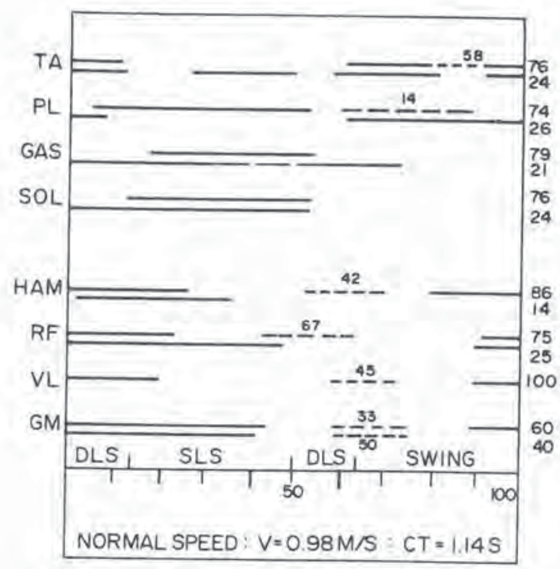


Fig. 2 Muscle Patterns

ELECTROMYOGRAPHICAL ANALYSIS OF THE AXIAL MUSCLES IN THE DOG'S LOCOMOTION ON THE TREADNILL.

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## INTRODUCTION

The analyses of the mammalian locomotion, especially in the field of neurophysiology and electromyographical kinesiology, have been made a great progress in the last ten years. The electromyographical analyses of the mammalian locomotion, however, have been done about the movement of the four legs. The auther studied electromyographical analysis of limb muscles and some axial ones during dog's locomotion. These results required the further analysis of axial muscles. The main motor muscles transferred from the axial muscles in fish to the limb muscles in terrestrial vertebrates after the appearance of the latter who evoluted from the former. It is, however, interesting problem in mammalian locomotion what a role the axial muscles as large as ever perform and what sort of coordination they have to the movement of four legs.

## MATERIALS AND METHODS

The six male mongrels with no locomotor deficiency were trained to perform the locomotion on the treadmill. Bipolar wire electrodes made of 120 $\mu m$ enamel copper wire were put into the seven epaxial trunk, two hypaxial trunk, six epaxial neck and two hypaxial neck muscles by means of a 27 gauge hypodermic needle. The electrodes were, without anaesthesia, injected into the muscles of the dog standing on the treadmill. An oscillograph was a nine channel electroencephalograph equipped with an ink-writing recording system. 16-mm motion pictures of the treadmill locomotion of the dog were taken at a velocity of 64 fps and were synchronized with the EMG of the oscillograph recording. Two accelerometers were put on the head and lumbar part. To investigate the relation between the muscle activity phase of axial muscles and the stance-swing phase of extremities, the locomotion speed of the dog was changed from 30 to 140 m/min. at an interval of every 10 m/min. The overlap of the stance-swing phase of forelimbs and that of hindlimbs changes according to the speed of locomotion: the phase of ipsilateral limbs, in the slow walk, synchronizes in a relatively long span of time, but diagonal limbs become to move in phase in proportion to the increase of locomotion speed.

## RESULTS AND DISCUSSION

Axial Neck Muslces [Fig. 1]

The epaxial neck muscles such as the splenius, the biventer cervicis, the complex, the longissimus capitis and the multifidus cervicis change their EMG activity according to the position of the head and neck in standing posture. These muscles do not often show the rhythmical EMG activity in the slow walk, but in the usual or fast walk they discharge synchronously four times, sometimes twice, in a stride. These burst discharges coincide with the landing of the forefoot. In trot, the epaxial neck muscles occur the periodical EMG activity coincident with the touching of each forefoot twice a stride, and upper acceleration of the head, at the same time, appears. In slow gallop, a leading leg exists because of asymmetrical gait. The dog gallops in a right leading forelimb in Fig. 1. The epaxial neck muscles show the burst activity with two peaks in a stride. These peaks coincide with the stance phase of each forelimb. Each EMG activity which appears in landing of each forefoot overlaps because of the overlapping of the stance phase of both forefeet. The hypaxial neck muscles, the longus capitis and the sternomastoideus, occur reciprocal electromyographical activity against the epaxial ones. The head and neck position during the locomotion on the slope at an angle of 15° is different from that on the level locomotion. The EMG activity of neck muscles during slope locomotion, however, does not change the phase to the stance-swing phase of forelimb, although it increases the strength.

The epaxial neck muscles occur the activity coincident with the landing of forelimb and may prevent the head and neck from falling against the reaction of the landing of forelimbs.

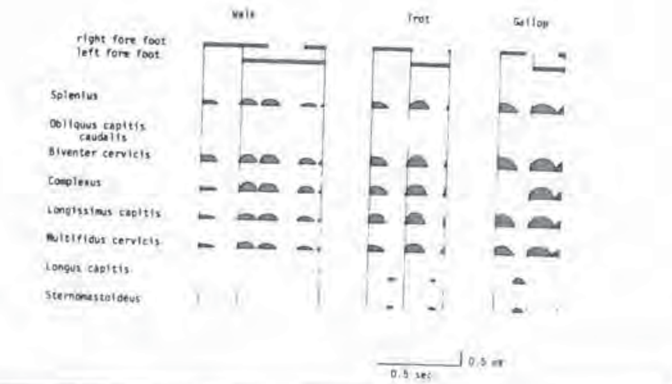
Axial Trunk Muscles [Fig. 2,3]

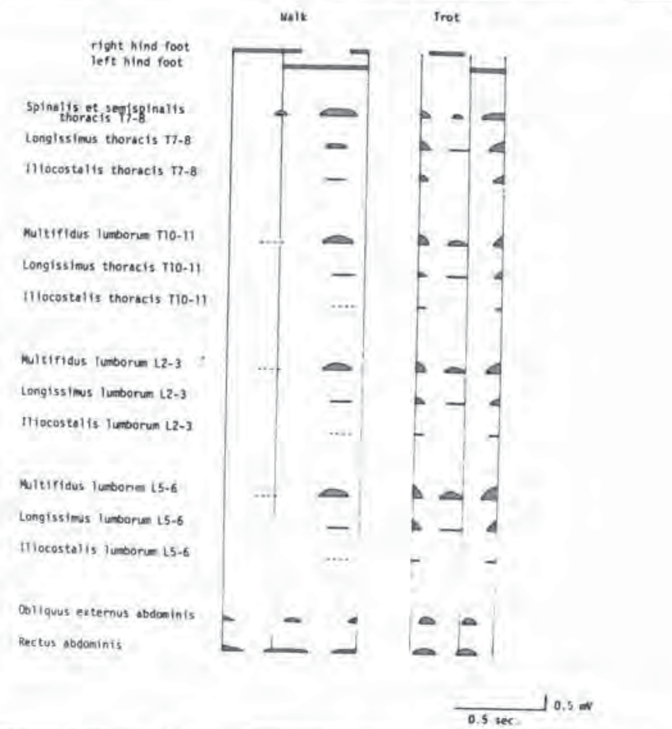
The epaxial lumbar muscles such as the multifidus lumborum (ML), the longissimus luborum (LL) and the iliocostalis lumborum (IL) are not active in the relaxed standing. The epaxial thoracic muscles, the longissimus thoracis (LT) and the iliocostalis thoracis do not often occur the activity, although the spinalis et semispinalis thoracis (SS) is almost always active. In slow walk, ML exhibits EMG once a stride in the latter half of the stance phase of the ipsilateral hindlimb. In proportion to the increase of the locomotion speed, it becomes active twice a stride because EMG appears in the latter half of the stance phase of the contralateral hindlimb. LL shows EMG once a stride and IL exhibits little EMG, in walk. All epaxial trunk muscles occur the activity at the same period. In trot, each muscle increases their activity, and LT and LL as well as ML become active twice a stride in the latter half of the stance phase of hindlimbs. SS occurs the activity ealier than other epaxial trunk muscles. In walk and trot, the nearer the epaxial trunk muscles locate to the vertebral column, the stronger they show the activity.

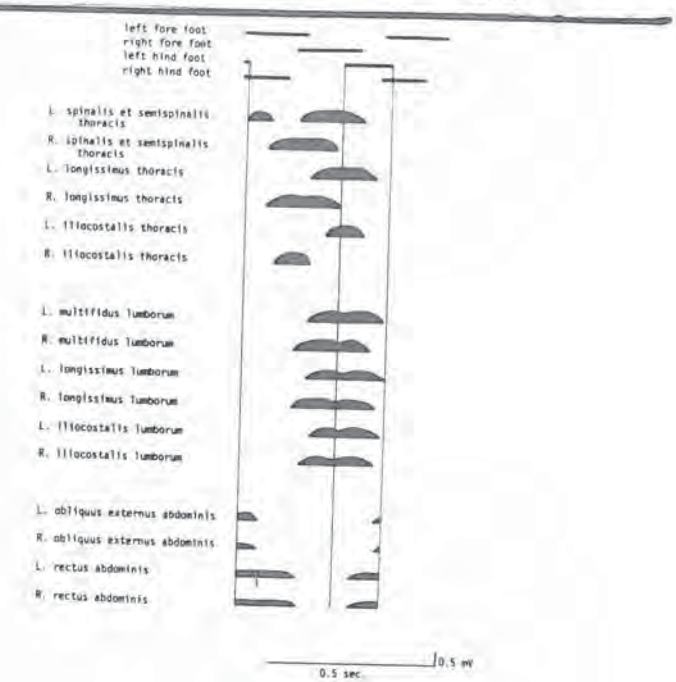
In symmetrical gaits, epaxial trunk muscles may work to prevent the trunk from being flexed and spinned by the force which the hindlimb exerts in the latter half of that stance phase, and may transmit the force to the forequarters.

In slow gallop [Fig. 3], the muscle activity pattern of left side of the body is different from that of the right side because of the asymmetric gait. In Fig. 3, the dog performs the right leading fore- and hindlimb, that is, transverse gallop. The activity of the muscles ipsilateral to the leading limb precedes that of the opposite muscles except SS. The epaxial lumbar muscles, although they have stronger activity, become active later than the epaxial thoracic ones. Almost all epaxial trunk muscles show the two-peak activity once a stride. Each peak coincides with the period when only the leading forelimb is on the ground, namely, to have to support the hindquarters on the fulcrum of the fore-paw, and when only the trailing hindlimb is in the stance phase, namely, to have to support the forequarters on the fulcrum of the hind-paw. The hypaxial trunk muscles, the obliquus externus abdominis and the rectus abdominis, become active reciprocally against the epaxial trunk muscles, although their activities overlapp each other.

Thus, the activity phase of trunk muscles to the stance-swing phase of four limbs in slow gallop is different from that in walk and trot.



ANTAGONISTIC INHIBITION IN DOUBLE-JOINT LEG MUSCLES DURING NORMAL GAIT CYCLE.

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## INTRODUCTION

In a normal human gait, the hip joint is extended and the knee extensors are activated prior to the heel strike (Batty and Joseph, 1966., Okamoto, 1970). In such double-joint movements where the joints are connected in series, the dynamic and neurophysiological features have been reported by Yamashita (1975), and Yamashita and Kumamoto (1976). That is: 1) the resultant force was found not to be the summation of the individual hip and knee joint forces, but was limited by the weaker joint force contributing to the resultant force along the functional force direction of the both joints' simultaneous movement; 2) the electrical behavior of the double-joint leg antagonists exhibits an antagonistic inhibition between them and may suggest which joint is limiting the resultant force. The present experiments were performed to elucidate a possible existence of an antagonistic inhibition between double-joint leg antagonists around the time of heel strike in a normal human gait.

## METHOD

Eighty healthy subjects of both sexes were employed in these experiments; 9 subjects ranging in age from 1.2 to 6 years who characteristically show an infant walking pattern and 71 subjects, ages 7 to 49, who show an adult walking pattern. Electromyograms (EMGs) were recorded from the Tibialis anterior (Ta), the Gastrocnemius lateral head (Gl), the Vastus medialis (Vm), the Rectus femoris (Rf), the Semimembranosus (Sm), the Biceps femoris (Bf), and the Gluteus maximus (Gm) with a multichannel electroencephalograph (at 60 mm/sec) and an electromagnetic oscillograph (at 200 mm/sec), utilizing surface electrodes. Front and side view motion pictures of the walking patterns (at 32 frames/sec), toe and heel contacts with the ground, and angle changes of the ankle, knee and hip joints were recorded simultaneously with the EMGs. Subjects were requested to walk at an adequate normal walking speed for their respective ages, and then, at speeds of about 20 % slower, and faster than normal. Some of the adult subjects were also requested to walk with their upper bodies flexed deeply at the hip joint, or with the knees flexed deeply and the upper body erect, so that in the former case the upper body load would be compensated for mainly at the hip joint or in the latter case at the knee joint.

## RESULTS

The subjects over 7 years of age exhibited adult patterns previously shown by Okamoto (1975). The predominant electrical discharge patterns of the Ta and Gl were summarized as follows: 1) a marked discharge of the Ta was observed at the early part of the swing phase, that is, during dorsiflexion and from the end of the swing phase to the early part of the stance phase, that is, during the shock absorbing action at the heel strike; 2) the main discharge of the Gl was observed at the latter half of the stance phase, that is, when the heel was lifted off (see Fig. 1). The discharge of the Ta at the mid-swing phase and the latter half of the stance phase, and of the Gl from the end of the swing phase to the early part of the stance phase varied. Thus, the subjects did not show any pathological pattern as far as ankle joint movements were concerned.

The Rf showed marked discharge prior to the heel strike (Fig. 1), so that the muscular tension developed sufficiently to bear the heel striking shock at the knee (Grillner, 1976). The discharge lasted into the first double leg support period. Although the discharge of the Bf started nearly the same time as that of the Rf in almost all subjects, it showed a definite tendency to decrease or even disappear immediately before the heel strike in more than 60 % of the subjects (Fig. 1). This tendency increased with increasing walking speed, i.e. slow, 46.5 %; normal. 66.2 %; fast, 72.1 %. During this period where the decreases or cessations in electrical activities of the Sm and Bf were observed, both single-joint extensors of the Vm and Gm showed marked discharges. The EMG pattern of the child group ranging in age from 1.2 to 6 showed the same typical infant's walking pattern as was reported by

Okamoto (1975). The youngest group who have been walking for only a couple of months did not show any spike cessations in the Sm and Bf during the period where many of the adults showed clear spike cessations in the muscles immediately before the heel strike. They were walking with forward swaying postures. The children above these ages who could walk with their upper bodies erect began to show spike cessations in the Sm and Bf during the period as mentioned above.

When some of the adults who demonstrated clear spike cessations in the Sm and Bf most of the time, were asked to walk with their upper bodies flexed deeply at the hip joint, the Sm and Bf showed continuous marked discharges and the Rf diminished its activity during heel contact. In such a posture, the upper body load was compensated for mainly at the hip joint instead of at the knee. When the subjects were required to walk with their knees flexed deeply and the upper body erect, the spike cessations in the Sm and Bf became pronounced.

The cinematographic and mechanical recordings show that the hip joint was extended before the heel strike and the knee joint was fully extended at the heel strike in most cases, but in some cases, the knee was already slightly flexed at the heel strike. Even in these latter cases, the electrical cessations in the Sm and Bf occurred just before knee flexion began.

## DISCUSSION

The discharges of the Vm, Rf, Sm and Bf observed before the heel strike in almost all subjects was programmed at the spinal level as had been suggested by Grillner (1976), and when the excitation level of the Vm and Rf increased beyound a certain level an antagonistic inhibition could be induced by these muscles. Although there is no direct evidence of an antagonistic inhibition, the findings observed in these experiments seem to be sufficient to conclude that the electrical silence observed in the Sm and Bf immediately before the heel strike in a normal gait is most probably due to an antagonistic inhibition, and, therefore the knee joint is the limiting factor in the double support period of the early stance phase.

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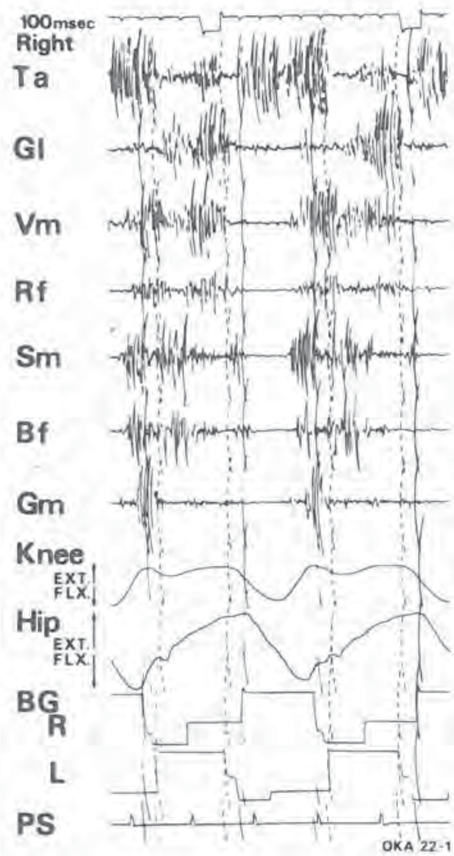


Fig. 1. The typical EMG and angle change patterns of the right leg muscles during normal gait cycle. BG: Foot contact signals from the right (R) and left (L) feet. The downward deflections show heel and toe contacts with the ground, and the periods betwen the solid and broken lines, double support periods. PS: Photo signals.

# ELECTROMYOGRAPHIC ANALYSIS OF ACQUISITION OF A NOVEL CLOSED MOTOR SKILL

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## INTRODUCTION

In attempting to understand the process for acquiring motor skills, many and varied approaches have been employed. Neurophysiologists and neurochemists have investigated the cellular and molecular level; psychologists and specialists in motor learning have investigated the end performance; and, recently, kinesiologists have investigated the muscular, kinematic and kinetic aspects. Poulton, (1957) examined the end performance. In classifying motor skills, he proposed that a closed skill is acquired under a stable and predictable environmental state, whereas an open skill is acquired under changing and unpredictable environmental conditions. Acquisition of closed skills involving the upper extremity has yielded conflicting changes in IEMG with increases, decreases and no changes being reported. To explain these differences, Payton, (1974) proposed that agonists begin as an undifferentiated field. During learning they become differentiated into prime movers and assistive muscles. The prime movers maintain their level of IEMG and the assistive muscles decrease their level. Lamb, et al. (1974) attempted to extend this propoundment to antagonists, but they were unable to support Payton's proposal. A consistent finding in these studies has been a decrease in movement time as skill is acquired. Because IEMG is a function of amplitude, frequency and time, three studies were reanalyzed--normalizing the data to time. Results indicated that the agonists made no change in IEMG. The antagonists significantly increased their activity, denoting that they may be an important controller in skill acquisition (Hobart, 1979).

Using Poulton's classification, the swing phase of gait progressed under stable and predictable environmental conditions and was considered a closed skill. The stance phase, particularly the first part, was considered an open skill. Because most of the previous investigations involved closed skills, the swing portion of gait was chosen. The purpose of this study was to investigate the IEMG of selected antagonistic muscles in the lower extremity during acquisition of a closed novel motor skill.

## METHODS

Twelve female and eight male normal adults, with a mean age of 22.9 years, served as subjects. The IEMG of the right gluteus maximus, long and short head of the biceps femoris, and medial hamstrings (semimembranous and semitendonous treated as one muscle) were simultaneously recorded using surface electrodes. After cleansing and abraiding the skin, to insure impedance measures of less than 4k ohms, 5mm gold cup bipolar electrodes were placed 1.5 cm on either side of the motor point of each muscle. Electrical activity of the muscles was amplified and recorded by a Grass 7B polygraph with 7P3B amplifiers and simple integrators. Sensitivity was set at 200 $\mu$V /div., the frequency response was 3Hz to 10KHz, and there was a 0.2 second time constant of integration.

A pre-test consisted of walking a distance of 6m while wearing a pylon which could be adjusted for subject's height and shoe size. The socket, made of orthoplast, opened posteriorly to allow fitting to the thigh and was padded at the knee for comfort. Straps secured the pylon to the thigh. Two ribbon switches attached to each shoe and connected to the polygraph recorded swing and stance phases of the gait cycle. Water-color pens attached to the mid-posterior aspect of each shoe recorded feet contact on a paper walkway.

Experimental procedures were explained and each subject was allowed one minute to practice standing. After an assisted 1.5m walk to the starting line, the pre-test data were gathered as each subject walked at his/her own pace for 6m. This was followed by a practice session which consisted of walking 80m with a one-minute rest period after each 12m. Instructions were given to help subjects acquire the skill. Using pre-test protocol, post-test data were gathered immediately after practice.

Acquisition of the task was determined by a reduction in step width, an increase in swing phase of the limb without the pylon, and more equal step length. These variables were tested statistically by a paired t-test. EMG data were converted to units by planimetry; units being a function of muscular activity and duration of movement. Because the phases of gait cycles varied greatly, it was necessary to normalize the data by dividing the IEMG by the duration of its activity yielding electrical activity per second. These data were then analyzed using a Hotelling $T^2$ multivariate test. Post Hoc analysis of significant data was accomplished by the Confidence Interval method. A .05 level was accepted for statistical significance.

## RESULTS

Kinematic variables were all significant at the .05 level; this indicated that through adequate learning the subjects had improved toward a normal gait. Statistical analysis of IEMG showed significant changes in muscle activity. Post Hoc analysis tested the confidence intervals for each muscle and revealed the main source for changes in IEMG as an increased activity of the short and long head of the biceps femoris and the inner hamstrings. The gluteus maximus made no change.

A significant change toward a more equal step length was the result of an increase in the left step length rather than a change in the side with the pylon. At the same time, there was a decrease in swing phase duration on the pylon side. Because IEMG was corrected for movement time deviation, it seems reasonable to attribute the increased muscle activity as a requisite to slow the increased velocity of the swing-through movement. As skill was acquired, subjects moved with more confidence; they were able to employ the antagonistic muscles in a more positive manner to control the movement by placing the foot in a desired position at the proper time.

Lack of change in the gluteus maximus was expected, as there is ample documentation that its participation in normal gait tends to be minimal. The significant increase of the IEMG of the remaining muscles does not support Payton's proposal as stated for agonists. Based on present data, it could be stated that in closed skills the prime antagonists tend to increase their IEMG, whereas the secondary antagonists show no change or decrease.

Results to date, along with reanalyzed data by Hobart, may offer a hypothesis that the antagonist's increased activity tends to be a major consistent variable during acquisition of a novel closed motor skill. Changes in the myotemporal variables, along with more efficient operation of the muscles, allow the agonists to provide necessary displacement and velocity changes. The concomitant increase of the antagonist's activity provides final control in placing the limb in a desired position at the appropriate time.

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NOTES

# ANALYTICAL METHODS

## Session 8B

# ADAPTIVE FILTERING IN RESPIRATORY ELECTROMYOGRAM PROCESSING

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## INTRODUCTION

Noise cancelling devices include fixed and adaptive filters. The former require prior knowledge of both signal and noise, and cannot be applied if the two bandwidths overlap. Adaptive filters utilize inputs from a primary and a reference channel, the former containing signal plus noise, the latter an estimate of noise alone. The filter modifies continuously the reference input and subtracts it from the primary until a stable output is reached which offers the best estimate of the desired signal uncontaminated by noise (1). In this study we investigated the use of adaptive filtering to minimize 60 Hz noise and electrocardiographic (EKG) artefact in electromyograms (EMG) recorded from the external intercostal muscles by means of electrodes applied to the chest wall.

## EXPERIMENT

External intercostal EMGs were recorded during spontaneous breathing via bipolar electrodes applied over the second right intercostal space, parasternally. A reference signal was obtained from electrodes applied to the acromion process and the anterior superior iliac crest on the left side. The primary and reference signals (DC - 10 KHz) were recorded on magnetic tape, together with the airflow signal derived at the mouth from a pneumotachometer (Fleisch No. 2).

## DATA PROCESSING

During preliminary studies it was established that greater than 90% of the power in the intercostal myoelectric signal is associated with frequencies below 500 Hz. In the present experiments the primary and reference signals were, therefore, lowpass-filtered at 500 Hz prior to being digitized at 1200 Hz. Discrete Fourier transforms were then performed on 417 ms segments of data before and after processing through linear and non-linear adaptive filters implemented on an IBM 370 computer. Those time windows that contained a cardiac artefact were centered on the QRS complex. The effectiveness of the filtering was assessed by: (i) visual inspection of the power density and amplitude spectra of the filtered and unfiltered signals (ii) computing the average residual power after filtering of 10 EKGs recorded between breaths and (iii) comparing the centroid frequency ($f_c$) of the EMG spectrum (2) for the bandwidth 25 - 500 Hz, before and after filtering.

## RESULTS

Spectral analysis revealed that, in our recording circumstances, line power was associated with artefactual peaks at 60 Hz and also its odd harmonics (i.e., 180, 300 and 420 Hz). These were removed with equal success by the linear and non-linear filters. However, the non-linear filter was notably more effective in minimizing cardiac noise, reducing it to 0.15% of the total EKG power in the primary signal, compared with 1.72% which remained after processing through the linear filter. The significance of this discrepancy is illustrated in Table 1, which compares $f_c$ values for four successive windows (listed from left to right) during a typical inspiration in one subject. The rows from above down indicate the position of the window in inspiration, by number; whether or not an EKG was present; and $f_c$ values for the primary signal (P), and after processing through the linear (L) and non-linear (N) filters.

TABLE 1

EFFECT OF FILTERING
ON $f_c$ VALUES (Hz) DURING INSPIRATION

| | WINDOW | | | |
|---|---|---|---|---|
| | 1 | 2 | 3 | 4 |
| EKG? | NO | YES | NO | YES |
| P | 142 | 78 | 151 | 97 |
| L | 160 | 133 | 155 | 144 |
| N | 156 | 151 | 154 | 157 |

These results indicate that adaptive filtering with a non-linear device yields values for $f_c$ which are consistent from window to window, and agree with similar values computed from the primary signal late in inspiration in the absence of an EKG, when a favorable signal-to-noise ratio minimizes the effect of 60 Hz interference. Signal distortion by the filter was minimal, and limited to frequencies below 50 Hz.

## DISCUSSION

Spectral analysis of respiratory EMGs has demonstrated that the power of the EKG and its harmonics is such that cardiac artefact must be eliminated if the EMG signal is to be analyzed carefully (2). Until now this prerequisite has inevitably dictated the simultaneous loss of substantial portions of the myoelectric signal with the information it contains. Similarly, recent studies of respiratory EMGs recorded over the chest wall required the use of a Faraday cage to minimize 60 Hz noise (3). These considerations seriously limit the application of frequency domain analysis to the clinical problem of detecting respiratory muscle fatigue. The results of this study suggest that adaptive filtering may offer a solution to these and similar noise problems in bioelectric signal processing. However, the filter's performance is critically dependent on the reference electrodes detecting every component of the EKG, and some of our early results suffered from the absence of a significant T-wave in the reference signal. This problem was easily resolved by more careful electrode placement, which was limited to sites of bony prominences to minimize the possibility of recording extraneous muscle activity in the reference signal. Finally, although these developmental studies have been performed with the help of major computing facilities, preliminary analysis indicates it should be possible to implement the same approach utilizing far less expensive equipment.

## CONCLUSION

Adaptive filtering is a technique that has rarely been applied to the problem of noise cancelling in bioelectric signal processing. During the present investigation we established that it was possible to remove interference associated with line power and also the cardiac artefact from respiratory EMGs, using such an approach, without significantly distorting the myoelectric signal.

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## ACKNOWLEDGEMENT

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COMPUTERIZED INTERPRETATION OF INITIATION AND RELAXATION OF EMG RECORDED DYNAMIC MUSCLE CONTRACTION

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## INTRODUCTION

A significant problem confronting researchers utilizing Electromyography (EMG) is an accurate and repeatable method of selecting the initiation and relaxation parameters of recorded dynamic skeletal muscle contraction (2) (3) (4).

The purpose of this study was to develop computer based criteria for establishing the initiation and relaxation points of EMG recorded dynamic muscle contraction. Presented in this paper is a method for determining these points directly from the digitization of the analogue data generated by muscle activity during high velocity human movement.

## RATIONALE

EMG recordings plotted during human, sport performance, movement patterns where dynamic muscle contraction and high velocity limb movement both occur are considerably different from EMG recordings taken during controlled isometric contraction experiments. During isometric data collection, EMG baselines are normally quiescent resulting in distinct points of initiation and relaxation of recorded muscular contraction. These points of contraction and relaxation are usually indisputable even to the inexperienced investigators eye. However, myograms taken of high velocity human limb movement rarely demonstrate a quiescent baseline just preceding an application of muscular force. Therefore, distinct parameters identifying the initiation and relaxation of the desired muscular contraction are not usually evident thereby providing easy analysis procedures. This lack of a steady EMG baseline during sport skill performance results from muscular tension which must be factored out before the actual contraction of muscular force application may be reliably analyzed. Pre-contraction, (and/or post-contraction), tension found on dynamic myograms normally stem from three potential sources: 1) griping an implement, 2) holding a static position just prior to performance, and 3) in a windup situation. Our laboratories primary variable of interest in the EMG analysis of human movement, is the activity of selected agonists during their application of force in movement production. Fundamental to the precise quantification of these dynamic muscle contractions is a reliable and unbiased mehtod for determining the points of initiation and relaxation of each contraction from its analogue data waveform. Most investigators sole criteria for establishing these parameters on a myogram is visual inspection. However, since no widely accepted standard establishes a criteria for this visual inspection, little unanimity exists among researchers in EMG as to what constitutes the initiation or relaxation of a recorded skeletal muscle contraction.

## METHOD

Myograms were taken during the execution of ice hockey snap shots, baseball throws, and sprint track starts.

Each experimental group consisted of four male subjects between 20 and 24 years of age, all highly skilled competitors in their areas of performance. Each subject completed three trials with four agonists, specific to each sport movement, producing EMG data. Data was processed by a totally automated program via the analogue to digital convertor (ADC) subsystem of an HS 4020 Computer. The ADC's are 10 bits wide and sample at 2000 $H_z$ coupled with a 1 $KH_z$ low pass filter which negates aliasing. Surface electrodes with the proximal electrode over each muscle motor point were connected from the subject to the amplifier by a flexible cable. The amplifier contains a differential input component with a high common mode rejection ratio plus variable gain and selectable low pass filters.

The raw EMG analogue data produced from the ADC's (or from magnetic tape) was deciphered into single precision floating point values and stored in separate disc files. With the analogue waveform reduced, $t_o$, the beginning point for data analysis (initiation of muscular contraction during force application) may be established.

Since RMS values have been previously demonstrated to be porportional to both muscle tension and the square root of the number of motor units firing this protocol was attempted as a technique to determine muscle contraction parameters (1). After RMSth was calculated the computer examined windows of 100 datapoints (50 msec.) starting from $t_o$. The establishment of two or more consecutive windows with an RMS greater than RMSth identified possible contraction initiation (S100). Similarly a window with an RMS less than RMS th identified possible contraction relaxation (E100). The actual initiation and relaxation parameters were established from these RMS windows and a computer examination of the peak amplitude just before and just after the points determined by the RMS values.

A complete operating system has been designed to allow any researcher to perform data collection, storage, analysis and output in conversational modes without previous computer knowledge.

## RESULTS

In all trials across all selected agonists computer determinations, of the initiation and relaxation of muscle contraction, using RMS as the defining criteria proved to be valid. The computer interpretations essentially agreed with or acceptable adjusted contraction parameters previously established by visual inspection (P=.01). Earlier approaches which included computer analysis of reversal points (peaks), amplitude deflections from the baseline and zero crossing failed to establish a reliable technique for solving the problem of muscle contraction qualification.

## CONCLUSION

A computer based interpretation of the initiation and relaxation of EMG recorded dynamic muscle contraction which also coincides with high velocity limb movement is possible through an application of RMS calculations to the collected data. We do not predict that our technique locates the exact point of muscle contraction or relaxation. However, this method does establish a protocol by which muscle contraction parameters may be consistently identified, thereby increasing data analysis precision.

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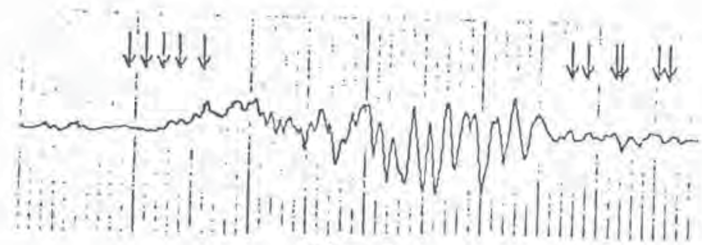

Fig. 1

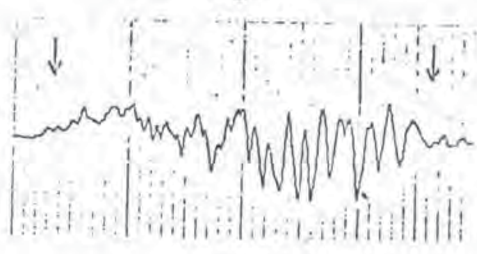

Fig. 2

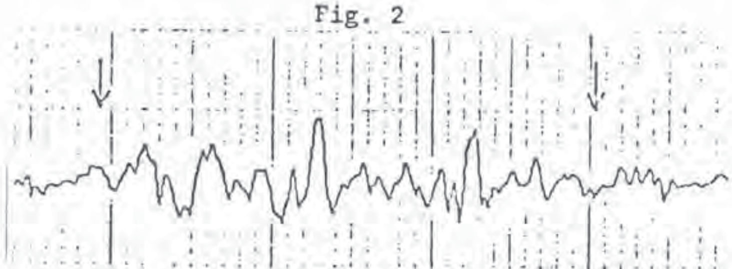

Fig. 3

Fig. 1 Analogue EMG data plot, Medial Head, Triceps Brachii (MHTB) during execution of an ice hockey snap shot. Possible initiation and relaxation contraction points are identified by the computer.

Fig. 2 Final computerized selected initiation (S100) and relaxation (E100) points of muscle contraction for MHTB based on RMS values and peak amplitude analysis.

Fig. 3 Flexor Carpi Ulnaris during same snap shot, contraction parameters identified, plus pre- and post- contraction tension problem illustrated.

QUANTITATIVE ELECTROMYOGRAPHY: A COMPUTARIZED METHOD FOR KINESIOLOGICAL STUDIES.

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## INTRODUCTION

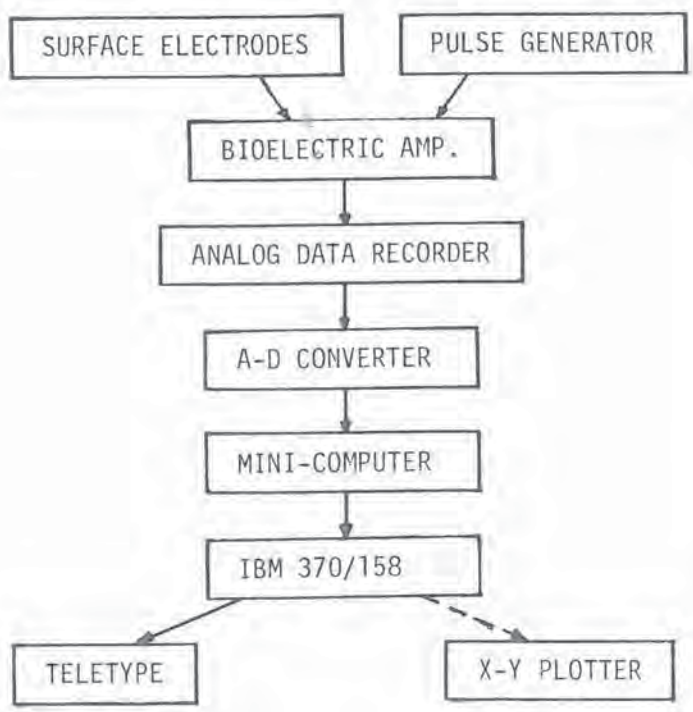
Evaluation of the electromyogram is traditionnally based on the examiner's subjective recognition of the pattern, and it is usually utilized to compare disabled patients to normal subjects in physical medicine, rehabilitation and related fields (Rose and Willison, 1967). In kinesiological studies, the electromyogram is utilized to study the role played by different muscles (Basmajian and Travill, 1961), or to determine the sequence and order of muscle activation (Lagassé, 1976 and Corser, 1974) in a given movement. EMG quantification is also conducted in terms of duration, amplitude and shape of the myoelectric potentials as well as in terms of pattern of recruitment (Eberstein and Goodgold, 1978). Thus EMG quantification has been a slow and laborious process.

Researchers (Willison, 1963, Hirose and Sobue, 1972 and Kopec and Hausmanowa-Petrusewicz, 1974) in clinical EMG have attempted to overcome these problems by utilizing computerized techniques of quantification. However, similar attempts in kinesiology still remain scarce. The purpose of this paper is to introduce an automated method, using digital computers, for surface EMG quantification.

## METHODOLOGY

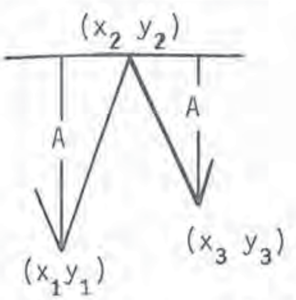
The instrumentation needed for EMG quantification is the following: Beckman silver-silver chloride bi-potential electrodes, a pulse generator, Hewlett Packard bioelectric amplifiers (8811A), an Ampex (FR 1300) eight-track analog to digital (A-D) converter, a Nova 1200 (Data General Corporation) mini-computer, an IBM (370/158) computer, a digital decwriter terminal and a Honeywell X-Y plotter. A block diagram (Figure 1) illustrates the arrangement of the hardware described above.

FIGURE 1: ARRANGEMENT OF THE RECORDING AND ANALYSING EQUIPMENT




The basic assumptions which underlie the quantification method presented in this paper are the following: first, the electromyogram is considered to be formed by potentials, each of which being shaped by a pair of an upward and a downward deflection (Hirose and Sobue, 1972); and, second, every significant deflection is at least 100 µV in amplitude (Willison, 1963). Considering the memory content of the minicomputer (16K), as well as the reliability of the reproduced wave form, the sampling interval selected to convert the EMG into digital values was 500 µsec. The pulse which was generated initiated the A-D conversion and was used to synchronize all EMG recordings. One second samples were thus analysed, the EMG being represented by 2000 X and Y coordinates, which were used as input for the IBM 370/158. A computer program was then utilized to quantify the upward and downward deflection of each potential, rejecting non significant deflections of less than 100 μV in amplitude, according to one of the precited assumptions. The modified electromyogram is then processed in reference to the vertical axis in terms of amplitudes (A), mean amplitude, standard deviation for the amplitudes and number of potentials (# pot), and, in reference to the horizontal axis in terms of intervals (I), mean interval and standard deviation for the intervals, as shown on Figure 2.

FIGURE 2: ANALYSIS OF THE MODIFIED EMG.




$A = y_2 - y_1$

$I = (x_2 - x_1) \times 500$

$\# \text{POT} = \dfrac{\text{Number of A}}{2}$

## APPLICATIONS

This method can be applied in several studies involving electromyographical data. In studies concerned with the respective contribution of several muscles in a given movement, the method described in this paper enables to accurately and rapidly quantify the activity of each muscle investigated (Hirose et al., 1974). It can also be utilized in studies concerned with the temporal and sequential order of muscle activation (Lagassé, 1976 and Lagassé et al., 1978) or in studies dealing with fractionated reaction time components (Lagassé and Hayes, 1973 and Clarkson and Kroll, 1978) and fractionated reflex latencies (Hayes, 1972).

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# A FACTOR - OF - FIVE IMPROVEMENT IN MYOELECTRIC SIGNAL PROCESSING

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## INTRODUCTION

Despite the wide attention given to the use of myoelectric activity for providing a quantitative estimate of muscle activity, the fidelity achieved using common myoprocessing techniques is poor. Work to date on myoprocesser improvements has been somewhat piecemeal, lacking coherent theoretical underpinning. This paper reports briefly on a project whose outcome was (a) the introduction of a novel myoprocessing technique; (b) the mathematical derivation of an optimal myoprocessor; and (c) the experimental verification of the efficacy of this processor.

## MULTI-CHANNEL MYOPROCESSING: A NEW APPROACH

The basis of the new processing technique and the motivation for the mathematical techniques used are provided by considering the physiology and electrophysiology of muscle. The mechanical output of the muscle is a combination of a large number of individual chemico-mehcnaical events. Similarly, the electrical activity of the muscle is a combination of the individual electrical events in the muscle. In both cases the macroscopic activity of the muscle is a weighted sum across space and time of a large number of individual events, but the weighting in the electrical case is radically different from the mechanical case. This is principally due to tissue attenuation which limits the pick-up region of the electrodes. In effect, the electrodes "look at" only a sample of the total population of active muscle fibres. As a result, a spatio-temporal sampling artifact ensues which is manifest as large-amplitude, low-frequency "noise" contaminating the myoprocessor output. Elimination of this artifact requires increased sample size. To accomplish this the new technique of combining multiple channels of myoelectric activity into a single myoprocesser output was proposed.

## MATHEMATICAL ANALYSIS

To determine the best way of processing each channel and combining all channels, a functional model of myoelectric activity under isometric conditions was formulated [1]. The model describes myoelectric activity as a random process whose instantaneous amplitude distribution for a fixed level of muscle activity is Gaussian. As myoelectric activity is a sum of conditionally independent events, the Law of Large Numbers justifies this assumption and experiments verify it. The amplitude distribution is assumed to be modulated multiplicatively by a static function of muscle force. In general, myoelectric activity is a non-stationary process as muscle force may vary, but as the frequency components of muscle force are almost an order of magnitude lower than those of myoelectric activity, the process may be described as stationary over the time available for processing. The shape of the power spectral density of myoelectric activity is assumed to be constant, independent of force, and further, is assumed to be a rational function of frequency. These assumptions permit myoelectric activity to be modelled as white noise passed through a linear, constant-coefficient filter, representing a combination of the frequency content of subcutaneous myoelectric activity and the filtering effects of transmission through tissue and detection by electrodes.

From this model an optimal estimator of muscle force based on observations of myoelectric activity can be derived. Because the frequency components of the desired signal, muscle force, differ from those of the input information, myoelectric activity, the problem is fundamentally nonlinear, and the well-developed linear filtering theory can not be applied even as an approximation. The approach taken was to combine state-space methods and statistical decision theory to define the optimum myoprocessor as the maximum likelihood estimator of muscle force [1].

The resulting optimal processor for the single-channel case is as follows:

Because the mathematical model assumed that the myoelectric activity observed at the electrodes was originally white noise, the first stage of processing is a prewhitening filter whose purpose is to convert the observed signal back into white noise. However, white noise is a convenient mathematical fiction, not a real process, and prewhitening must be approached with caution to avoid merely amplifying bacground noise. The second stage is variance estimation. The signal is squared and averaged. The third stage is "relinearization," the inversion of the assumed static relation between force and signal amplitude. A derivable property of this processor is that the output signal-to-noise ratio is constant, independent of force. [1] Note that this mathematical analysis encompasses and places in perspective almost all of the recent work on improving myoprocessors--prewhitening, averaging, non-linear demodulation.

Solution of the multiple-channel case requires some additional simplifying assumptions about the covariance matrix relating the activities of the individual channels. These are (a) that the form of the static relation between force and myoelectric signal amplitude be the same for all channels and (b) that the correlation between channels be constant, independent of force. These are the spatial equivalent to the earlier assumption of constant power spectral density shape.

The resulting optimal multi-channel myoprocessor prescribes spatial prewhitening of the myoelectric activity. This is an appropriately scaled linear recombination of channels to produce equal-variance, uncorrelated signals. These are then combined and processed as in the single-channel case.

## EXPERIMENTAL VERIFICATION

To verify the efficacy of the optimal myoprocessor, four channels of surface myoelectric activity plus corresponding isometric muscle force were simultaneously recorded from the biceps brachii muscle of normal and amputee subjects. This data was processed digitally off-line to extract the required signal parameters (statistical bandwidths, covariances, etc.) and to compare the effects of various stages of the optimal processor. Knowledge of the statistical bandwidth permits prediction of processor performance. Agreement of experimental results with theoretical predictions was excellent. For example, signal-to-noise ratio of a single channel of myoelectric activity rectified and first-order, low-pass filtered with a 250 ms. step-response rise time (a common processing technique) was predicted as 8.70 and measured as 8.56. This excellent agreement strongly justifies description of surface myoelectric activity as an amplitude-modulated Gauss-Markov random process

The signal-to-noise ratio obtained by optimally combining four channels of activity, averaging with a 250 ms. step-response rise-time, and half-power "relinearizing" (equivalent to assuming the muscle force/myoelectric activity relationship to be square-law) is 46.27. This is 5.41 times better than the performance of the common myoprocessor and corresponds to a peak-to-peak deviation about the mean of $\pm$ 6%.

As a final test the optimal myoprocessor was implemented using analog hardware to provide an on-line real-time estimate of muscle force. The performance of the on-line myoprocessor is shown in Figure 1, which includes a recording of the corresponding isometric muscle force. The extent of the improvement achieved is quite striking.

## ACKNOWLEDGEMENTS

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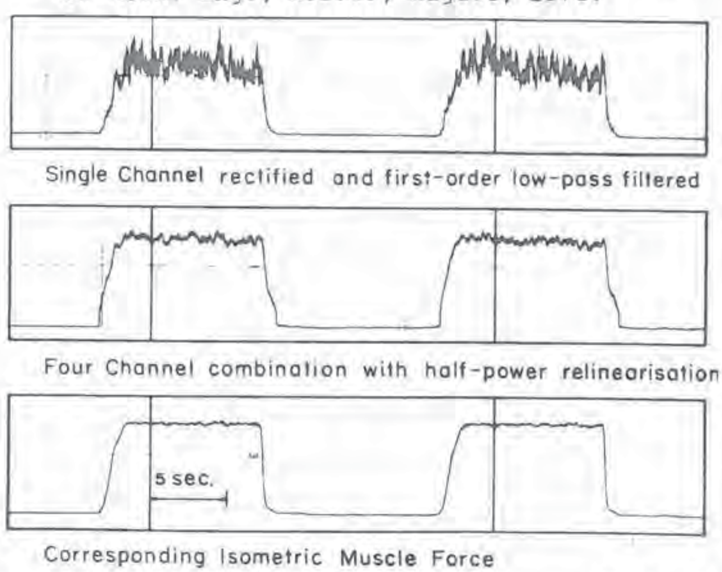


Fig. 1. Output of common myoprocessor (Top) and improved myoprocessor (Middle) vs. muscle force (Bottom).

# THE EFFECT OF ELECTRONIC PROCESSING ON THE QUANTITATIVE EMG SIGNAL

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## INTRODUCTION

The collection and evaluation of electromyographic (Emg) data from active muscle tissue is a commonly used step in the biomechanical analysis of human motion. Particularly when qualitative relationships are sought, Emg data in direct form can be useful (Basmajian, 1978; Kelley, 1971). Attempts to produce quantitative measures of Emg muscle activity usually involve electronic processing of the Emg signal including rectification, RC integration, and voltage or timing discharge networks (Tursky, 1964). However, when this is done, some modification of the original Emg signal will result due to factors such as frequency filtering and phase shifts inherent in these processing circuits (McLeod, 1973).

Investigators using Emg data are faced with the question of what influence electronic data processing will have on their final results and conclusions. Much of current Emg literature reports the results of RC integrating networks, and many of these have used different time constants (values of RC).

The purpose of this study was to compare an Emg signal processed by rectification only with the same signal processed by integration using several different time constants. Resulting measures of total Emg activity and the location of Emg peaks and dips were statistically analyzed.

## PROCEDURES

Emg data were collected from 11 normal adult subjects via Beckman miniature surface electrodes placed over the biceps brachii. Electrode output was routed through a Hewlett Packard 8811A Bioelectric (Differential) Amplification System (freq. resp. = 5 - 3K Hz), and data were then recorded by a Hewlett Packard 3968A FM Tape Recorder. After being recorded, the Emg data were processed with a Coulbourn Modular Data Reduction and Recording System for rectification and/or integration. The subject movement involved simultaneous elbow flexion and shoulder extension as described by Alon (1978). Initiation and termination of the movement range were indicated by signals from appropriately aligned photocells.

Data from a movement lasting 0.5 second were used for this analysis. The Coulbourn processing system converts data to digital form through a voltage controlled oscillator and prints out summed values over a minimum interval width of 0.1 second. Therefore, five data points were determined for each Emg record. The data from each subject were replayed from the FM tape and treated five different ways: rectification alone, integration with time constants of 0.02, 0.03, 0.1, and 0.2 second.

## ANALYSIS AND RESULTS

A two factor (both repeated) analysis of variance was performed on the data with a program available at the University of Maryland Computer Center. The first factor was treatment with five levels (A1 = rect.; A2 = 0.02,RC; A3 = 0.03,RC; A4 = 0.1,RC; A5 = 0.2,RC) and the second factor was time, also with five levels (B1 = 0.0-0.1; B2 = 0.1-0.2; B3 = 0.2-0.3; B4 = 0.3-0.4; B5 = 0.4-0.5 second). To be able to compare total Emg (area under each curve) with like units, each Emg record for each subject was normalized to the value recorded for the first time interval (B1). In this way all curves were made to start from the same point. The five treatment curves averaged for all subjects are shown in Figure 1.

Both data treatment and treatment-time interaction were found to have significant effects ($P<0.01$). Student-Newman-Keuls post hoc tests were performed, and the results are summarized in Tables 1 and 2.

## DISCUSSION

Since each time interval in the data record is of constant width (0.1 second), the total area under the curve is proportional to the total normalized Emg values for each treatment. These treatment totals are what the analysis of variance compares, and, therefore, the A factor shown in Table 1 is representative of the total Emg over the 0.5 second interval. There is no significant difference in total Emg as determined from the rectified data and integrated data with time constants of 0.02 or 0.03

second. However, integrating the data with time constants of 0.1 or 0.2 second resulted in significantly higher total Emg.

The interaction effect may be used to determine the location of peaks and dips in the data by comparing the data across time for each of the types of treatment. Examination of Table 2 reveals several interesting facts. The last time interval (B5) is clearly a dip in the data for the rectified and the two shortest time constant treatments. The peak in the data is at or near B2 for these treatments although it is not pronounced for the rectified treatment. For the longer time constant treatments (RC = 0.1, 0.2) the data peak shifts to the right as evidenced by the ascendance of values at B3 and B4 in the hierarchy. In fact, this time shift tends to further obscure the location of the dip in the data.

## CONCLUSIONS

When processing Emg data to obtain quantitative relationships, consideration should be given to the use of rectification alone instead of rectification and RC integration. Rectification appears to be an adequate technique for determining total Emg activity providing that units are kept consistent. In fact, rectification alone more closely agrees with short time constant (fast response) integration than does integration with long time constants (slow response). When curve shape is of primary importance consideration should be given to the fact that integrating with short time constants produces a slightly more distinct pattern than rectification alone. However, integrating with long time constants results in significant distortion of the curve shape.

If integration with a time constant of 0.02 second is used as a criterion measure for quantitative Emg, statistically similar results may be obtained by using rectification alone. Integration with time constants longer than 0.1 will not yield results in agreement with the criterion measure, and, in fact, should be considered less desirable than simply rectifying the Emg data and appropriately adjusting units.

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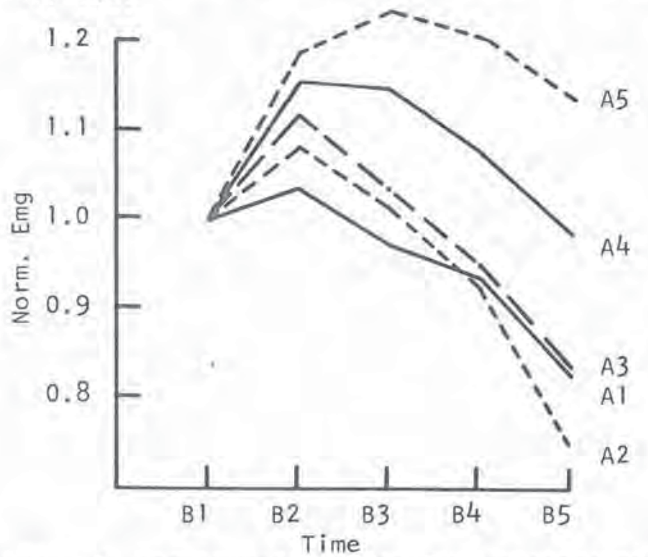


Figure 1. Normalized Emg for five treatments.

A2 A1 A3 A4 A5

Table 1. Total Emg compared. Values on common underline are not significantly different.

A1: B5 B4 B3 B1 B2

A2: B5 B4 B1 B3 B2

A3: B5 B4 B1 B3 B2

A4: B5 B1 B4 B3 B2

A5: B1 B5 B2 B4 B3

Table 2. Emg compared across time for each treatment.

## KEY TO FIGURE AND TABLES

| | |
|---|---|
| A1 = Rect. | B1 = 0.0 - 0.1 second |
| A2 = 0.02,RC | B2 = 0.1 - 0.2 second |
| A3 = 0.03,RC | B3 = 0.2 - 0.3 second |
| A4 = 0.10,RC | B4 = 0.3 - 0.4 second |
| A5 = 0.20,RC | B5 = 0.4 - 0.5 second |

NOTES

# MOTOR UNIT PROPERTIES I

## Session 9A

THE MUSCLE UNITS OF CAT TIBIALIS POSTERIOR: CLASSIFICATION BASED ON UNIT NEUROMECHANICAL PROPERTIES AND WHOLE MUSCLE HISTOCHEMISTRY

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## INTRODUCTION

Glycogen depletion studies have shown that muscle fibers of single cat medial gastrocnemius (MG) muscle units are histochemically homogeneous and that physiological properties of a unit are consistent with histochemical properties of its constituent fibers (Burke et al., 1973, Burke and Tsairis, 1973). MG was initially said to have three distinct muscle unit types: FF (fast-contracting, fast fatiguing); FR (fast-contracting, fatigue resistant) and S (slow-contracting, fatigue resistant). Units with physiological properties intermediate between the FF and FR type units were also encountered and subsequently designated F (int.) by Burke (1975). This classification scheme has also been found applicable to three other muscles of the cat hind limb: tibialis anterior and extensor digitorum longus (Dum, 1978) and flexor digitorum longus (Dum et al., 1978).

These studies suggest that the tetrapartite classification scheme might have quite general applicability to the muscle units of cat hind limb muscles. As a further test, we have examined another heterogeneous muscle. Our approach involved selecting a muscle which would permit representative sampling of its muscle unit population within a single experiment, and then comparing the mechanical properties of these units to the histochemical properties of muscle fibers in the same muscle. These experiments were supplemented by another series utilizing the glycogen depletion procedure on single units.

As a test muscle, we selected the cat ankle inverter, tibialis posterior (TP) which offered the advantages of having only approximately 60 motor units (Boyd and Davey, 1968) and having its efferent axons distributed over a sufficiently broad spinal root level (1½ segments, Romanes, 1951) that single alpha axons could be functionally isolated much more rapidly than in larger muscles. These features enabled us to analyze neuromechanical properties of 23-33% of the total unit population per experiment before preparing the muscle for histochemical analysis.

## METHODS

Single alpha motor axons innervating TP in deeply anesthetized adult cats were functionally isolated from ventral root filaments. Optimal muscle length for each unit's twitch or tetanus was used to determine: 1) peak tetanic force; 2) non-potentiated and potentiated twitch force and contraction time (CT); 3) the profile of unfused tetanus ("sag" test); and, 4) fatigue resistance. After these tests, a block of the muscle was frozen, serial-sectioned and stained for NADH-tetrazolium reductase and myofibrillar ATPase activity at pH 9.4.

## RESULTS

Measurements of muscle unit fatigability, "sag" susceptibility, peak tetanic force output and CT were found sufficient to classify 103 TP muscle units (6 cats) into four categories. A fatigue index (Burke et al., 1973) alone separated TP units into FF, FI and FR-S types. The "sag" test was essential to further separate type FR "sagging" from type S "non-sagging" units. A significant ($r = 0.93$, $P < 0.01$) correlation was found between the percentage of unit types and corresponding fiber types.

Table 1 summarizes the relative distribution and select properties of the four muscle unit types in cat TP and the distribution of their corresponding muscle fiber types.

## DISCUSSION

Distributions of mechanical properties in a representative sample of TP units and of histochemical properties of the muscles' fibers are compatible with Burke's (1975) tetrapartite muscle unit classification scheme. Furthermore glycogen depletion of FF, FR and S units in TP (manuscript in preparation) revealed that these units are composed of FG, FOG and SO fibers, respectively. No glycogen depletion data are yet available for FI units to determine if they are composed of the FI fibers identified in this muscle, a correlation suggested by:

Table 1. Classification and Physiological Properties of Muscle Units of Cat Tibialis Posterior[a]

| | | FF | FI | FR | S |
|---|---|---|---|---|---|
| % of each unit type | | 29.6 | 8.1 | 20.4 | 41.9 |
| | | FG | FI[c] | FOG | SO |
| % corresponding fiber types[b] | $\bar{x}$ | 38.6 | 5.7 | 17.1 | 38.5 |
| | S.D. | 3.8 | 2.2 | 4.5 | 4.7 |
| | Range | 32.3 - 42.4 | 2.6 - 8.5 | 10.7 - 24.0 | 35.1 - 46.4 |
| Classification Parameters | | | | | |
| Fatigue Index[d] | $\bar{x}$ | .05 | .58 | .87 | .98 |
| | S.D. | .05 | .15 | .09 | .10 |
| | Range | .00 - .17 | .31 - .83 | .74[e]- 1.09 | .81 - 1.21 |
| "Sag" | | (+) | (+) | (+) | (-) |
| Peak Tetanic Tension (g) | $\bar{x}$ | 108.1 | 46.0 | 33.8 | 10.1 |
| | S.D. | 47.5 | 17.9 | 19.2 | 5.4 |
| | Range | 26.4 - 219.8 | 25.0 - 87.4 | 12.0 - 94.9 | 2.2 - 23.5 |
| Potentiated Twitch CT (msec) | $\bar{x}$ | 31.2 | 34.5 | 34.2 | 49.1 |
| | S.D. | 5.2 | 5.7 | 5.2 | 9.8 |
| | Range | 23.0 - 49.6 | 28.8 - 42.8 | 23.0 - 47.6 | 33.6 - 76.2 |

[a] Total sample = 103 units

[b] Nomenclature of Peter et al., 1972

[c] FI fiber described in Crockett-McDonagh, 1979

[d] Fatigue index of Burke et al., 1973

[e] Unit with .74 fatigue index judged to be type FR on the basis of cumulative fatigue index of Reinking et al., 1975

1) similarities in unit and fiber type percentages; and, 2) intermediate physiological properties of the FI unit.

Of the four physiological properties used for muscle unit type classification, fatigability and "sag" were the most valuable for separation of unit types. Fatigue is an important parameter with respect to muscle unit and whole muscle function. "Sag" susceptibility, too, would seem to be just as fundamental to muscle mechanics since natural movements largely involve unfused muscle unit contractions.

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# EFFECT OF MOTOR CORTEX STIMULATION ON RECRUITMENT ORDER OF FAST AND SLOW ANKLE EXTENSOR MOTOR UNITS IN THE CAT

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## INTRODUCTION

In both animals and human subjects, many studies of the recruitment order of motor unit during a tonic and ballistic contraction have shown that individual motor units are recruited in such a stereotyped order that the units producing smaller and slower twitch are activated earlier than those yielding stronger and faster one. In contrast to these reports, the reversal of recruitment order has been recently reported in the rapidly isometric contraction of the human toe extensor (Grimby & Hannerz, 1977). Kanda et al. (1977) further demonstrated that the recruitment order could be changed by the cutaneous afferents. The present study was performed to test whether the reversal of fixed recruitment order is induced or not by the activity of motor cortex.

## METHODS

Cats were anesthetized with urethan-chloralose (30mg/3mg/kg). The animals were held to a rigid frame and then laminectomized in a lumbosacral portion. Nerves to the left hind limb were cut except those to the triceps surae muscle. Temperatures of body, back and muscle oil pools were maintained at 37 - 38°C by infrared lamps. A ventral root was dissected to the functionally isolated single axons to the medial gastrocnemius muscle (MG), lateral gastrocnemius muscle (LG) and soleus muscle (Sol) without cutting ventral roots. The functionally single axon was confirmed by all or none behavior of both the antidromic action potential and the twitch tension of its motor unit. The MG, LG and Sol were connected to the mechano-electric transducer for measuring the contractile properties of single motor units. Reflex discharges responding to muscle stretch or vibration at 160Hz and 40 - 160μm amplitude were recorded with bipolar electrodes on a thin filament.

In addition to the experiment of a single motor unit, another experiment was performed to examine whether the similar results could be obtained even during the simultaneous responses to the MG vibration of two or three axons in a natural filament of the MG nerve. The filament was cut at its entry into the muscle and placed on bipolar wire recording electrodes.

The right pericruciate cortex was exposed by craniotomy for electrical stimulation. Monopolar stimulating electrodes of stainless steel, insulated except at the very tip, were inserted in the contralateral pericruciate motor cortex to the depth of 2 - 3mm. Repetitive current pulse were used for stimulating it (rate: 400 impulses/sec, duration: 0.2msec).

## RESULTS

Fig. 1 shows the relation between the axonal conduction velocity and the reciprocal of twitch contraction time of 43 motor units. The clear tendency toward shortor contraction time in the motor units with higher conduction velocity was found ($r=0.182$, $p<0.001$). Since the conduction velocity is said to be significantly correlated with the motoneuron size, larger motoneurons would correspond to fast motor units and smaller ones to slow motor units.

Fig. 2A and 2B show the effects of cortical stimualtion on the MG stretch reflex of the fast unit with larger conduction velocity and the slow unit with lower one respectively. The reflex discharge of the fast unit (A) showed the phasic pattern and was facilitated by the repetitive cortical stimulation. On the contrary, the slow unit of record B exhibited the tonic pattern, and was inhibited by the stimulation. Of 43 motor units studied, the activity of all 11 fast-phasic units were facilitated, those of 3 and 4 fast-tonic units were facilitated and inhibited respectively. In the slow units, all 24 tonic firing units were inhibited but one phasic firing unit was facilitated. Namely, the larger motoneurons with higher conduction velocity have shorter contraction time, phasic firing pattern and are facilitated by the cortical stimulation, while the smaller ones with lower conduction velocity have the opposite properties (Fig. 1).

Further, the cortical control described above were confirmed by the second experiment (see Methods). It was found that the reflex discharge to vibration of the MG unit with slower conduction velocity was depressed at the same time that the MG unit with faster conduction velocity was activated by the cortical stimulation.

## DISCUSSION

The present experiments show that stimulation of the contralateral motor cortex predominantly facilitates the reflex activity of ankle extensor motoneurons with short contraction time and phasic firing pattern, while inhibits the reflex discharges of slow-tonic units. This finding is in accordance with the report of Endo et al. (1975) that fast and slow motoneurons in a pyramidal cat are subject to differential control by the stimulation of motor cortex. Recently, Kanda et al. (1977) has demonstrated that the MG motoneurons with relatively low threshold for reflex activation can be inhibited by a system of cutaneous afferents at the same time that higher threshold MG motoneurons are powerfully excited.

Together with the reports described above, the results obtained in this experiment suggest that a stereotyped recruitment order according to motoneuron size could be modified by the synaptic input systems different from primary spindle afferents.

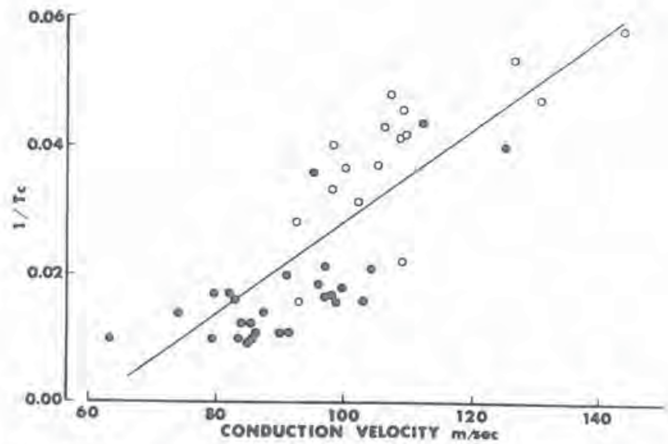


Fig. 1 Relation between the reciprocal of twitch contraction time and axonal conduction velocity in 43 triceps surae motor units. All of the motor units showed either facilitatory (open circle) or inhibitory responses (filled circle) to cortical stimulation.

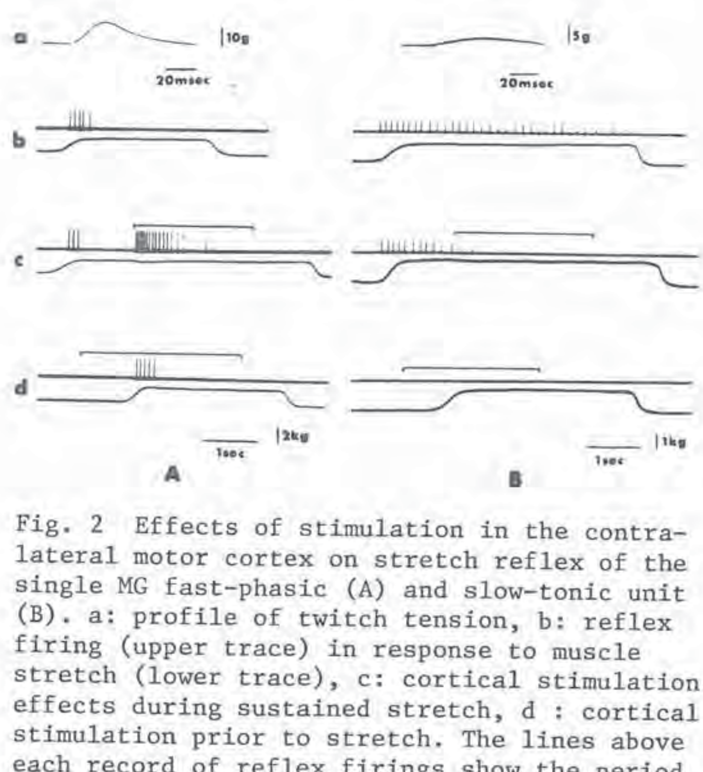


Fig. 2 Effects of stimulation in the contralateral motor cortex on stretch reflex of the single MG fast-phasic (A) and slow-tonic unit (B). a: profile of twitch tension, b: reflex firing (upper trace) in response to muscle stretch (lower trace), c: cortical stimulation effects during sustained stretch, d : cortical stimulation prior to stretch. The lines above each record of reflex firings show the period of cortical stimulation.

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INDIRECT ESTIMATION OF INNERVATION RATIO AND SPECIFIC TENSION OF MUSCLE UNITS OF CAT TIBIALIS POSTERIOR

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## INTRODUCTION

The intrinsic force-generating capability of muscle tissue (specific tension) has, historically, been thought to be approximately equal in animals of widely varying size, whether measured for whole muscles or single fibers (Close, 1972; Schmidt-Nielsen, 1979). For single muscle units of cat medial gastrocnemius (MG), however, Burke and Tsairis (1973) estimated a much lower specific tension value for type S units (0.6 $Kg/cm^2$) than for type FF (1.7 $Kg/cm^2$) or type FR (2.9 $Kg/cm^2$) units. The FF and FR indirect estimates of specific tension were in agreement with direct measurements based on glycogen depletion of single units. No direct measurement of specific tension is yet available for the type S unit of MG.

The present study involved indirect estimation of specific tension output of type FF, FR and S units of the cat ankle inverter, tibialis posterior (TP). This required: 1) measurement of mean muscle unit force and mean cross-sectional area of the unit's constituent muscle fibers; and, 2) estimation of mean unit innervation ratio (number of muscle fibers per unit).

## METHODS

Muscle units (N = 103) of TP from 6 cats were classified as type FF, FI, FR or S as based on measurements of their fatigability, "sag" susceptibility, peak tetanic force output and potentiated twitch contraction time (Crockett-McDonagh, 1979). Muscle fibers of TP were classified as FG, FI, FOG or SO (nomenclature of Peter et al., 1972; see Crockett-McDonagh, 1979, for FI fiber description), and cross-sectional area of these fibers was measured by planimetry from camera lucida drawings magnified 800X. The percentage of each fiber type was determined from counts of 2100-6500 fibers per muscle.

The FI unit was excluded from estimations of specific tension because both the unit and fiber type had a small representation in TP.

## RESULTS

Table 1 shows that the mean innervation ratio was estimated to be 547 fibers/unit for FF units, 363 for FR units and 393 for type S units. Mean specific tension was 4.9 $Kg/cm^2$ for FF, 3.6 $Kg/cm^2$ for FR and 1.5 $Kg/cm^2$ for type S units. Standard deviations of all these means were based on a propagation of errors analysis (Young, 1962). The three innervation ratios were not significantly different from one another ($0.05 < P < 0.10$), but the mean specific tension value for type S units was significantly different ($P < 0.05$) than the corresponding values for type FF and FR units.

In Table 1 standard deviations are also given for the four measured variables required to estimate mean specific tension. These include means for tetanic force, percentage of fibers and units of a particular type and fiber cross-sectional area. Values for force and for cross-sectional area are all significantly different ($P < 0.05$) from each other for FF, FR and S units. Mean percentage of FOG fibers is significantly less ($P < 0.05$) than that for FG and type SO fibers, while mean percentage of type S units is significantly greater ($P < 0.05$) than that for both FF and FR units.

## DISCUSSION

Of the variables on which innervation ratio is based (percentage of fibers and units of a particular type), the standard deviations of unit percentages are quite large, an error which is compounded in the propagation of errors calculation (Young, 1962) of the standard deviation of mean innervation ratio. A larger sample size than the 6 experiments on which these data were based would tend to reduce the variability in unit type percentages and might, consequently, add significance to the differences in mean innervation ratio for type FF as compared to FR and S units.

The finding for TP of significantly greater mean specific tension for type FF and FR compared to type S units is supportive of the finding of Burke and Tsairis (1973) for differences in specific tension between type F and S units of MG. The present findings and those of Burke and Tsairis (1973) present an interesting biophysical problem and indicate a need for further morphological, biochemical and histochemical studies on contractile protein content, activity and force-generating properties of different muscle fiber types.

Table 1. Indirect Estimation of Average Innervation Ratios and Specific Tensions in Tibialis Posterior Muscle Units[a]

| | FF (FG) | FR (FOG) | S (SO) |
|---|---|---|---|
| (1) % fibers of each type | 38.6<br>(4.5) | 17.1<br>(4.5 | 38.5<br>(4.7) |
| (2) No. of fibers of each type[b] | 9843 | 4361 | 9818 |
| (3) % units of each type | 29.6<br>(11.0) | 20.4<br>(11.4) | 41.9<br>(5.2) |
| (4) No. of units of each type[c] | 18 | 12 | 25 |
| (5) Innervation Ratio (line 2/line 4) | 547<br>(213)[d] | 363<br>(225) | 393<br>(67) |
| (6) Peak tetanic tension (g) | 109.9<br>(48.0) | 30.6<br>(13.0) | 9.6<br>(5.5) |
| (7) Fiber cross-sectional area ($\mu^2$) | 4078<br>(491) | 2342<br>(143) | 1619<br>(224) |
| (8) Specific unit tension (line 6/line 5 x line 7) ($Kg/cm^2$) | 4.9<br>(1.7)[d] | 3.6<br>(2.0) | 1.5<br>(0.6) |

[a] Values presented are mean (± S.D.)

[b] Based on 25,500 fibers in TP (Crockett-McDonagh, 1979)

[c] Based on 60 motor units in TP (Boyd and Davey, 1968)

[d] S.D. of innervation ratios and specific tensions derived from propagation of errors analysis (Young, 1962)

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# H-REFLEX RECOVERY CURVE-SINGLE FIBER EMG STUDIES

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The H-reflex recovery curve has been the subject of debate through the last three decades (Matthews, 1970). The recovery of single motoneurons after the primary inhibition period has not been studied before. In the study of the excitability and other properties of motoneurons, in normal and diseased states, accurate analysis of these recovery curves is of clinical as well as physiological importance. In this work the recovery curves of the H-reflex with identical stimuli were studied with single fiber EMG using the method of Trontelj (1973).

## MATERIAL AND METHODS

In this investigation the recoveries of 120 motor units of the soleus muscle were studied by single fiber electromyography (SFEMG) while recording the H-reflex (Sabbahi Awadalla, 1976). Twenty-eight normal young adults (18-31 years) were tested. The H-reflex was elicited by stimulating the posterior tibial nerve unifocally at threshold strength, and the soleus muscle action potential was recorded using a unipolar MEDELEC single fiber electrode (SF-25). To monitor the superficial EMG of the muscle, Copland-Davis (Palmer) surface electrodes were used. Recording from a single muscle fiber was identified by the criteria of Stålberg and Ekstedt (1973) and Trontelj (1973). Both visual and auditory monitors were used to verify single muscle fiber recording. Two identical square wave pulses (one millisecond in duration) were given with varying interstimulus intervals. The recovery of the motoneurons at $H_2$ was identified, and the inhibition period was noted.

## RESULTS

Results showed that motoneurons recover in an orderly way so that every neuron has a definite time of inhibition after which it would recover. Recovery of the potential was all-or-none, and most of the motoneurons recovered after 80-250 msec. (fig 1). However, some motoneurons had a late recovery (up to 2200 msec.). They were those neurons excited by low voltage pulses.

The period over which the recovery was less than complete, was called the recovery fringe, and ranged from 2-100 msec. At the beginning of this period the motoneuron recovered once in ten consecutive trials. However, at the end of the fringe period it was fully recovered. A linear relationship was noticed between the recovery time and the recovery fringe (fig 2). Those motoneurons which had an early recovery showed a short recovery fringe (2-15 msec.) while neurons of late recovery demonstrated a longer fringe time (100 msec.). A mild increase in the stimulus strength shortened the recovery fringe but not the recovery time. The recovery time of any motoneurons was consistent and reproducible throughout several tests, so long as conditions were not changed. This means that a partially recovered conventional H-reflex was always produced by the same motoneurons and not simply by the same proportion of motoneurons.

Conduction time as measured by SFEMG was compared to the surface EMG, and from this relationship a tentative classification of motoneuron types was developed (Freund et al, 1975). It was noticed that those motoneurons which recovered earliest were the more tonic while those with late recovery were the more phasic neurons. This indicates a higher excitability of the more tonic motoneurons to the Ia volley than the more phasic ones, a finding which is in accord with studies on cats by Henneman et al, (1965).

## DISCUSSION

The finding that most of the motoneurons recovered after 80-250 msec. indicates an excitable type of neuron. They are probably the tonic types which are heavily used in movement (Henneman, 1974). Kernell (1966) reported that the total membrane resistance of these tonic motoneurons is higher making them more excitable to external stimuli. These are the motoneurons which more easily are driven by the muscle spindles (Burke, 1973). It has been demonstrated that the soleus muscle is formed mainly of tonic type of motor units (Johnson et al, 1973). It seems that these motoneurons are less affected by the inhibitory mechanisms causing the primary inhibition period. Moreover, our results demonstrated phasic type of motoneurons in the soleus muscle. Those are the motoneurons showing a long inhibition period and a late and fluctuating recovery. They form a small percentage of the motoneuron pool of the soleus muscle and are less excitable to external stimuli.

The fact that some motoneurons do not recover before 3000 milliseconds stresses the importance of eliciting the reflex every five seconds, in the H-reflex methodology to ensure the recovery of all motoneurons.

The intimate relationship between the recovery time and the recovery fringe indicates that the fringe is a function of the excitatory state of the motoneurons more than the motoneuron itself. When the motoneuron is a highly excitable type (tonic) it showed a faster and more stabilized recovery, while those of low excitable type (phasic) were delayed and less stabilized.

## CONCLUSION

It was demonstrated that the H-reflex should be elicited every five seconds in order to obtain a full recovery of the motoneuron pool. Also, the soleus muscle showed a larger percentage of excitable tonic motoneurons. Motoneurons recover from the primary inhibition in an orderly fashion. Motoneuron types may be studied in man by single fiber EMG and H-reflex recovery curves.

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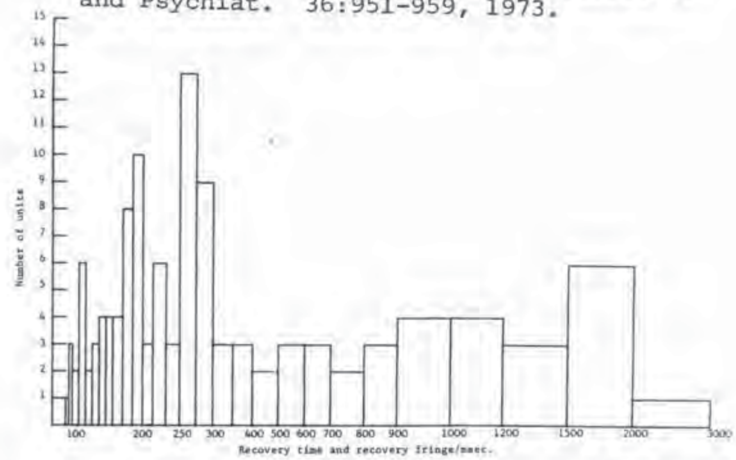


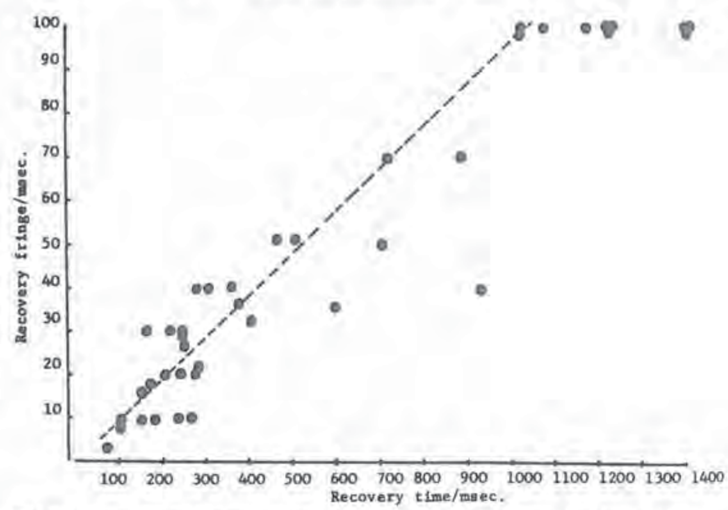


Fig 1: Recovery times and recovery fringe of 117 motoneurones.

Fig 2: Recovery time has an approx. linear relationship with the recovery fringe of various motoneurones. (40 units)


ACKNOWLEDGMENT: Dr. M. Sabbahi was supported by the gov't of the A.R.Egypt during this study.

NOTES

# MODELLING

## Session 9B

MODELING (AND VERIFICATION) OF THE MOTOR UNIT ACTION POTENTIAL TAKING INTO ACCOUNT FINITE MUSCLE FIBRE LENGTH.

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Chris W. Pool, Zoological Lab. University of Amsterdam, Amsterdam. The Netherlands.

## INTRODUCTION

The anatomical, physiological and/or pathological properties of a motor unit strongly determine the motor unit action potentials (muaps). The relations between the parameters of the motor unit and its muap can be studied by simulation. For such simulations a correct description of the extracellular potential field of a single muscle fibre, the anatomical structure of the m.u. and the geometry of the electrode are important.

Most of the muap simulation models are more or less based on the geometrical structure of the m.u. (George '70; Boyd et al. '78; Griep et al. '78), but these models did not take into account the finite length of the muscle fibres. For a single striated muscle fibre with a finite length Dimitrova ('74) described a model in which two depolarized zones were considered. For the further development of muap models it is also necessary to include the proper geometrical structure *and the finite muscle fibre length*. One of the first attempts in this direction was recently published by Dimitrov (1978). He studied the influence of the desynchronization in the fibre activation on the extracellular potential of a number of muscle fibres, but a spread in fibre and endplate positions was not taken into account.

## REFINEMENT OF THE MUAP MODEL.

The present paper deals with a refinement of our muap model (Griep et al. '78) to the finite fibre length. The original model was based on the superposition of single muscle fibre potentials of each fibre belonging to the motor unit. The parameters which characterize each fibre were specified i.e. the spatial position, the diameter, and a time dispersion concerning the arrival times of the single fibre potentials at the electrode. This time dispersion mainly depends on the position of the endplates, which are usually situated in a relatively small oblique zone.

The validity of the refined model will be tested by comparing simulated and measured muaps.

When finite muscle fibre length is introduced in the model two major aspects have to be examined: a) the geometrical structure of the muscle, b) the relation between the shape of the single fibre potential and the position of the depolarization zone on the fibre membrane.

Ad a. The Extensor Digitorum Longus muscle of the rat was chosen for investigation. In Fig.1 the mean geometry of this muscle and the simplified geometry used for the model is given.

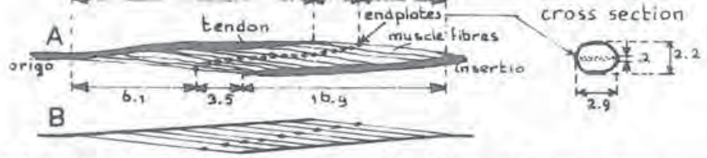


Fig. 1. A. Mean geometry of five EDL muscles (endplates identified by acetylcholinesterase). B. Model representation of the EDL muscle.

Ad b. With respect to the depolarization zone on the muscle fibre membrane, three phases can be distinguished: 1) the emergence of the action potential in the endplate region, 2) the propagation of the a.p. along the muscle fibre, 3) the disappearance of the a.p. at the transition of the muscle fibre in the tendon.

The extracellular single fibre potential was approximated by a modified tripole model (Rosenfalck, 1969).

Since the electrical conductivity of muscle is almost pure resistive it was assumed that the total membrane current at each moment during all three phases was zero (no charge accumulation in a resistive medium). This means that the tripole can be considered as two dipoles of which the "sinks" coincide. The membrane current (tripole) is time independantly coupled to the intracellular potential. If the membrane current is twice integrated along the fibre axis, the intracellular potential is obtained (see Fig. 2A). It was also assumed that the distances between the three poles are fixed during the phases of emergence, propagation and disappearance of the a.p. In Fig. 2B, 9 intracellular potentials are drawn which are present at successive time intervals during the three phases.

The emergence of the action potential (first phase) was described by a linear increment with time of the first dipole strength (corresponding intracellular potentials at t1, t2, t3 are drawn) followed by a linear increment

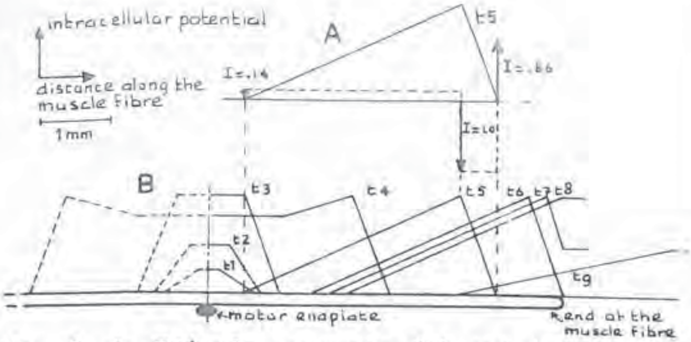


Fig. 2. A. Tripole currents (I) and intracellular potential. B. Intracellular potentials at successive points in time.

of the second dipole strength (see curves t4 and t5).

During the active propagation along the muscle fibre both dipoles have maximum strengths (see Fig. 2A and curves t5 and t6). If the potential diminishes at the end of the fibre the dipole strengths decrease linearly one after the other (curves t7, t8, t9).

## RESULTS.

We are testing the validity of this model by comparing simulated and measured muaps. Using the model a histochemically identified m.u. (Pool et al. '78) was simulated. The m.u. consisted of 58 fibres in a territory of about 1.5x1.5 mm2.

For a first verification, muaps were simulated and measured extraterritorially (on the muscle surface) in the endplate region, along the muscle fibres and in the region of the end of the muscle fibres. See Fig. 3.

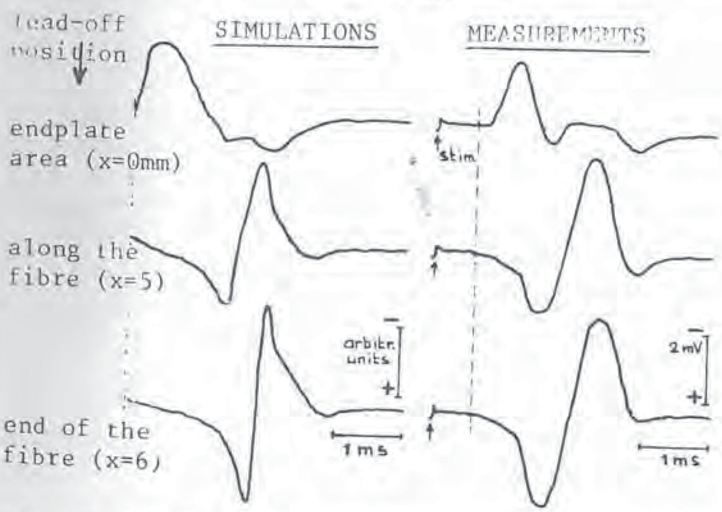


Fig. 3. Simulated and experimentally obtained motor unit action potentials.

## CONCLUSIONS.

These preliminary findings based on the quite simple assumptions described above indicate that there is qualitatively a good agreement between simulated and experimental results. The adaptation to the finite fibre length gives more realistic results when compared with other muap models. The presented extensions of the model enable the verification and also permit the study of all kinds of experimentally obtained muaps.

The verification of the model will be extended for intraterritorial muaps on the basis of the precise geometry of the muscle fibres of the m.u. in relation to the electrode. These results will also be presented.

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# MODELLING OF THE EMG RESPONSE TO APPLIED TORQUE PERTURBATIONS OF HUMAN FOREARM POSITION

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## INTRODUCTION

The advent of programmable force generators has made it convenient to study the use of proprioceptive information by human subjects during simple motor tasks. A linear analysis of the response of forearm muscles to pseudo-random torque changes applied at the elbow shows flexor and extensor EMG activity to be highly dependent on the kinematic variables of elbow rotation (angular position, velocity, and acceleration). Nonetheless, these linear models have some residual error and it seems appropriate to ask whether this error is primarily due to a nonlinear dependence on the kinematic variables or reflects a dependence on nonkinematic information or motor system nonstationarities.

## EXPERIMENT

A fast-response electric motor (time constant of 4 ms) was used to generate pseudo-random sequences of torque pulses at the elbow joint. Adult subjects were asked to resist these disturbing torques and to maintain their arm in a nearly constant position. The direction of these torques could not be anticipated by the subject. EMG activity of the biceps and triceps muscles was rectified and ensemble-averaged (intramuscular and surface electrodes yielded equivalent results). Angular position, angular acceleration, and applied torque were also measured. Each experimental record was the ensemble average of 10-20 consecutive trials. All averaged variables were adjusted to have a zero mean and were low-pass filtered with a cutoff frequency of 50 Hz.

## DATA ANALYSIS

The experimental averages were analyzed in two ways. First, statistical correlation methods were used to obtain the first two terms of a Wiener-Volterra representation of the averaged EMG signal:

$$\mathrm{EMG}(t_j) = \sum_i h_1(t_i)x(t_{j-i}) + \sum_i \sum_k h_2(t_i,t_k)x(t_{j-i})x(t_{j-k}) \qquad \text{Eq. 1}$$

where $h_1$ and $h_2$ are the first and second-order kernels, and x is the pseudo-random torque sequence. Second, least-squares methods led to parametric models of the averaged EMG signals in terms of the kinematic variables:

$$\mathrm{EMG} = b_0\Theta + b_1\dot{\Theta} + b_2\ddot{\Theta} + c_0\dot{\Theta}^2 + c_1\dot{\Theta}^3 + c_2\ddot{\Theta}^2 + c_3\ddot{\Theta}^3 \qquad \text{Eq. 2}$$

where $\Theta$ is the angular position of the elbow. The residual error was computed as the squared difference of the experimental averages and the model predictions. To increase the reliability of the parameter estimates, the data base for the fitting was comprised of three different pseudo-random sequences.

## RESULTS

The linear terms of Eq. 1 and 2 provided a good description of the ensemble-averaged EMG, with comparable residual errors. The best fit to Eq. 2 was obtained when each kinematic variable was provided with its own time delay (typically > 50 ms for position, 25 ms for velocity, and 45 ms for acceleration). The modelling error (residual error/EMG variance) was typically 15-30%.

This error was only slightly reduced by the inclusion of the second-order kernel ($h_2$), which conferred a small asymmetry in the response to flexion versus extension torques. Likewise, the modelling error was not significantly lowered by the addition of nonlinear terms in the kinematic variables ($c_0$-$c_3$), which resulted in a relative saturation of the reflex EMG response to large velocities and accelerations (see Fig. 1).

Much more pronounced changes in modelling error were obtained when one set of kernels or coefficients was used to describe the reflex activity under widely different experimental conditions or tasks. For example, a nonlinear scaling of the EMG response was present when the pseudo-random sequence was imposed upon different levels of background torque.

The modelling error was largely unaffected by increasing the number of trials per ensemble average beyond ten. However, significant variations in the parameter estimates could be observed if short averaging intervals were used, apparently due to nonstationarities in the reflex response rather than to a biased spectral content of the torque sequence.

## CONCLUSIONS

The residual error of linear models for the EMG response to pseudo-random sequences seems to depend more on motor system nonstationarities and the choice of operating point than on low-order nonlinearities when continuous disturbances are used. It is likely that the contributions by particular muscle or CNS nonlinearities are greater when discrete input disturbances are used, but such inputs are less physiological and unduly complicate the investigation of the adaptive behavior of the motor system.

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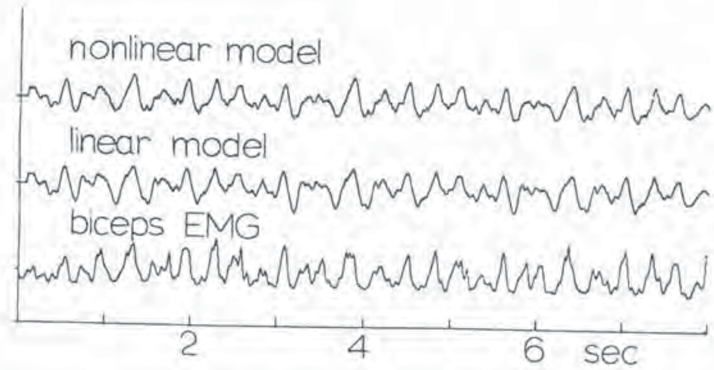


Fig. 1 Comparison of the ensemble-averaged EMG signal with the model of Eq. 2. The linear component (associated with coefficients $b_0$-$b_2$ of Eq. 2) is shown above the output of the complete nonlinear model.

## ACKNOWLEDGMENT

The authors are grateful to Dr. Carlo A. Terzuolo for his helpful criticisms and support. Financial assistance was provided by USPHS Grant NS-15018.

# MODELING OF THE ELECTROMYOGRAM: EFFECT OF MOTOR UNIT SYNCHRONIZATION

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## INTRODUCTION

The basic premise in developing a mathematical model for the interference electromyogram (EMG) has begun by synthesizing a sequence of pulses characterizing the activation of a single motor unit (MU) involved in an isometric contraction. A number of these sequences, representing the active MU population, are then summed to generate a synthetic interference pattern EMG. (Person & Libkind, 1967; Agarwal & Gottlieb, 1975; DeLuca & Van Dyk, 1975; Shwedwk, Balasubramanian & Scott, 1977). Various models have resulted from different formulation methodologies used to depict the observed phenomena relating to the muscle group under study.

Measurements of the energy density spectra of real EMG waveforms have consistantly shown that there is a shift of energy from the higher to the lower end of the spectrum during steady contraction and with fatigue. The basis for this spectral shift are not known but one suggested mechanism is MU synchronization which is known to occur under both of these conditions.

Synchronization can be considered as being the result of different patterns of MU activation. Three patterns of particular interest are: 1) simultaneous MU firing with identical Inter Spike Intervals (ISIs), 2) MUs firing in concert but slightly shifted in time, or 3) increases in the duration of MUAPs. All of these may occur as a function of the type, level, and duration of a contraction. The inclusion of such activity in an EMG model has been formulated to various degrees by Deluca & Van Dyk (1975), Lago & Jones (1977), and Libkind (1972), but simulation as such has been minimal.

## MODEL

Our model is similar to that of Parker et. al. (1977) if it were extended to the bipolar case, to that of Shwedwk et. al. (1977) if a random weighting coefficient were added, and to that of Calvert and Chapman (1977) if random shifts were considered as arrival latencies.

The motor unit action potential train (MUAPT) for the lth MU of the kth group $u_{l,k}(t)$ is commonly represented as the result of an impulse train $g_{l,k}(t)$ being passed through a system with impulse response $h_{l,k}(t)$ as shown in Figure 1. (Agarwal & Gottlieb, 1975). The function $h_{l,k}(t)$ defines the shape of the individual MUAP.

The interference EMG is modeled by summation of several groups of $n_k(t)$ MUAPTs with each group characterized by a different mode of MU activity. Three distinct modes of MU activity were studed: (k=1) synchronous (simultaneous), all units have same $g_{i,k}(t)=g_{j,k}(t)$, (k=2) synchronous with a random delay $\tau$ between trains $g_{i,k}(t)=g_{j,k}(t+\tau)$, and (k=3) asynchronous (different g).

The ISIs were assumed to follow either a Gaussian or a Weibull distribution. The MUAP was assumed to be either biphasic or triphasic (Agarwal & Gottlieb, 1975). The autocorrelation function and power density spectra were calculated and compared for different modes of synchronization (Trusgnich, 1978).

## SIMULATION RESULTS

For the first study, identical motor units were distributed among the three modes such that $n_1 + n_2 + n_3 = N$. The aim was to see what effects changes in the relative degree of synchrony within the active MU population had on the power density spectrum. Simulation with a biphasic MUAP, a Gaussian ISI distribution and N = 100 revealed no shifting of energy between high and low portions of the spectrum when various values for $n_1$, $n_2$, and $n_3$ were used.

The second simulation differed from the first in that motor units from mode 2 (synchronous with random time delay) had a lengthened MUAP duration.

Figure 2(a) shows the averaged power spectral density with all MU in group (3) and in Figure 2(b) 10, 52 and 36% of MUs are in the three groups respectively. A biphasic MUAP and

and Gaussian distribution for ISIs were assumed in this simulation. A significant shift in energy to lower frequencies is quite evident.

The third series of simulation were made with asynchronous motor units, some with increased MUAP duration. This simulation also produced a significant shift of energy to lower frequencies.

## DISCUSSION

Libkind (1972) reported the shifting of energies from higher to lower frequencies in the power density spectrum through the formulation of synchronous with a random time delay in MU activity. This formulation is similar to our first series of simulation which revealed little if any such spectrum shift. Only when the random time delay is combined with increased MUAP duration (second series of simulations) is spectral energy shift evident. But this shift is also evident in the third simulation as a result of increased MUAP duration alone.

These simulations suggest therefore that the observed shift in energy from high to low portions of the spectrum under various physiological conditions is not due to changes in the relative degree synchrony of the MU population but to changes in the MUAPs of the active population.

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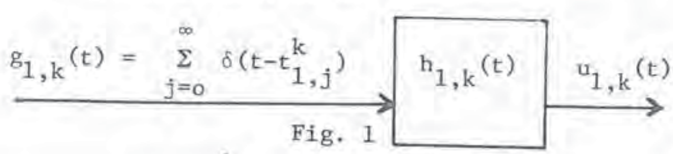


Fig. 1

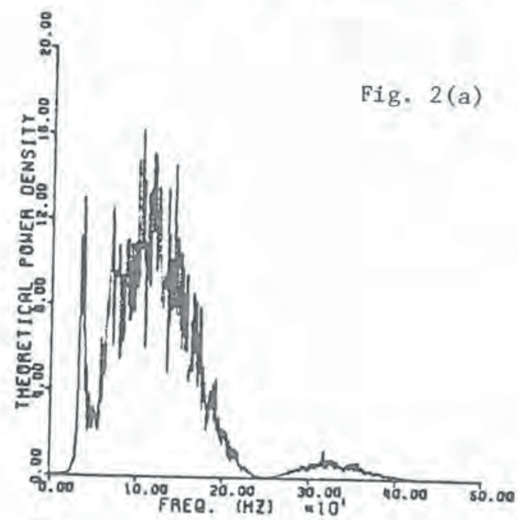


Fig. 2(a)

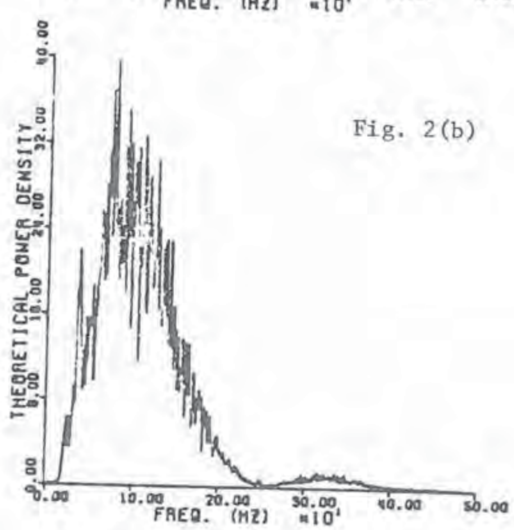


Fig. 2(b)

MATHEMATICAL ANALYSIS OF COMPOUND EMG SIGNALS

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## INTRODUCTION

A fundamental requirement imposed on Electromyography (EMG) for clinical use in the diagnosis of Neuromuscular Impairment, is the capacity of discriminating between Neurogenic and Myogenic Diseases. In the case of Needle Electrode EMG, the measurement of the Motor Unit Action Potential (MUAP) provides most of the desired information.

In this work a theoretical approach is presented which suggests possible use of Surface Electrode EMG as a diagnostic tool: preliminary results are discussed.

## THEORY

The model which will be described assumes the Motor Unit (MU) as the functional unit of the Neuromuscular System. The MU comprises the motoneuron and the muscle fibers innervated by the nerve axon.

The myoelectric activity of the MU is the Motor Unit Action Potential Train (MUAPT) whose properties have been studied by a number of authors (De Luca, 1975; Gath, 1974).

The j-th MUAPT may be represented by the pulse random process $u_j(t)$ defined as follows:

$$u_j(t) = \sum_{1k}^{N_j} h_j(t-t_jk) \qquad (1)$$

where:

- $h_j(t)$ is the impulse response of the j-th MU
- $t_jk$ the instant of firing
- $N_j$ the number of AP's in the MUAPT.

The compound signal s(t) obtained by summation of a number N of MU's can be expressed as:

$$s(t) = \sum_{1j}^{N} u_j(t) = \sum_{1}^{N} \sum_{1k}^{N_j} h_j(t-t_jk) \qquad (2)$$

A schematic model for the compound signal s(t) is illustrated in Fig. 1. If it is assumed that all MU's have the same impulse response h(t), than the model of the compound signal s(t) can be simplified as in Fig. 2. Its mathematical expression is that of the convolution integral:

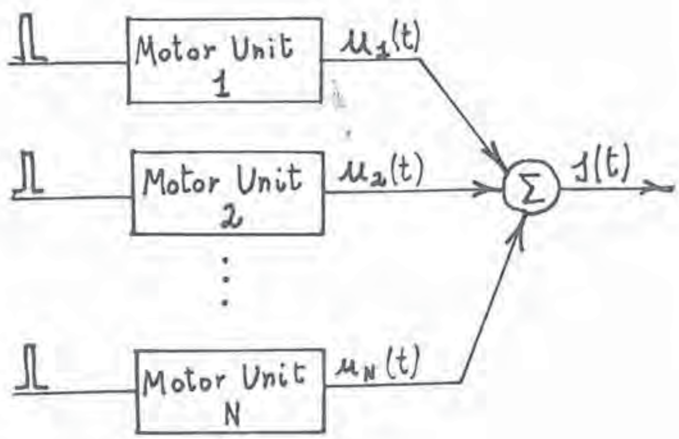


Fig. 1

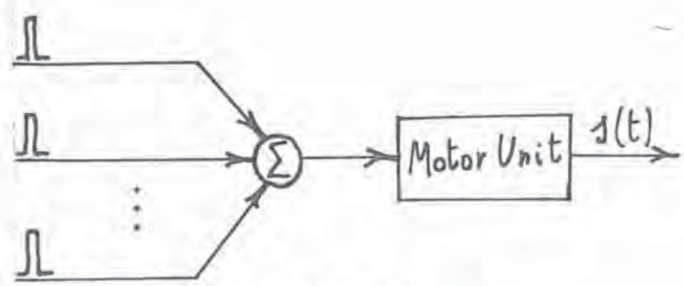


Fig. 2

Fig. 1 - A schematic model for the compound signal.

Fig. 2 - Simplified Model.

$$s(t) = n(t) * h(t) \qquad (3)$$

where n(t) is the time distribution of the active MU's.

Performing the Fourier Transform of expression (3) one obtaines:

$$S(\omega) = N(\omega) \cdot H(\omega) \qquad (4)$$

On the basis of Physiological Knowledge it is suggested that:

- Myogenic Pathologies be responsible for the shape of the dominant elementary MUAP, that is $H(\omega)$.
- Neurogenic Pathologies be responsible for the distribution $n(t)$, that is $N(\omega)$.

The proposed method aims at calculating the time distribution of the active MU's in the case of normal subjects, utilizing eq.(4).

## EXPERIMENT

The Biceps Brachii Dx of 10 young healthy subjects was tested by means of Surface Elec - trode EMG during maximum voluntary contraction in isometric conditions. In order to avoid muscle fatigue, only 5s of the EMG signal were considered for analysis.

## DATA ANALYSIS

The analysis was carried out on a HP2100s minicomputer. After filtering at 800Hz, the measured EMG compound signal was sampled at 5000Hz, A/D converted and memorized on a digital magnetic tape in blocks of 1024 samples. A normal MUAP was also sampled at 5000Hz and memorized as a binary file on the disc user area.

The Fourier Transform (FFT) of the discharge density function was calculated following eq.(4) Inverse Fourier Transform was performed on it, yielding the time distribution of the active MU's, $n(t)$.

## RESULTS AND DISCUSSION

The time distribution of the number of active MU's presents a maximum between 30ms and 50ms, for all subjects. This is in agreement with physiological findings which indicate frequencies in the range from 20Hz to 30Hz as the most utilized by the contraction mechanisms at high contraction levels.

The shape of the time distribution appears to be characteristic of the subject and strongly dependent on the number of points adopted for the calculation of FFT. Implications of this fact as for statistical significance of the EMG signal time epoch considered, is at present not yet fully understood.

The subject-dependent aspect of the distribution might be related to the existence of individual contraction patterns. In order to calculate the "average" distribution in healthy subjects, it is suggested that the whole procedure be repeated with data obtained by means of over-threshold stimulation.

## CONCLUSIONS

The proposed model allows calculation of the time distribution of active MU's from compound EMG signals. The distributions calculated in the case of ten normal subjects, show a maximum at values which are in good agreement with physiological findings.

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## ACKNOWLEDGEMENT

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NOTES

# MOTOR UNIT PROPERTIES II

## Session 10A

## FIRING RATE MODULATION OF CONCURRENTLY ACTIVE MOTOR UNITS

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### INTRODUCTION

In order to better understand the voluntary control of human skeletal muscles, a comprehensive system has been developed for the signal acquisition and detailed analysis of the firing patterns of several motor units recorded simultaneously. This system incorporates new and improved methodology in all of its phases, including: a) micro-computer control of the experiments, b) electrode and multiple-channel signal recording, c) digital computer storage of recorded data, d) pre-decomposition signal conditioning (filtering), e) decomposition of the recorded superimposed action potential trains, and f) statistical analysis display of these firing pattern parameters with the force output.

### EXPERIMENT

The deltoid, biceps, and first dorsal interosseus muscles of the right upper limbs of 5 normal, right-handed males were studied. Their ages ranged from 18 to 35. For each of the muscles, the limb was positioned to minimize the non-linearity in the relationship between the force measured by an appropriately positioned force transducer and the contractile force produced by the muscle. The force was displayed to the subject as a horizontal line on an oscilloscope, along with a target force line which the subject attempted to match. The position of this target line was controlled by a small micro-computer system. The subjects were asked to track increasing and decreasing force as well as constant force isometric contractions, ranging from zero to near maximal levels. The force varying contractions were performed at three force rates: 10%/sec., 20%/sec., and 40%/sec.

After attempting several electrode designs, it was found that a modified version of a commercially available DISA (13k80) bipolar needle electrode was suitable for the experiments. Three channels of EMG signals could be recorded from the needle: one differential between the two wires in the needle, and two differential between the cannula and each of the wires. Although this provided one redundant channel of information, the signal to noise ratio was slightly improved. Because the needle electrode could be repositioned after the initial insertion, it was possible to obtain good quality data for the entire series of isometric contractions with usually only one insertion per muscle.

The EMG signal was amplified by DISA differential amplifiers with the bandwidth set at 1 kHz to 10 kHz, recorded along with the force output on an FM tape recorder, and later transferred to digital storage. The data was greatly compressed by saving only the small portions of EMG signal with action potentials above the background noise level. Both the duration of the recorded action potentials and the background noise were then reduced through the use of a digital filtering program.

### DATA ANALYSIS

The compressed and filtered EMG signal was decomposed into its consistent motor unit action potential trains using an interactive computer program. The program automatically assigns each action potential to a particular motor unit based upon a probability-weighted, template-matching algorithm. If the "match" fails to meet preset error limits, the operator is consulted for assistance. The operator is able to both make and change assignments based upon a multichannel visual analysis coupled with the statistical information provided by the computer Thus, the overall decomposition accuracy and the number of motor units which may be isolated is considerably greater than any automatic or visual analysis alone. The speed of decomposition greatly surpasses that of visual techniques alone.

### RESULTS

Through the use of this system it has been possible to record sufficiently stable EMG signals during constant-force and force varying isometric contractions ranging to near-maximal voluntary effort. Through the use of the decomposition scheme it has been possible to accurately identify and separate (classify) all the action potentials of typically 4, and up to 8, simultaneously recorded motor unit action potential trains detected by one electrode. Figure I shows the firing times and interpulse intervals of 6 action potential trains. Each dot represents a single firing, its vertical position indicating the time elapsed since the previous firing (interpulse interval). Occasionally, two or three of these action potential trains actually arise from different fibers of the same motor unit. This is illustrated with the bottom two action potential trains in Figure 1. The activity of 5 distinct motor units can therefore be observed. Usually 3 to 6 distinct motor unit action potential trains can be obtained.

Statistical analysis of the firing patterns has yielded information concerning the recruitment, "decruitment" and discharge properties of motor units from functionally different muscles. These properties are in the process of being categorized. The analysis performed to date indicates that concurrently-active motor units tend to rapidly modulate their firing rates in unison to compensate for small force fluctuations. The percentage change in firing rate due to this modulation is often two or more times greater than the percentage change of these rapid force fluctuations. The percentage change in firing rate to a slower overall change in force appears to be much less. This can be seen in Figure 2. The time varying firing rate of 4 motor units are displayed. Although masked by the averaging, the slowly firing motor unit fires irregularly until it is "fully" recruited at 12 seconds when the force is increased above 30%. The other three units respond to this increase in force with an initial increase in firing rate, but then return to their approximate previous rate. This is a phenomenon which is seen consistently throughout the data that have been analyzed. It has also been found that some motor units fire in exact synchrony for periods greater than one second; yet, except for modulating their firing rates in a similar manner, they fire relatively independently prior to and after these periods.

## DISCUSSION

From these results emerges a distinction between the roles of recruitment and rate coding in the generation of isometric muscle tension. Motor units appear to be recruited to the degree necessary to sustain an overall tension level. Fluctuations in tension about this level appear to be made by rapidly altering the firing rates of all currently active motor units. This permits very rapid and precise alterations in the tension. The overall firing rate of motor units may also increase to some extent as the overall tension increases. This physiological control scheme has great appeal from an engineering viewpoint.

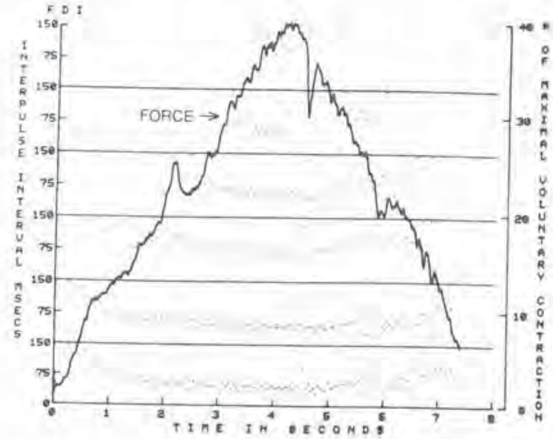


Fig. 1: Firing times and interpulse intervals of six action potential trains as detected by one electrode, during a force-varying isometric contraction.

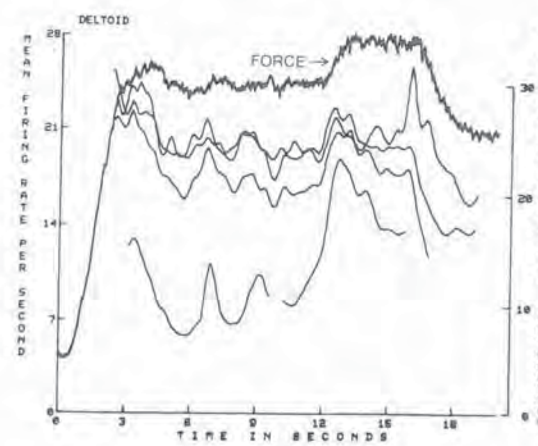


Fig. 2: Time-varying firing rates of four concurrently active motor units and the corresponding isometric force produced.

## ACKNOWLEDGEMENT


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# PATTERNS OF MU ACTIVITY IN THE COURSE OF A MOVEMENT

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## INTRODUCTION

Since Adrian and Bronk's pioneering work (1929), obvious technical difficulties have limited the study of Motor Unit (MU) activity to that of steady isometric or slowly increasing isometric contractions. During the past ten years, using an original wire electrodes technique (Maton et al, 1969), we have studied and quantified the MU recruitment and discharge frequency in various types of voluntary contractions, including anisometric ones.

The purpose of the present study is to compare the results obtained in these different kinds of contractions.

## METHODS

The subjects were seated. The elbow axis was conincident with the rotation axis of a mechanical device, and the hand was semi-prone. The subjects carried out elbow flexions in the horizontal plane. The experimental situations were the following: a) isotonic isometric contractions, ie. different levels of force ranging from 1 to 21 kg. (Maton, 1977) maintained for 30 sec.; b) anisotonic isometric contractions, ie., ramp contractions of different rates, from rest to .5 to 14 kg (Maton, 1977); c) anisometric contractions, ie., movements performed at different velocities against various pure inertia ranging from .179 $m^2$ kg to .504 $m^2$ kg (Maton and Bouisset, 1975), the amplitude of movement being limited to 30° around the rest position of the elbow, and the brake of the movement being achieved either voluntarily or by striking a target; d) anisometric contractions, ie., movements performed against various loads ranging from 1 to 5 kg (Maton, to be published).

MU activity of the biceps brachii was recorded by bipolar wire electrodes inserted into the belly of the muscle with a hypodermic needle (Maton et al, 1969).

## RESULTS

The patterns of MU activity in the course of a contraction were consistent for a given type of contraction: a) in isotonic isometric contractions the MU discharged at a steady rate the value of which depended on the level of force; b) in anisotonic isometric contractions the MU's instantaneous discharge frequency increased with force when the velocity was low and was at first higher and then decreased for higher velocities (Fig. 1); c) in movements performed against pure inertia, the first interval between two consecutive MU discharges was shorter than the following ones and its value depended on the peak velocity and more generally on mechanical work whatever the type of braking (Fig. 2); d) in movements performed against load the pattern was the same as in (c) for the MU's which were silent before the onset of the movement.

## DISCUSSION

Two opposite patterns of MU activity were found: i) a pure isometric pattern which consists of a steady rate of MU discharge; ii) a pure kinetic pattern which consists of a first interval of MU discharge shorter than the following ones. In three out of the four types of contractions , the MU's exhibit one or the other pattern, but in anisometric isometric contractions the pattern depends on the velocity of the contraction. Thus, it seems that the MU activity patterns are differentiated by this criterion.

According to Desmedt and Godeaux (1977) it is possible that contractile potentiation may contribute to the force increase in the case of pure kinetic patterns.

The pure kinetic pattern corresponds to a preprogramming of movement as shown by Maton and Bouisset (1975). This based on the fact that i) the biceps brachii is the first muscle to be active in elbow flexion, and ii) there exists a consistent relationship between the peak velocity and the interval between the first two consecutive discharges of a MU at the beginning of the myoelectric burst which precedes the movement.

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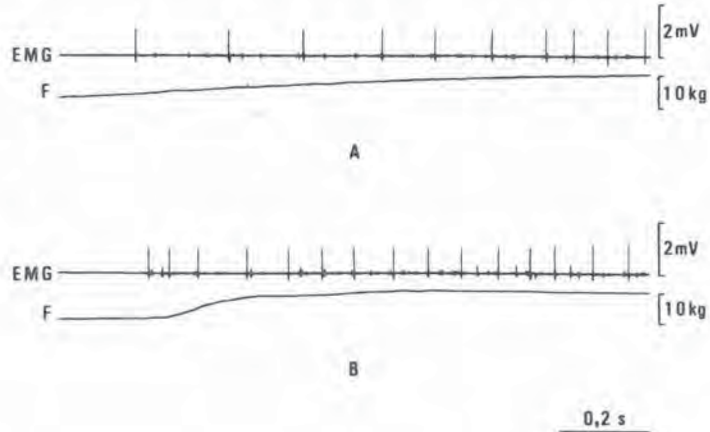


Fig.1:Activity of a single MU during anisotonic isometric contraction performed at low rate(A),and at higher rate(B).
EMG:Biceps brachii MU activity.
F:External force measured at the wrist.

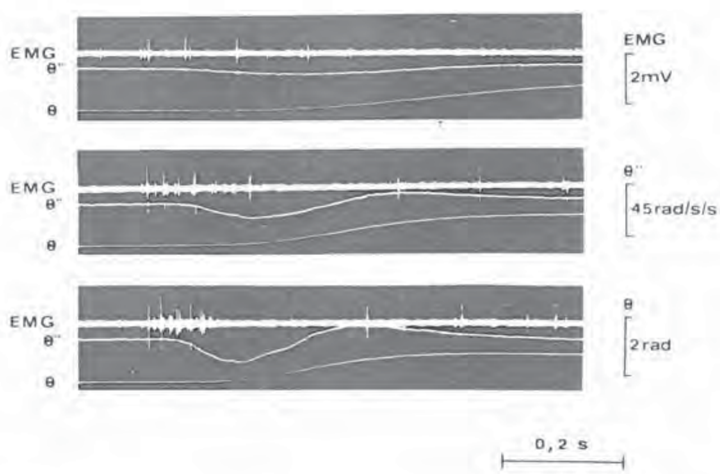


Fig.2:Activity of a single MU during movements performed against inertia.
EMG:Biceps brachii MU activity.
θ'':Tangential acceleration.
θ:Angular displacement.
The three sets of records correspond from top to bottom,to increasing velocities.

DYNAMIC AND STATIC COMPONENTS OF MOTONEURONE ACTIVATION IN MAN

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The silent period (SP) in the electromyogram (EMG) of a voluntarily activated muscle following its response to a tendon tap (Hoffmann, 1920) and its correlation with the firing behaviour of a single motor unit (SMU) under various conditions of activation will be discussed in this paper.

In man, when muscle tension is to be initiated, a SMU begins firing in a manner that depends on the required gradient of effort increase (Gurfinkel et al., 1970, 1972). Once this gradient, dF/dt (Fig.1a), is to be high enough, SMUs come into action in a typical pattern (Fig.1b) described by impulse sequ-

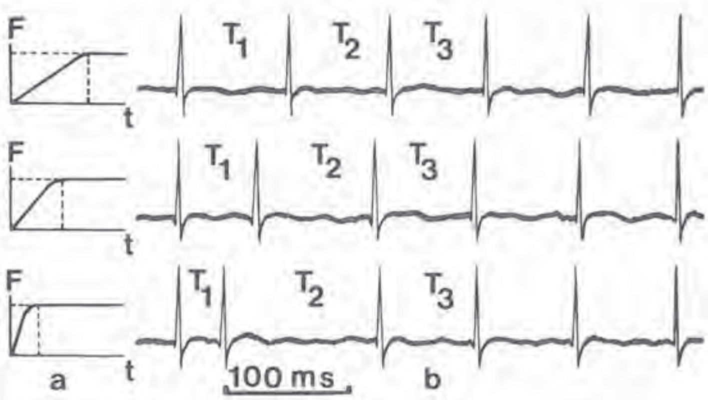


Fig.1 Initiation of muscle tension with differing gradient of effort

ence $T_1 \ll T_2 > T_3$. The sum of the first two inter-spike intervals is close or equal to the doubled value of a mean interval, $\overline{T}$, in the following steady firing; $\overline{T}$ and $T_3$ are equal with precision of 10 ms.

In cat, the same initiation pattern of repetitive firing of spinal motoneurones (MNs) has been reported (Gustafsson, 1974); injection of a short current pulse through the recording microelectrode resulted in an impulse sequence practically identical to the above, provided the current step was high enough. This was considered as evidence that the initiation pattern described is natural for MNs and that the neurone afterhyperpolarization alone might be quantitatively capable of producing this pattern.

It is known that the recurrent influences (Traub, 1977) may result in lowering the MN firing rate after an increase, hence causing longer intervals including possibly the SP as such. The common view which considers unloading of muscle spindles after a reflex discharge of MNs (Matthews, 1972) as the main factor responsible for the SP, seems quite reasonable. However, a well known variety of factors influencing the SP after a reflex response to electrical shock, stretch or tendon tap, is a reason to question its easy interpretation. Such decisive factors as the level of background activity, firing pattern of SMUs and the parameters of stimulation(Struppler, 1975), as well as its purely cutaneous components (Shahani & Young, 1973) must also be taken into account.

Thus, the relationship between the SP and the temporal pattern of the single MU firing behaviour seemed worth studying; a report has appeared elsewhere (Mirsky, 1975). A strong correlation has been shown between the mode of recruitment of a particular MN and the duration of its electrical silence after the monosynaptically elicited response of MNs.

Normal subjects (12 volunteers, age range 17-43 years) were asked to activate a SMU and to try to maintain its steady-state activity using acoustic and/or visual feedback; after several trials, subjects usually perform this task without being much disturbed by the procedure of eliciting the reflex. EMGs of SMUs from the soleus, gastrocnemius, tibialis anterior and rectus femoris muscles were recorded using intramuscular fine-wire electrodes, 62-100 μ in diameter, and conventional amplifiers, scopes and re-

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corders. Ankle or knee jerks were produced by light tendon taps; in control measurements, electrical stimuli were applied to the tibial nerve in the popliteal fossa.

The SP was found practically equal in duration to the sum $T_1 + T_2$, or to the doubled mean interval $\overline{T}$ before, or often to the sum of the first two intervals after the SP, $T_1' + T_2'$ (Fig.2), provided the steady firing was efficiently maintained, i.e. $\overline{T} = \overline{T}'$. The maximal devia-

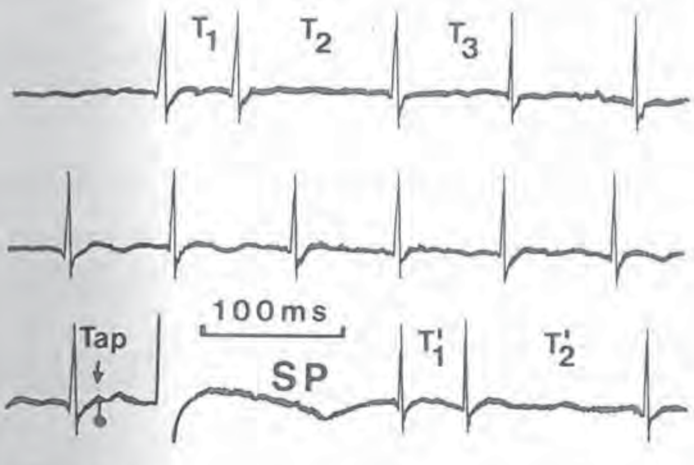


Fig.2 Effect of ankle jerk on the steady firing of a MU (soleus muscle)

tion of the ratio $SP/\overline{T}$ from 2.0 was 1.5, the latter being typical when a subject had difficulties in keeping MU firing stable. When a steady firing of a MU from tibialis anterior is influenced by changes in the MN-pools of triceps surae caused by electrical or mechanical stimuli, the first two intervals after a stimulus, $T_1^{\bullet}$ and $T_2^{\bullet}$ (Fig.3), show the

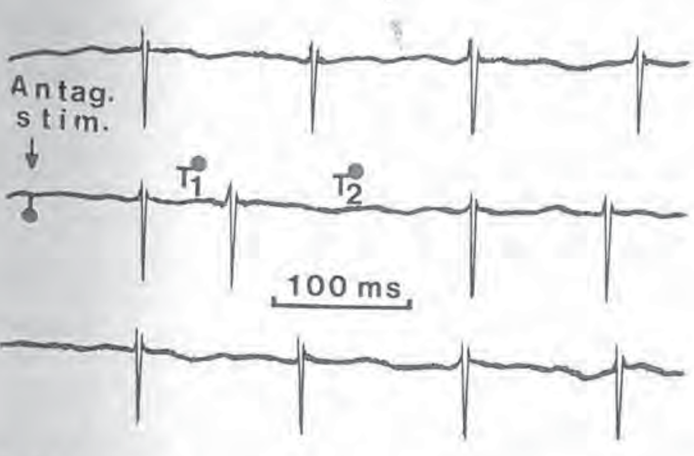


Fig.3 Modulation of steady firing (tib. ant.MU) by antagonist stimulation

same trend of being equal in their common duration to both $2\overline{T}$ and $2\overline{T}'$. All that the weak antagonist stimulation has done is to modulate firing of tibialis MUs, for none of the tibialis records showed synchronous response.

Considered in the aggregate, the described MU firing behaviour during initiation of tension, after a synchronous response to a tendon tap, and when modulated by antagonist stimulation, suggests that the SP after a synchronous MN discharge can be formed by long $T_2'$- or $T_2^{\bullet}$-intervals of single MNs, and the well known "rebound" after the SP - by short $T_1'$-intervals. A hypothesis will be discussed assuming that during initiation, the MN-pool is being activated by a shift in the excitability of the segment (an amplitude, A, of this shift, is reached in time, t), and the future $\overline{T}$ is determined by A, whereas in the slope of this shift, dA/dt, the needed $T_1$ is coded. Such a view is in keeping with previous results and theory (Tsetlin, 1973).

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# SINGLE-MOTOR-UNIT ELECTROMYOGRAPHIC ANALYSIS OF FUNDAMENTAL FREQUENCY CONTROL DURING PHONATION

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## INTRODUCTION

Fundamental frequency ($F_o$) of phonation is controlled by the five intrinsic laryngeal muscles, several extrinsic laryngeal muscles, transglottal pressure, and several other variables. During voluntary control of $F_o$, several of these control parameters are intercorrelated, so cause-and-effect relationships between $F_o$ and the activities of individual muscles have been difficult to establish. At the level of single-motor-unit firings, however, individual muscles can be assumed to be statistically independent. Therefore, if an effect on $F_o$ related to single-motor-unit activity can be found, it can be used to uncorrelate the muscles from each other.

A talker attempting to sustain a steady tone normally produces random perturbations in $F_o$ that are typically 3-4% of the nominal value. It has been unclear to what extent these perturbations are due to the controlling musculature rather than to other random effects. This study was undertaken to determine whether $F_o$ perturbations are related to single-motor-unit firings, so that a standard averaging technique can be used to investigate the causal relationship between contractions of individual muscles and their effects on $F_o$.

## METHODS

A subject was asked to sustain a steady tone for several breaths. Electromyographic (EMG) activity, obtained through hooked-wire electrodes from a laryngeal muscle under study and the simultaneous voice signal obtained through a standard microphone were recorded and input to a digital computer. A measure of instantaneous fundamental frequency as a function of time was derived from the voice waveform. The resulting $F_o$ waveform was offset by approximately its average value and amplified to exaggerate the perturbations. Isolated single-motor-unit firings were identified in the EMG waveform. Then, samples of the EMG waveform and the $F_o$ perturbation waveform were aligned around the single firings and averaged. The sample window extended from 100ms before to 300 ms after these firings.

## RESULTS

Figure 1 shows the results of one such calculation. The muscle under investigation here is the cricothyroid, whose function as a vocal fold tenser, and hence as $F_o$ raiser, is well known. The subject produced a fundamental frequency of about 100 Hz, which is in the lower part of his range in order to keep the number of recruited units and their firing rates low. Nineteen firings, each isolated by an interval great enough to insure against overlap of their mechanical effects, were isolated.

The upper panel in Fig. 1 shows the averaged EMG signal, which exhibits a pulse only at the lineup point, as expected. The lower panel shows the average $F_o$ perturbation. This signal is approximately at baseline both to the left of the lineup point and at the far right of the window. However, there is a positive pulse beginning immediately after the lineup point. The pulse reaches its peak amplitude of 1 Hz at a latency of about 70-80 ms. This pulse appears to represent the average effect on $F_o$ of a firing of a single motor unit in the cricothyroid muscle.

A similar calculation was performed on a strap muscle, an extrinsic muscle of the larynx whose possible function in lowering $F_o$ has been a source of some controversy. The results, not illustrated here, indicate that a systematic $F_o$-lowering effect exists for this subject when the fundamental frequency is already low, but not when the subject produces an $F_o$ in the middle of his range.

## CONCLUSION

The results show that randomness in the controlling musculature contributes to $F_o$ perturbations and that a single-motor-unit averaging technique can be used to study laryngeal control of $F_o$. This result is itself interesting, since $F_o$ perturbations have been used both as an indicator of psychological stress and as an indicator of vocal pathology. The magnitude of the output $F_o$-perturbation pulse in relation to overall perturbation indicates that very few motor units were firing at rates low enough to show the effects of individual twitches. The dynamics of this pulse represent the combined effects of a motor-unit contraction, of laryngeal mechanics, and of phonatory mechanics. Since contraction times can be estimated from animal experiments, these data can be used to test the dynamic performance of phonatory models. Finally, these experiments should be extended to other muscles and to other measures of phonatory performance to further elucidate the nature of the phonatory mechanism and its control.


ACKNOWLEDGMENT

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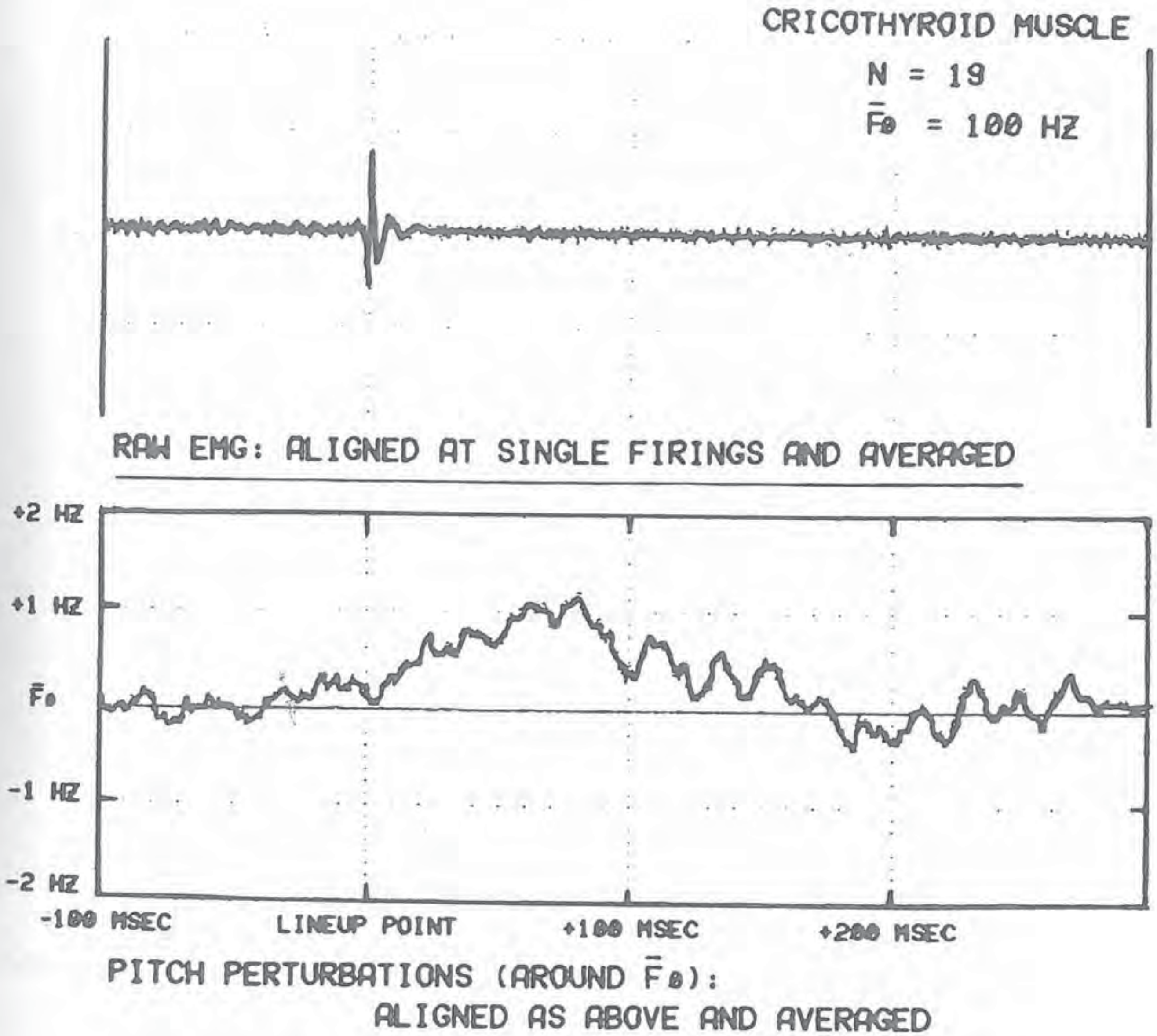


Figure 1

# NOTES

# BIOFEEDBACK

## Session 10B

THE ROLE OF SENSORY FEEDBACK OF INTEGRATED EMG IN THE ABSENCE OF PROPRIOCEPTION

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## INTRODUCTION

Recent controlled investigations (Bird & Cataldo, 1978; Gianutsos, et al., 1979) have confirmed the previously reported clinical studies on the usefulness of sensory feedback of integrated electromyographic activity (IEMG) in the treatment of hemiparesis (Brudny, et al. 1976, 1979) and dystonia (Korein & Brudny, 1976). Considering the different pathophysiology and variability of anatomical sites involved, a common mechanism may be operational in mediating utilization of feedback of IEMG during the restoration of motor control, which may perhaps explain its efficacy in a variety of neuromotor disorders.

In our opinion, visual feedback of IEMG is conceived as a modal substitute for inadequate or disturbed processing on cerebral levels of kinesthetic information that occurs as the result of various etiologies. Such substitute feedback may conceivably undergo intersensory translation to ultimately effect surviving brain areas concerned with motor control and thereby influence the restoration of functional movement. The case of a patient with loss of proprioception due to brachial plexus trauma serves as an illustrative example.

## CASE REPORT

A 35-year-old male patient injured in a car accident has suffered traumatic amputation of right forearm and avulsion of the left brachial plexus. Repeated EMG studies indicated total denervation with the exception of the suprascapular and dorsal-scapular nerves. Exploration of the left brachial plexus was planned one year later. In the event that reconstruction proved unfeasible, an alternative approach was proposed. Namely: the reinnervation of the left biceps muscle by surgically transposing a viable nerve. The procedure would be supplemented by IEMG feedback training before and after surgery. If successful, biceps contraction could provide the limb with an assistive capacity. The right ulnar nerve, lacking functional significance due to forearm amputation, was selected for possible transfer. Preoperative training was carried out utilizing surface electrodes for transducing the EMG activity in the small stump of preserved flexor carpi ulnaris muscle. Within four weeks the patient learned and retained volitional and skillfully graded contractions of the muscle with no further need for feedback. The training paradigm consisted of making feedback contingent on desirable EMG response as described elsewhere (Brudny, et al., 1976). The muscle contraction was initiated by instructing the patient to employ visual imagery to effect the spreading of his absent right fingers.

The surgical exploration revealed that reconstruction was impossible due to extreme separation of nerve remnants and extensive scaring. Sheathed in a silastic tube, the right ulnar nerve was then transposed subcutaneously across the chest with an end-to-end anastomosis to remnants of the degenerated left musculocutaneous nerve. Eight weeks later following several feedback sessions during which the patient was instructed to employ the same imagery, volitionally generated EMG potentials appeared over the left biceps (Fig. 1).

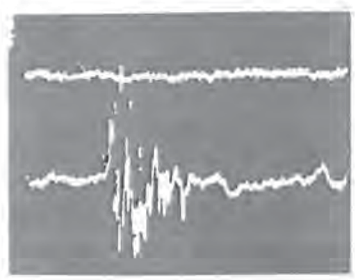

Fig. 1. Reinnervation of left biceps muscle and evidence of voluntary motor activity. On rest fibrillation and positive wave presence (upper tracing), on volition polyphasic motor units of 25 milliseconds duration and 50 microvolt amplitude (lower tracing). Calibration 10 milliseconds, 10 microvolts.

With training, these potentials gradually increased in magnitude and the patient reported subsequently that he was able to think in terms of flexion of the left elbow to effect motor activity mediated by the right ulnar nerve (Fig. 2).

Fig. 2. Patient attempting flexion of the left elbow while observing the EMG feedback displays. Note appearance of silastic tubing subcutaneously over right clavicle and neck.

Three months later there was EMG evidence of normal motor units induced volitionally without benefit of feedback and trace contraction of the left biceps muscle was noted (Fig. 3).

Fig. 3. Patient alternating volitional motor activity (upper and lower tracings) with rest (middle tracing) without benefit of feedback. Amplitude of motor units 400-600 microvolts, duration 14-16 milliseconds, frequency 10-12 per second. Calibration 10 milliseconds, 100 microvolts.

## DISCUSSION

Despite the fact that functional restoration was not documented in the above case due to patient's loss to follow up, the theoretical and practical implications of the case are of considerable interest.

Viewing motor control in cybernetic terms, a closed loop sensorimotor interaction is facilitatory to the acquisition of motor skill. In the described case vision provided a means for linkage between sensory input and motor output. Visual imagery provided a mechanism for initiating efferent neuronal signal flow and then visual feedback provided means of augmenting what remained of the proprioceptive loop prior to surgery as well as reestablishing a substitute closed sensorimotor loop during the postoperative training. When feedback was phased out, the patient successfully performed in an open loop fashion. This assumption is supported by the findings during the initial surgical exploration which revealed that spontaneous reinnervation is beyond any reasonable probability.

The above considerations imply that the brain possesses adequate plasticity to utilize substitute modes of feedback in effecting central motor control. Parallels can be drawn between the present case and what is encountered following brain insult (Brudny, et al., 1979). In the latter case, where it is evident that much brain tissue remains intact, the possibility that this reserve may be utilized for functional restoration with aid of substitute sensory feedback looms as an enticing topic for future endeavors.

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IMPLICATIONS OF VARIABILITY IN SPASTICITY OBSERVED DURING EMG FEEDBACK TRAINING

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## INTRODUCTION

EMG activity in spastic muscles may vary greatly even when stimulus conditions are kept fairly constant. During a study in which subjects used EMG feedback to attempt relaxation of spastic flexors during passive stretch, fluctuations in spasticity were measured continuously in baseline and training conditions, affording an opportunity to relate variations in spasticity to other aspects of subject behavior. An understanding of the basis of this variability is needed if the effects of feedback training are to be properly assessed.

## EXPERIMENT

Subjects were 5 spastic hemiparetics between 20 and 60 years old, victims of CVA or brain trauma. Each subject was seated comfortably and the spastic arm was subjected to repeated extension/ flexion cycles at controlled angular velocities by means of a Cybex Isokinetic Dynamometer. EMG activity from four different muscle placements was monitored with surface electrodes, integrated, and stored as digital values on computer disk memory every 1/10 second. During training, the computer also generated EMG feedback displays of integrated muscle activity on a screen facing the subject. In baseline and training sessions, each subject tried to keep spastic flexors relaxed while the wrist or elbow was flexed and extended at a velocity chosen to elicit light to moderate spastic activity. Performance was assessed concurrently with a 4 channel scope displaying the EMG voltages and a DC pen recording of the integrated muscle activity levels quantified by the computer. Retrospective analysis included calculation of average EMG activity during static and dynamic flexion and extension, and computer plotting of such averages.

## RESULTS

Baseline testing revealed typical velocity dependent spasticity in the elbow and/or wrist flexors of all subjects. All showed slight spasticity even at speeds of 12°/sec. and below. During passive extension, EMG activity in spastic flexors built up rapidly. It showed an approximately exponential decrease when the limb was held in static extension but usually persisted, especially if the stretch rate had been high. Passive flexion would abruptly decrease any EMG persisting from the previous extension. At low flexion speeds, small amounts of EMG activity sometimes continued in spastic flexors, while at higher speeds these muscles usually relaxed entirely. Probably the more rapid stretch of muscle spindles in extensors had a stronger reciprocal inhibitory effect on spastic flexors.

In static flexion, complete relaxation usually ensued, but in all subjects the persistent repetitive firing of large action potentials of regular waveform was noted at times. Such potentials have been described previously by Regenos and Wolf (1). It is not yet known what proportion of these are true single motor units; some can be shown to consist of smaller units firing in closely locked synchrony. Subjects had difficulty suppressing such isolated units by voluntary effort. Some learned that moving the head or the normal arm might terminate firing. Initiating passive stretch sometimes silenced these potentials.

One puzzling observation was that biceps spasticity increased substantially when its initial length was reduced by hand supination. In general, spasticity can be expected to decrease if the dynamic stretch cycle is applied to a shortened muscle. Biceps' discrepant behavior might be due in part to a postural factor, since it has an increased role in elbow flexion with the hand supine, and may be facilitated when so positioned passively. Flexing the wrist during supination would decrease biceps spasticity somewhat. This suggests that hand supination increases biceps spasticity by imposing additional mechanical stretch on wrist flexors in the forearm group which cross the elbow. This would increase their stretch reflex response during elbow extension, which could elicit additional spasticity in biceps in spite of its foreshortening.

During initial baseline, these five subjects and others examined more briefly showed less EMG activity in spastic muscles when flexed and extended in the Cybex than when manipulated similarly by hand. Clinicians relying on manual examination to measure spasticity may overestimate its role in interfering with voluntary movement. (See also Sahrmann and Norton, 1977.) Spasticity levels typically decreased over the 3 to 6 baseline sessions preceding feedback training as each subject became more accustomed to the experimental setting. Often levels would decrease during the first 10 minutes of each training session, even when feedback was unavailable, probably due to a general settling down of the subject's reflex reactivity during quiet sitting.

Spasticity varied considerably both within and across sessions, even when stretch velocity and other conditions were constant. (See Fig. 1.) Patients might attribute a generally bad session to tension, lack of sleep, etc. Sudden increases in spasticity within sessions might follow coughing, sneezing, or large movements, analagous to the effect of the Jendrassik maneuver in increasing stretch reflexes. Sudden decreases in spasticity were also seen, usually following movements of the head or of the contralateral limb. Such movements might cause persistent large motor units or groups of smaller units to cease firing.

Each subject received at least 10 training sessions with EMG feedback. Extension/flexion speed was kept low during training and was increased only if spastic flexors were quite relaxed. Three subjects did not show clear and consistent performance gains, and spasticity remained quite variable. They did learn to avoid movements which increased spasticity, and two discovered that brief movements of the head or the unaffected arm would sometimes decrease spasticity, but there was no evidence of direct and consistent inhibitory control. Performance with and without feedback differed little. The other two subjects made substantial performance gains, often attaining zero spasticity in elbow flexors even during fairly rapid (90°/sec) elbow extension. Their strategies differed markedly. One subject made use of reciprocal inhibition: during extension he would push or think about pushing with triceps, and this would reduce flexor spasticity. He followed the feedback display closely and could subdue spasticity promptly at the start of each session. By contrast, the remaining subject always required at least the first quarter of a daily training session to achieve zero spasticity. Her strategy was to think about the arm becoming relaxed while avoiding doing anything to stir up spasticity. Comparison of trials with and without feedback were inconclusive, but she never relaxed quite so thoroughly in no-feedback sessions as in her best sessions with feedback. With learned control in these two subjects came a marked reduction in the variability of spasticity.

Generalization of learned relaxation from trained to untrained muscles was inconsistent and incomplete. The muscle used for feedback usually relaxed more thoroughly and consistently than its synergists. Figure 1 shows substantial variation in brachioradialis during a session in which feedback was derived from biceps. Biceps activity was lower and less variable during the same session.

No objective measures of relaxation outside the training situation are available, but the most successful subject thought the training helped him to keep his arm straighter during gait. Three other subjects reported instances of improved active elbow extension or of the appearance of extension in some fingers which might possibly be related to reduced spasticity resulting from the training.

## DISCUSSION AND CONCLUSIONS

The results suggest that therapists using EMG feedback to train relaxation during passive stretch could mistakenly attribute spasticity decreases within and across sessions to the effects of learning when the real cause may be only adaptation to the training situation. Minor changes in limb positioning and handling can also cause spasticity decreases easily misconstrued as learned inhibition. Since small head and limb movements can cause temporary reductions in spasticity, patients may learn to make such movements as an indirect and inefficient means of manipulating feedback. The variability of spasticity makes assessment of change more difficult, but to the extent that a patient succeeds in learning to inhibit spasticity, variability will also be reduced, so that variability can be used as an important secondary index of learning progress.

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## ACKNOWLEDGEMENTS


The authors are grateful for the programming assistance of Mr. Harold Clifford and Mr. James Malone. The study is supported by NINCDS grant NS 12280 and RSA grant 16/P-56808/4-15.


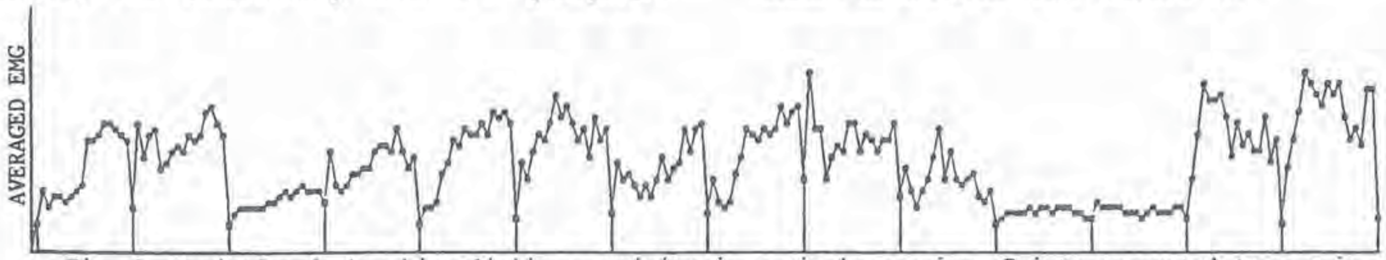


Fig. 1. Variation in brachioradialis spasticity in a single session. Points connected to x axis are EMG averages for 40 sec. rest periods, during which muscle was almost completely relaxed. Remaining points represent spasticity averaged during each 50° elbow extension at 12°/sec.

# EMG BIOFEEDBACK INSTRUMENTATION FOR CLINCAL USE.

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## INTRODUCTION

Instrumentation for the measurement and display of skeletal muscle EMG is widely used both for electrophysiological research and, clinically, for EMG biofeedback. Initially, research oriented equipment was used in both applications (Basmajian, 1974). More recently, equipment has appeared on the market which has been "specifically designed " for clinical use. These "clinical" devices are often purported to be superior to research equipment when used for EMG biofeedback therapy.

Following a request to the authors of this paper for assistance in setting up clinical EMG biofeedback facilities at the Rehabilitation Institute of Montreal, a study was made to determine the best type of equipment for this application. The conclusions reached in this study, and the results obtained using various biofeedback techniques on different categories of patient, are presented in this paper.

## METHODS

" Clinical " EMG biofeedback equipment from six different manufacturers were examined and compared with research oriented equipment consisting of:

- two TECA AA6 MkIII EMG amplifiers
- a 2-channel storage oscilloscope
- a high fidelity audio system
- two digital displays showing integrated full-wave rectified EMG with integration periods of 1 second or 10 seconds

The parameters of interest included:

(1) Sensitivity of the equipment over a wide range of EMG levels.

(2) Degree of immunity from external interference and artifacts.

(3) Suitability of the feedback information to both patient and therapist.

(4) Convenience of operation of the equipment.

(5) The ability of the equipment to function alone without additional monitoring instrumentation.

Thirteen patients participated in this study with diagnoses including:paraplegia, quadriplegia, hemiplegia, brachial plexus injury and traumatic amputation followed by re-attachment of the limb.

## RESULTS AND CONCLUSIONS

A. Choice of a Clinical EMG Biofeedback System

A number of conclusions were reached in this study which bear on the choice of EMG biofeedback equipment:

(1) Measurement of EMG is fraught with many possibilities for error arising from external electrical interference, movement artifact, bad electrode contact, faults (intermittent or continuous) in the measuring instrument, crosstalk from adjacent muscles, etc. This problem exists for both the research oriented and " clinical " instruments and is especially severe when the equipment is used in the electrically noisy hospital environment. The best way to detect (and therefore eliminate) such errors is to provide the therapist with an oscilloscope display of the raw EMG. A minimal amount of training and experience is required to be able to distinguish between EMG signal and noise. None of the " clinical " instruments examined had oscilloscope displays. Thus the research oriented equipment is superior in this regard.

(2) The oscilloscope display is also a useful mode of EMG biofeedback for the patient. He can be quickly taught to recognize even small changes in the raw EMG activity. The display gives instantaneous information as to the state of the muscle and this is a desirable feature in many treatment procedures. On the other hand, the visual displays used in most " clincial " instruments consist of either an analog panel meter or a series of lights. They have very limited range, thus necessitating the use of nonlinear data compression or thresholds. This can be confusing to both patient and therapist by making, for example, a muscle appear relaxed when the activity has merely gone below an often ill-defined threshold. The oscilloscope display does not suffer from this limitation, having excellent

sensitivity over the whole EMG amplitude range.

(3) The audio feedback produced by many " clinical " biofeedback instruments consists of an amplitude or pitch modulated tone, or a frequency modulated sequence of clicks. These sounds were found to be very aggravating to patients and therefore were sometimes counter productive. On the other hand, patients have found that audio feedback of the raw EMG (using a good quality audio amplifier) is very acceptable and is useful in discriminating even small changes in the EMG level.

(4) A digital display of the full-wave rectified and integrated EMG, with a 1 second period, is an excellent form of biofeedback for many applications. It is available on some " clinical " instruments and was also included in the research oriented equipment used in this study.

On the basis of the above considerations, and others, the " clinical " biofeedback instruments were decisively rejected by both research and clinical personnel in favour of the research oriented equipment. The greater portability of some " clinical " equipment was not considered to be a significant advantage, even for gait training, since the small preamplifiers of the research oriented system could be fitted with long extension leads.

B. Clinical Experience with the Chosen System

Surface electrodes were used in all instances. When there was difficulty in locating the optimum electrode postion, a hand-held electrode was used. This electrode incorporated two small saline-soaked pads with adjustable spacing. It was used to quickly scan the whole muscle area.

The training goals for patients fell into four broad categories:

(1) Muscle Relaxation

For relaxation of a single muscle, the digital display with a 1 second period was most often favoured by the patients and produced the best results. Audio feedback was a helpful adjunct. For relaxation of two muscles simultaneously, the 2-channel oscilloscope was used with audio feedback from one of the muscles if desired. These procedures were successfully used for treatment of spasticity in hemiplegia.

(2) Muscle Strengthening

The digital and audio displays were very useful forms of feedback during isometric muscle strengthening exercises. The digital display was used to set clearly defined goals which presented stimulating challenges to the patients. The oscilloscope display was used in training patients to produce graded muscle contractions. Patient response to the oscilloscope display was uniformly favourable. Many of them said that it gave them a better " feel " for what was happening in the muscle than they received from any other form of feedback. In the case of one patient with a brachial plexus injury, a five-fold increase in the mean level of biceps EMG was observed over a fourteen month period. A significant increase in muscle function also occurred.

(3) Muscle Strengthening with Inhibition of Antagonist

A number of patients were successful in using the 2-channel oscilloscope to maximize one EMG trace and minimize the other. The audio feedback was usually connected to the muscle to be strengthened. Progress was monitored by the therapist using two digital displays with 10 second integration periods. This procedure has been successfully used on patients to inhibit agonist - antagonist co-contraction.

(4) Gait Training

When using biofeedback to improve gait, the greatest degree of success was achieved when the initial training was done with the patient stationary (e.g. practising stance or swing from a standing position). Patients with severe gait disability were not able to pay sufficient attention to feedback information while walking. Audio feedback was used predominantly, but visual feedback was occasionally used when the patients were stationary. The improvement in gait was monitored from time to time by recording EMG signals from the appropriate leg and hip muscles during gait, and displaying them on a strip chart recorder along with the phases of gait.

## SUMMARY

In the clinical use of EMG biofeedback, research oriented instrumentation was found to be significantly superior to many of the recently introduced " clinical " devices.

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## ACKNOWLEDGEMENTS

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# DESIGN AND EVALUATION OF A CLINICALLY-ORIENTED MYOFEEDBACK DEVICE

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## INTRODUCTION

Investigators have reported success in the rehabilitation of certain neuromuscularly impaired patients by the use of auditory electromyographic (EMG) feedback, also known as myofeedback (Brudny et al. 1974, and others). Many of these studies were done in a research environment under the supervision of investigators highly trained and motivated in the use of myofeedback. However, the acceptance of myofeedback in a clinical environment has not been widespread. Some of the reasons for the limited acceptance by clinicians have been identified and a device has been designed with these factors in mind. The function of the device is to indicate, to the clinician and patient, when the mean-rectified value of the surface EMG signal surpasses a predetermined threshold level. This indicates when a muscle has contracted beyond a certain force level (contract mode).

## CLINICAL REQUIREMENTS FOR A MYOFEEDBACK DEVICE

The following drawbacks have limited the acceptance of commercially available myofeedback devices in clinical environments:
1) physically cumbersome
2) complicated adjustment of controls
3) time consuming electrode application and removal
4) high cost

With these factors in mind, a device (first generation) was designed and built for use as a therapeutic and evaluating tool. In addition, several other features were implemented:
1) an electrode that does not require conductive paste or gel
2) reliable performance and rugged construction
3) a means of displaying information suitable for documentation in patient's records and to indicate patient's progress
4) capability of home-use by the patient
5) Battery operation with test indicator.

After a clinical evaluation, several modifications were made to alleviate design problems and include the following additional suggestions:
1) separate threshold and amplifier gain controls were combined into one control.
2) a smaller and more reliable pasteless electrode was developed (described below).
3) a volume control was added so the audible tone would not disturb other patients.
4) a visible indicator accompanied the audible tone for hearing-impaired patients.
5) in addition to the contract mode, a relax mode was added whereby the tone (and light) is turned on when the muscle has relaxed below a certain force level.

The pasteless electrode, an important feature of the myofeedback device, uses the stainless steel housing of two junction field-effect transistors as both the recording surface and front-end of a differential amplifier. The transistors are housed in a metal shell which also serves as the ground electrode. The electrode unit is sufficiently small to be used on hand muscles, and requires only 10 seconds to become electrically stable. A Velcro strap attachment permits the electrode to be used in a stationary manner, or it may be used as a prot to rapidly explore many different muscles.

The second generation device is currently being clinically tested. Figure 1 illustrates the device in conjunction with the improved pasteless electrode used as a probe.

## CLINICAL EVALUATION

One Physical Therapist in each of six institutions was assigned a myofeedback device to be used in conjunction with his or her normal case load. Records were kept for an average of 30 days under four general categories of application: 1) initial evaluation, 2) therapeutic exercise, 3) program evaluation and 4) self-exercise. The number of cases for which the device was used with respect to application and muscle is shown in Table 1 for the lower extremity, and in Table II for the upper extremity. At the end of the study, each therapist was asked to evaluate the device with respect to:
1) ease of usage
2) things liked best and least about the device
3) suggestions for improvements and future studies.

## DISCUSSION

Examination of Tables I and II reveals the large number and variety of uses over 30 days, and thus the acceptance of the device in a clinical setting. The request for several modifications made by clinicians, which were incorporated into the second generation device, further indicate the acceptance of the device and the clinician's willingness to improve it.

The two Tables also show that clinicians used the device on a large variety of muscles of the upper and lower extremities and in all four application categories (predominately in therapeutic exercise), demonstrating its versatility. The device was applied in more cases involving the lower limbs than the upper limbs. While this might indicate that the device and technique are more suitable to the lower extremity, it must be pointed out that the majority of patients seen by physical therapy departments have dysfunctions of the lower limbs.

The utilization of the device for self-exercise is important. This indicates that the design of the device is sufficiently simple for unsupervised patient use, requiring a minimal amount of training in its use.

## CONCLUSION

The value of myofeedback has been clearly shown, but its potential for patient care in a clinical or home setting is yet to be realized (Gonella et al., 1978). A main reason for this discontinuity is the lack of myofeedback devices on the market that meet the needs of clinicians in a busy environment. When a myofeedback device was designed and built with the requirements of clinicians as the primary design factor, an evaluation revealed that the device had a high level of acceptance. Thus, while a medical device may function well from an engineering point of view, it is imperative that it meet the needs of the user. Indeed, the development of the device reported herein emphasizes the need for interaction between engineers, medically related personnel, and patients in the design procedure.

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TABLE I
APPLICATIONS IN LOWER EXTREMITY

| Muscle or Muscle Group | Initial Eval. | Therap. Exercise | Program Eval. | Self Exercise | Total |
|---|---|---|---|---|---|
| Quadriceps | 29 | 142 | 82 | 34 | 287 |
| Ant. Tib. | 16 | 64 | 25 | 14 | 109 |
| Hamstrings | 11 | 29 | 22 | 10 | 72 |
| Glut. Med. | 0 | 32 | 22 | 14 | 58 |
| Glut. Max. | 3 | 14 | 8 | 4 | 29 |
| Gastroc-Soleus | 0 | 9 | 12 | 0 | 21 |
| Hip Flexors | 0 | 5 | 1 | 0 | 6 |
| Extensor | 1 | 2 | 2 | 0 | 5 |
| Peroneals | 0 | 2 | 0 | 0 | 2 |
| Adductors | 0 | 1 | 0 | 0 | 1 |
| Total | 50 | 300 | 174 | 66 | 590 |

TABLE II
APPLICATION ON UPPER EXTREMITY

| Muscle or Muscle Group | Therap. Exercise | Program Eval. | Self Exercise | Total |
|---|---|---|---|---|
| Intrinsics | 8 | 7 | 7 | 22 |
| Ext. Carpi Radialis | 8 | 7 | 6 | 21 |
| Biceps | 17 | 3 | 1 | 21 |
| Middle Deltoid | 18 | 0 | 0 | 18 |
| Triceps | 10 | 3 | 1 | 14 |
| Ext. Digit. | 13 | 0 | 1 | 14 |
| Finger Flexors | 3 | 0 | 0 | 3 |
| Pronator Teres | 1 | 1 | 1 | 3 |
| Post. Deltoid | 2 | 0 | 0 | 2 |
| Scapular Muscl. | 2 | 0 | 0 | 2 |
| Total | 82 | 21 | 17 | 120 |

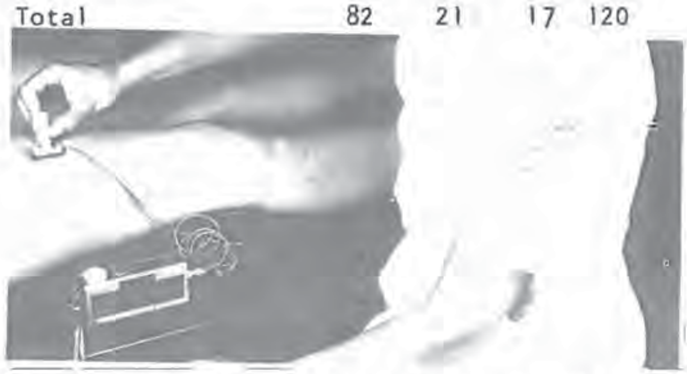

Fig. 1: Myofeedback device and pasteless electrode used as a probe.

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# N O T E S

# LOCOMOTION IV

## Session 11A

# ELECTROMYOGRAPHY OF THE VASTUS MEDIALIS IN NORMAL SUBJECTS DURING GAIT

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## INTRODUCTION

Several investigators believe that a selective function of the vastus medialis is terminal knee extension. This is supported clinically by the observation that vastus medialis atrophy often co-exists with lack of terminal knee extension. Anatomically the vastus medialis has been seen to contain two separate portions differing in their angle of pull on the patella. The upper fibers, running 15-18 degrees medially have been termed the vastus medialis longus (V.M.L.) and the lower fibers running 50-55 degrees medially have been termed the vastus medialis oblique (V.M.O.) There are no reports describing the phasic activity of these two portions of the vastus medialis during gait. The purpose of this study was to quantify the phasic activity of the vasti on normal subjects during gait and to determine if there is a difference in the phasic activity of the two portions of the vastus medialis.

## METHOD

Ten men and seven women volunteered to serve as subjects. Their ages ranged from 21 to 40 years. Fine wire intermuscular electrodes were placed in the vastus intermedius, vastus lateralis, vastus medialis oblique portion and vastus medialis longus portion. The vastus medialis oblique was sampled in the center of the prominent belly four centimeters proximal to the supreme border of the patella and just medial to its medial border. The vastus medialis longus was sampled just lateral to the oblique portion. Insertions were checked by visible contraction to electrode stimulation. Footswitches were placed on the subject's side and the data was telemetered to the recording equipment. Raw E.M.G. signal printed from a Honeywell visicorder was used for analysis. Subjects were asked to walk at their normal pace, footpace, toe-in and toe-out. Two runs of each were recorded for analysis.

## DATA ANALYSIS

Phasic activity analysis was performed manually on the visicorder paper records of raw E.M.G. signals. Three consecutive steps midway in each run were analyzed. All data for each muscle was normalized in percent of gait cycle time based on the interval of time between two consecutive heel strikes. A components of variance test was used to determine how much error was introduced by differences between subjects, differences within a subject on two consecutive runs and due to quantification technique.

## RESULTS

At free gait velocity vastus medialis oblique and vastus medialis longus activity occurred from 87% to 20% and 86% to 21% of the gait cycle respectively. Vastus intermedius and vastus lateralis activity occurred from 86% to 22% and 89% to 20% of the gait cycle respectively. Eight subjects showed secondary activity around toe-off but this was very inconsistent. The activity of the V.M.O. and V.M.L. during fast walking occurred from 85% to 18% of the gait cycle. Results from the toe-in and toe-out gait was inconsistent. The components of variance test showed that with the exception of the V.M.O. the greatest inconsistency occurred within each subject producing an average of 58% of total variation. Differences between subjects accounted for 28% of variation. The analysis technique accounted for 14% of variation.

## DISCUSSION

The summary of the subjects free velocity trials showed that phasic activity of all the vasti group occurred nearly simultaneously. This supports statements that the quadriceps act as a unit to produce knee extension. No differentiation could be made between the phasic activity of the V.M.O. and V.M.L. It can be presumed that both portions of the vastus medialis co-contract with the other heads of the quadriceps to maintain patellar alignment. Inconsistent secondary activity may be explained by variations in plantar flexion control necessitating active quadriceps contraction. Fast walking produced phasic activity occurring 3% sooner. This seems due to the inability of passive knee extension during early swing to meet the timing of heel strike. The components of variance test unexpectedly showed that the greatest percent of variation occurred within subjects. This lends validity to the necessity of averaging many steps when determining phasic activity during gait. In general, the least error was due to quantification technique although it is a slow tedious procedure and objective standards are difficult to establish. Differences produced by different evaluators may be large and for this reason computerized processing of phasic activity is preferrable.


## SUMMARY

No difference in the phasic activity of the vastus medialis oblique portion and vastus medialis longus portion was seen in normal subjects during free velocity gait. Fast walking produced onset of activity 3% sooner. The greatest source of variation was within each subject rather than between subjects or due to E.M.G. quantification technique toe-in and toe-out walking produced highly variable results.


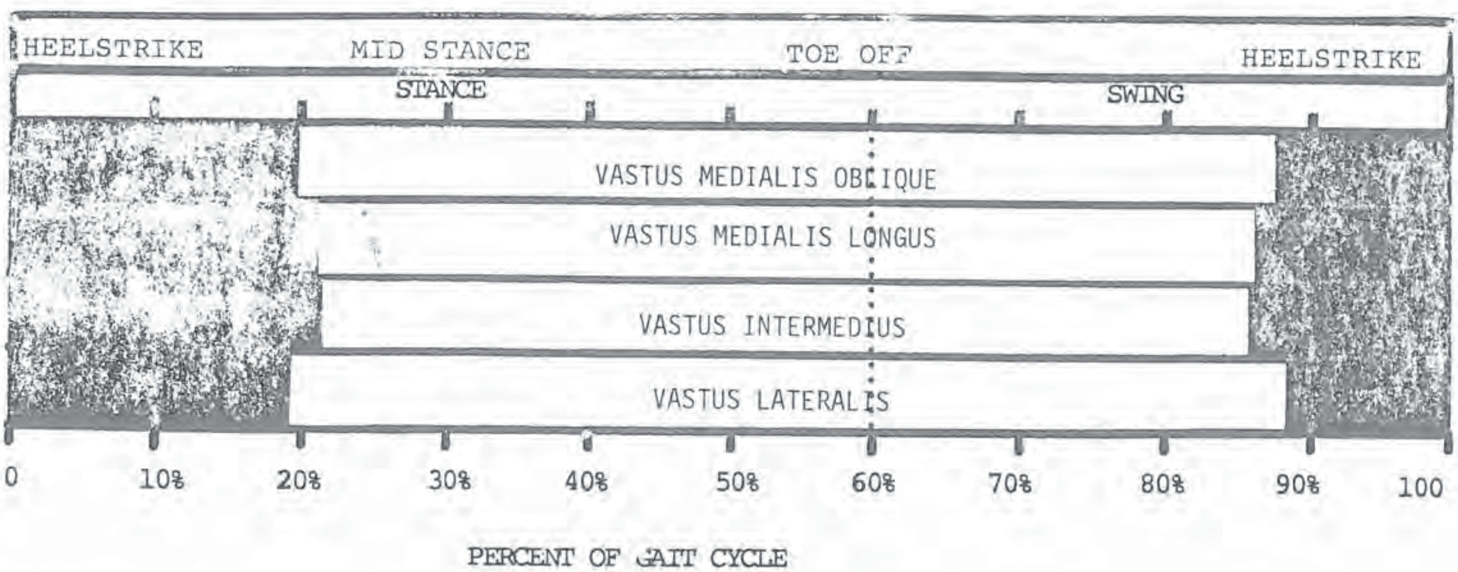


Fig. 1 Mean phasic activity of vasti group in normal subjects during free velocity gait.

# THE USE OF IMPLUSE MOMENTUM FOR QUANTIFICATION OF GAIT DISORDERS WITH THE AID OF FORCE PLATES

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## INTRODUCTION

Dynamic force plates have been in use for many years in gait laboratories and rehabilitation institutions. It seems that lately there has been a tendency of many investigators to abandon this tool since no adequate clinical technique for evaluation of gait has emerged from the use of force plates. When working with force plates in conjuction with prosthetic fitting work, one realizes that many abnormalities are reflected in the ground force traces. However, the correlations between the actual gait deficiencies and the way they are reflected in the force information have not been established. We believe, therefore, based on our own experience, that the force plate is an extremely informative tool. However, investigators have not developed as yet a clear data processing and interpretation method which can be used by the clinical team. The present work describes just one method of processing which seems attractive for clinical use.

## THE CONCEPT OF ANALYSIS

The ground reaction forces are directly responsible for the fluctuation of the center of gravity. Using a cartesian coordinate system, we can relate separately to the vertical, anterio-posterior (A-P) and lateral dynamics of the c.g. The three movements are interrelated through various constraints but of major importance appears to be the A-P component which is responsible for the forward ambulation.

In Figure 1 there is a typical example of A-P ground forces of both legs in normal gait. These, if superimposed, will result in the curve describing the acceleration of the c.g. and through integration, velocity and displacement can be obtained. If we integrate the force by time we will obtain the impluse generated by the leg in the named direction

$$\int F dt = mv_2 - mv_1. \qquad (1)$$

The impulse is represented therefore by the area under the curve. It can also be seen from Equation 1 that the impulse equals the difference in linear momentum between two points.

The A-P force information is composed of two major phases for each leg: the pushing phase and the braking phase. The impulse of each such phase represents the gross ammount of activity on the part of the patient during each such phase (not in terms of energy). This impulse quantity could therefore serve as a good means of evaluation of the gait performance during the various phases. One immediate result which proves the quality of the technique is the gait consistency test (Seliktar, 1978). Since the use of walk path with two incorporated force plates may attract too much attention of the walker and alter his gait, the data has to be examined for consistency. Assuming that the gait is a cyclic phenomena, then the boundary conditions at every two equivalent points of consecutive cycles should be the same. Expanding this concept into its analytic form the total impulse between two equivalent points should be zero for a consistent cycle. Additionally, the actual impulse values of the phases can be used to assess quantitatively the quality of performance.

## EXPERIMENTAL RESULTS AND DISCUSSION

The consistency test was applied to fifty gait tests of normal subject, amputees and hemiplegics. The limit of consistency was determined in a previous study to be 8% offset from zero in terms of percentage of the total measured area.

Out of the fifty test, nine were found to have been bad tests due to offset exceeding 8% (inconsistent). The rest of the tests were found consistent with offset well below 8%. The low percent inconsistency found in the good tests (0-8%) is due to both data acquisition inaccuracies and natural limited inconsistency of consecutive gait cycles. The acquisition errors and repeatability of cycles were separately tested and the offset was found to be less than 8% for each.

The variation of the phase impulse between various pathologies was interesting. Normals displayed a certain degree of assymetry between the two legs. Excessive push by one leg was compensated by reduced push of the other leg. This feature applied also to the various combinations of the four phases. Above knee amputees normally would display a reduced braking impulse on the prosthesis, which indicates compensation for instability. Increased braking activity by the prosthesis may produce a large enough moment to bend and buckle the knee while load bearing. The amputee being aware of this, automatically compensates either by exerting his hip extensors during the braking phase or reduces his stride forward with the prosthesis, practically initiating his stance phase with a leg which is almost in a vertical position. By proper realignment of the prostheses toward stabilization the braking impulse is increased. It is evident from this study that this method can serve a good quantitative measure for prosthetic alignment. This however, necessitates the conduction of the above consistency test to ensure the dealing with correct data.

The whole analysis was carried out with normalized data. The force was divided by the body weight W and the time scale was normalized by dividing it by the time period T of the walk cycle. This compensated to a certain extent for variations between subjects in terms of body built and speed of walking. This created a more homogeneous data and resemblance between the impulse quantities of the same phases in different individuals.

A phenomenon which we termed "impulsiveness of gait" could be observed from the result. The total measured impulse, positive + negative, for one complete cycle for normal individuals given in a nondimentional form as described above ranged between $8 \times 10^{-2}$ and $12 \times 10^{-2}$. Rarely the individual data exceeded these values upwards or downwards. The high values indicated the impulsive gait while the lower values were referred to as less impulsive gait. The individual phases were invovled about one quarter of this value, e.g. $2 \times 10^{-2}$ to $3 \times 10^{-2}$ per phase, positive values for pushing and negative for braking.

For the various pathologies several combination of phase impulses were observed. In general five logical combinations exist when one leg is abnormal:

1. All the phases are equal impulsewise (in absolute values). This is more like the normal gait feature although, as stated above, all normals displayed a certain assymetry.
2. Pushing on the affected leg is higher and braking is higher, e.g. the affected leg is more active than the normal.
3. Pushing on the affected leg is lower and braking is lower, e.g. the normal leg is more active than the affected one.
4. Pushing on the affected leg is higher but braking is lower.
5. Pusing on the affected leg is lower but braking is higher.

The first possibility is very rare but it may be obtained with certain below-knee amputees and with hemiplegics which are almost fully recovered. The second is very unlikely to occur but the third, fourth and fifth do occur in the different pathologies. Normally, the particular combination is correlated with the particlar pathology. For instance, the third possibility is typical to above-knee amputees with unstable knee. In neuro-muscular diseases this technique is more suitable for the conduction of quantitative assessment of performance for follow-up purposes.

## CONCLUSIONS

The present work describes a simple quantitative technique for evaluation of gait. Its accuracy can be evaluated on the spot and it can be used for clinical follow-up and prosthetic alignment. The data presentation is simple and adequate for use by the clinical non-engineering personnel.

## ACKNOWLEGEMENT

The work was carried out at the Biomechanic Laboratory of the Lowewnstein Rehabilitation Hospital and at the Julius Silver Institute for Biomedical Engineering at the Technion Israel Institute of Technology.

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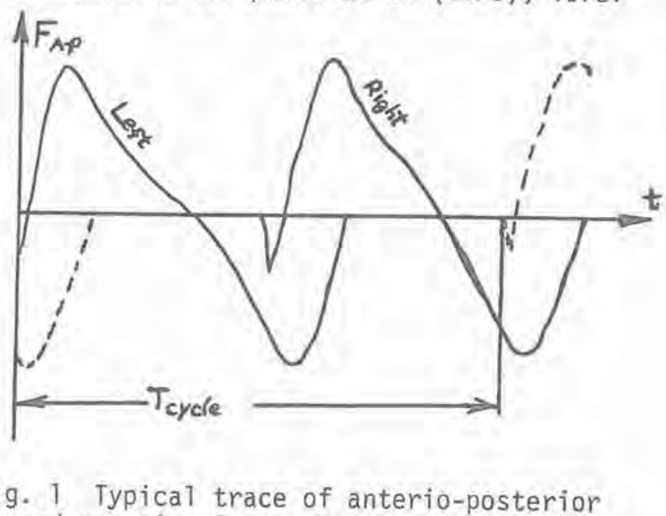


Fig. 1 Typical trace of anterio-posterior ground reaction forces in normal walking.

MOTOR EVALUATION PROCEDURE FOR TESTING HEMIPLEGIC LOCOMOTION.

S. Boccardi, C. Frigo and A. Pedotti, Systems Theory Center-CNR- and Bioengineering Center - Politecnico di Milano, Italy.

## INTRODUCTION

Despite the great number of works dealing with qualitative descriptions of the clinical appearance of hemiplegic gait, few of them are based on rigorous measurements and data processing. A better understanding of the differences between normal and pathological gait as well as

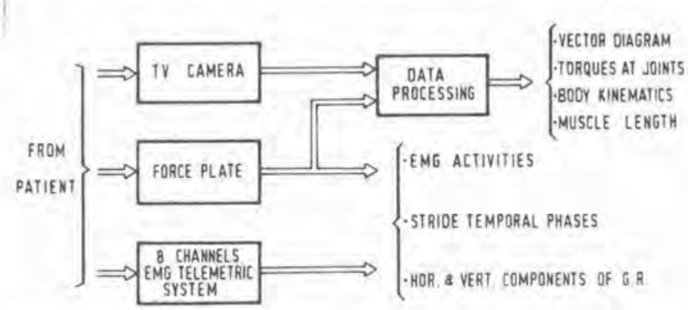


Fig. 1 - Scheme of Motor Evaluation Procedure

a correlative analysis of the biomechanical and neuromuscular events in each patient are the necessary tools for an objective evaluation of treatment methods and an important basis for improving the knowledge of the mechanism of the disability.

This paper presents a motor evaluation procedure(MEP) which is used to investigate hemiplegic patients. This procedure does not disturb the natural gait of subjects and it provides simultaneously the main variables characterizing the biomechanics, the physiology and kinesiology of each stride.

## PROCEDURE AND EXPERIMENTAL APPARATUS

Three different kinds of measurements are simultaneously performed on the patient walking along a pathway 10 meters long.

Fig.1 shows a schematic representation of MEP. The telemetric system with eight channels has been developed for this special purpose. The system involves the use of surface electromyography and microswitches under each foot, so that the EMG signals can be sincronized with the phases of the stride and with the other variables measured (ground reaction and kinematics).

The most relevant part of MEP provides a graphic representation of the mechanical torques at hip, knee and ankle joints from direct measurements of ground reaction and kinematics. It consists in : a) a force plate with four piezoelectric transducer; b)a special made hybrid computer which provides, on-line, on a display the vector representing, in instants of time uniformly discretized, the ground reaction with its amplitude, inclination and point of application. Simultaneously a TV camera takes the subject wearing suitable markers for a better identification of the various body segments and their centers of relative rotation.

The image of the instantaneous resultant vector and that of the subject walking join in a mixer which performs a superimposition. The vector sampling and the TV images are syncronized between them so that a vector appears for each image. The mixed images, providing a graphic representation of the instantaneous values of torques at joints, can be directly observed or recorded by a video tape recorder. Fig. 2 is a typical picture of a hemiplegic patient directly taken from the monitor.

In this picture it can be easily observed the resultant vector acting on the lower limb of

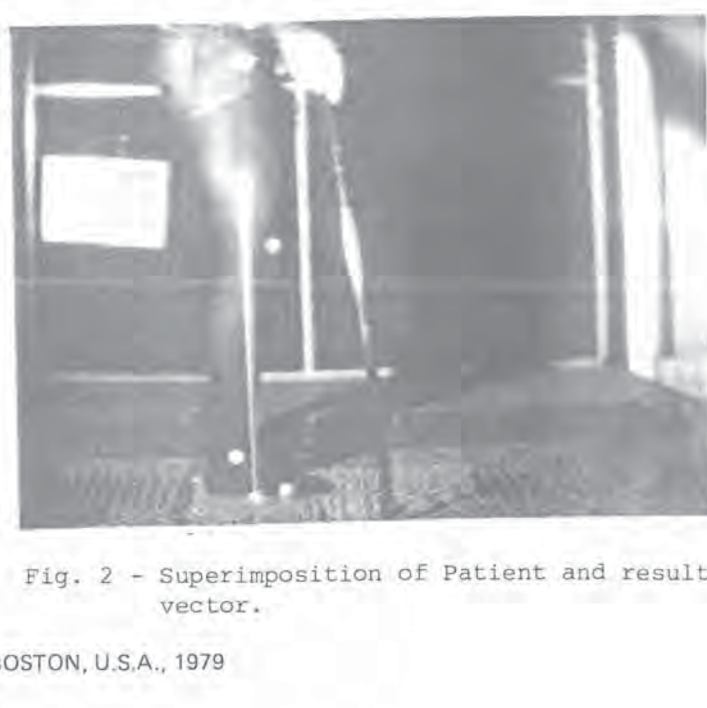

Fig. 2 - Superimposition of Patient and resultant vector.

the patient and the markers pointing out the rotation centers of various joints. An immediate evaluation of the torques applied to each joint is then possible by simply measuring the vector and its arm with respect to each joint. The angular coordinates are processed by a digital computer which provides the instantaneous length of muscles (see Frigo and Pedotti,1978).

Moreover, by simply storing the sampled vectors it is possible to obtain the vector diagram of gait which represents a significant synthesis of it (see Boccardi et al., 1977).

## RESULTS

The final goal of this research it to obtain for each patient a correlative analysis of the following variables: a) Torques at joints; b) angular and body coordinates; c) EMG activity of the main muscles; d) length of the considered muscles; e) vector diagrams; f) temporal phases of each stride.

The subjects are selected according to the following criteria: 1) onset of hemiparesis at

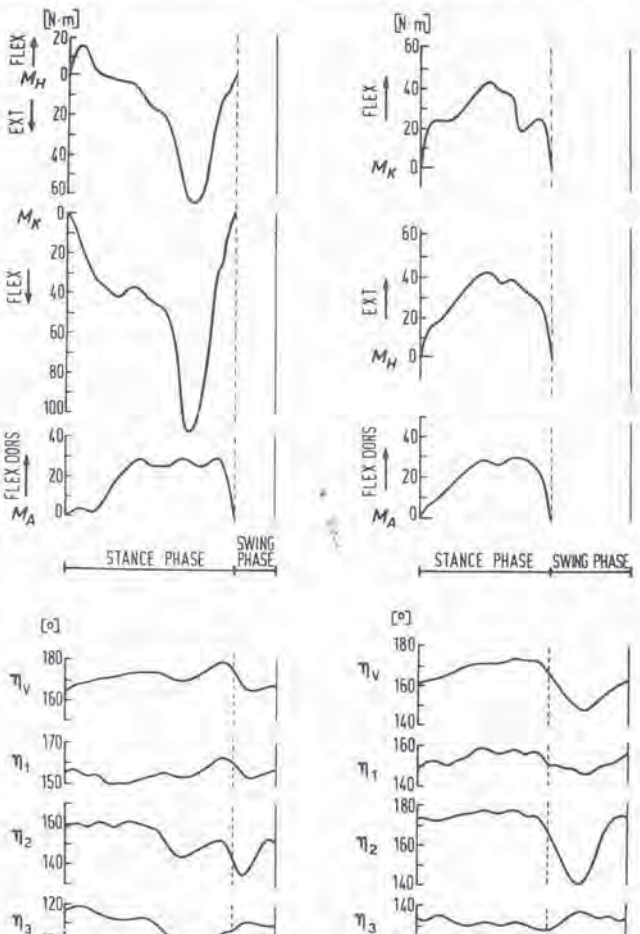


Fig. 3 - Some of the obtained results.See text

least three months before; 2) ability to walk with or without a cane. Several investigations are performed in different periods during rehabilitation.The results will be compared with those already obtained from normal subjects (see Pedotti, 1977, Pedotti et al. 1978, Frigo & Pedotti, 1978).

Fig. 3 shows some of the results obtained from the presented procedure. They refer to the healthy leg(on the left)and to the affected leg (on the right). From the top to the bottom are reported: a) the mechanical torques at the hip, knee and ankle. b) The angles between thigh and vertical, thigh and pelvis, thigh and shank, shank and foot respectively.

It is important to note that the torques are only those produced by ground reactions. Therefore they appear zero during the swing phase (see Pedotti et al. 1978).

Fig. 3 suggests the following preliminary observations:

a) Torques on the affected leg are always of the same sign during the stance phase.
b) The angular coordinates of the same leg point out the strategy used by the patient in order to flex the thigh.
c) The torque at knee of the healty leg is remarkably flexor during the second part of the stace phase,requiring the activity of extensor muscles. The EMG activity of muscles not here reported for reason of space, are in agreement with the torques obtained.

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# GAIT TIMING IN NORMAL AND HEMIPARETIC CHILDREN AND ADULTS


| | | |
|---|---|---|
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| V. L. Grillo | Istituto Pediatrico G. Gaslini, | Genova, Italy |
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The analysis of gait timing is an easy and inexpensive procedure for the assessment of a patient's performance and for the evaluation of the efficacy of treatment. By means of special switches(tapeswitches) placed under the heel and under the first metatarsal joint with adhesive tape it is possible to record the following main parameters: step and stride duration,duration of the brake, foot-flat, push-off and swing phases on each side, duration of the double and single support phases. These time intervals may be recorded on an analog chart recorder using different pen positions for the coding of the various phases of interest. They may also be automatically measured by a microprocessor system for later processing according to specific programs (1,2). Despite the more cumbersome data processing involved, analog recording was chosen for this investigation because allows an immediate visual inspection and evaluation of the gait pattern.

This is a preliminary report of a work in progress carried out on normal, hemiparetic and C.P. children at the Istituto G.Gaslini in Genova and on normal and hemiparetic adults at the Ospedale Generale Regionale in Monza. The work on hemiparetic subjects has been specifically focused on the evaluation of functional electrical stimulation devices (F.E.S.).

## MATERIALS AND METHODS

Normal subjects - Seventeen normal children (12 females and 5 males) within the age range of 9 to 13 years and 11 normal adults (5 females and 6 males) within the age range of 18 to 28 years were fitted with the switches and asked to walk barefooted back and forth along a 12 meter walkway at their natural cadence. In some cases the subject was also asked to walk faster or slower than normal. Five to eight complete gait cycles (10 to 16 steps) in between turns were analysed. The average value and standard deviation of each phase normalized with respect to its cycle duration were computed.Observation of variability of gait led to the conclusion that the analysis of less than 8-10 steps is not statistically sound.

Hemiparetic subjects - The same procedure was applied for the assessment of hemiparetic subjects walking with or without a functional electronic peroneal brace,before and after a period of treatment.Both the orthotic and the therapeutic effect of the treatment are under investigation. Statistical evaluation of these results is not as easy because of the wide range of individual performances. Only results of individual cases will therefore be reported at this stage of the work.

## RESULTS

Normal subjects - A first observation concerns the relationship between natural cadence (steps/min) and age. A scatter diagram of these two variables is shown in Fig.1. A pattern of decreasing cadence with increasing age is evident in the children group even if the correlation coefficient is not very high.No specific correlation appears for the adult group. A correlation between gait phases and cadence (more marked in the children group) showed that the normalized brake, push-off and swing phases increase with cadence while the foot-flat phase decreases.

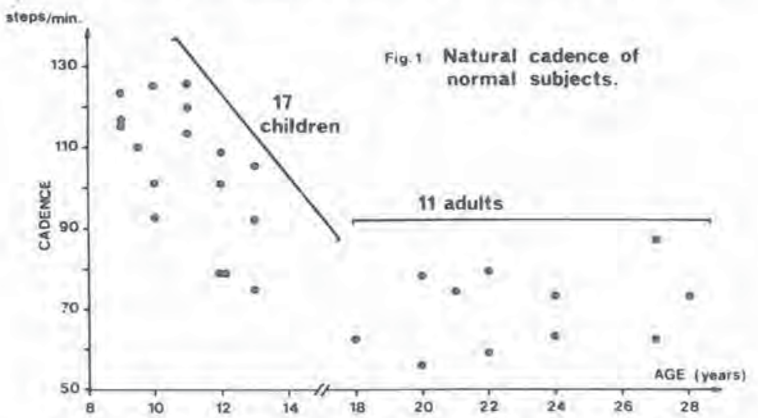
Fig.2 shows the average values and the standard deviations of the four phases for three cadence ranges of the children group and for the adult group. This correlation was observed both in individual subjects walking at different cadences and in different subjects walking at their natural cadence.The analysis of the single and double support phases is under way.

Hemiparetic subjects - Eleven hemiparetic adults and twelve hemiparetic children were examined. The main characteristics of the hemiparetic gait appeared to be: a) a toe strike or a very short brake phase on the affected side; b) a shorter foot-flat and a longer push-off phase on the affected side; c) a cadence much lower than normal (particularly in adults); d) a greater degree of variability from step to step; e) foot-flat longer and swing phase shorter on the non-affected side (particularty in adults).

EFFECT OF F.E.S. - Seven children within the age range 7 to 13 (4 males, 3 females) and four adults were considered to be good candidates for F.E.S. and used a peroneal brace as an orthotic device for 20-30 min. daily. Within one to two months the gait of six children improved (very substantially in two cases) while one did not change significantly even if it appeared clinically better. One of the improved cases is reported in Fig.3. Increase of brake phase duration on the paretic side is evident as well as a better matching with the normal pattern shown in Fig.2. Natural cadence is also increased.

Data on therapeutic benefit on adults are not yet available, however an orthotic effect has been observed in most cases. Fig. 4 shows a recording from a hemiparetic subject walking with the stimulator OFF and ON. Toe strike on the paretic side and weight shifting during the foot-flat phase on the non-affected side are evident. Both problems disappear with the stimulator ON even if the gait remains markedly asymmetric.

F.E.S. appears to have an orthotic and a therapeutic effect on gait timing and not only on muscle force as previously reported (3). From the functional standpoint this effect is certainly more relevant.



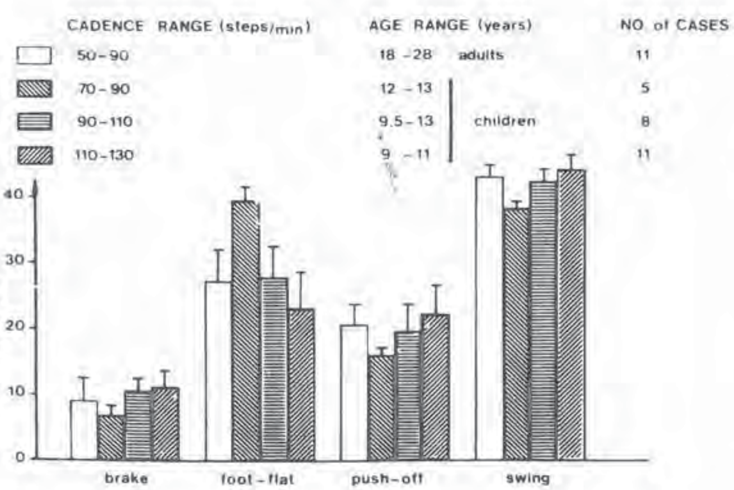




Fig.2 Gait phases in normal children and adults

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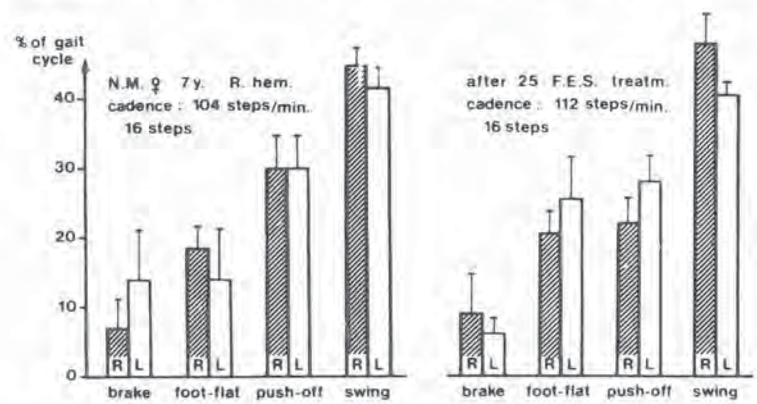


Fig.3 Natural gait of subject N.M. at the beginning of F.E.S. treatment and one month later.

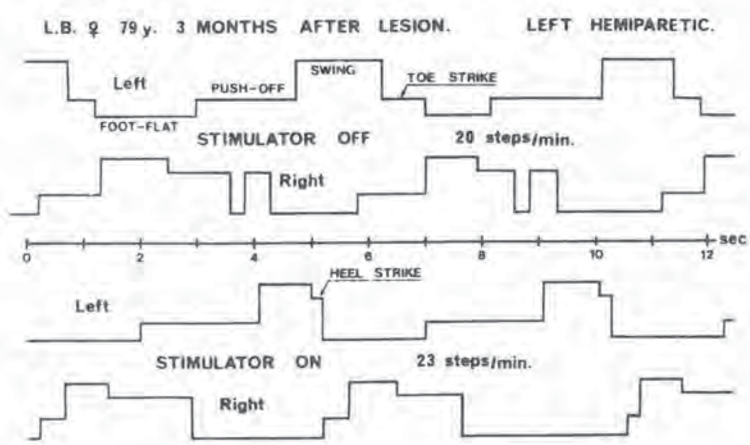


Fig.4 Recording of hemiparetic gait without and with F.E.S.

# GAIT ANALYSIS OF THE LEG AFTER 'CHARNLEY' KNEE ARTHROPLASTY

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## INTRODUCTION

Subjective measurement of pain and range of static movement are tests that are used for assessing the function of patients who have undergone knee surgery. Since the advent of electrogoniometry there has been increasing use of further functional assessment tests such as range of movement, velocity, cadence and stride length (Lamoreux, 1972). All these factors affect the forces acting at the joints in the lower limb during walking and must in particular dictate the forces acting on any prosthesis inserted into these joints (Paul, 1970). Using similar techniques to those described by Collopy, et al., (1977) we conducted cadence and range of movement studies at the knee on patients who had undergone knee joint replacement using the Charnley 'load-angle inlay' prosthesis.

## MATERIAL AND METHODS

29 patients of mean age 55 years were examined and subjected to gait analysis 2-3 days prior to, and 3-5 months after unilateral knee replacement using the Charnley knee prosthesis. 21 patients were diagnosed to have rheumatoid arthritis and 6 patients general osteoarthritis. 45 age-matched (mean age 54 years) with no symptom of knee joint disorders were also tested under the same conditions.

Transducers were attached to the thigh and calf by 'velcro' straps and the signals were analysed in the central processor of a polarised light goniometer system. The output from the processor is in the form of a variable voltage related to the angular displacement of the transducers. A simple heel switch placed in the shoe was used to indicate heel strike of the measured leg and the cycle time for each patient.

## RESULTS

The range of sagittal movement at the knee during walking of patients before knee surgery was 14 degrees compared with 73 degrees for the normal group. However, after replacement with a knee prosthesis, the average improvement was by 22 degrees but still only 50 per cent of the normal range at the knee (see Figure 1). Post operatively, all the patients had on average increased their cadence to nearly as high as the normal group (Figure 2), however, their stride length was smaller and consequently their velocity less. There was a large difference between the smallest and largest value of knee movement range, but all the patients improved in knee flexion after surgery.

## DISCUSSION

The gait analysis of these patients has shown quite clearly that there has been an improvement by all the patients in their range of movement at the knee, walking speed and stride length when examined after a period of 3-5 months post-operation. However, compared with the normal age-matched group they were well below the performance of walking ability but adequate to fulfil the basic functions of level walking without pain. The greatest improvement was with the patients who had rheumatoid arthritis, presumably because of knee extensor muscular weakness and a larger incident of flexion contracture before surgery. The range of movement tests confirm however that further physiotherapy is still needed so that patients are able to walk up and down stairs, and sit and rise from a chair.

It is necessary to conduct pre and post operation kinesiologic tests such as those conducted to provide dynamic assessment of joint functional movement under weight-bearing which tells us much more than static un-loaded flexion and range of movement studies previously undertaken.

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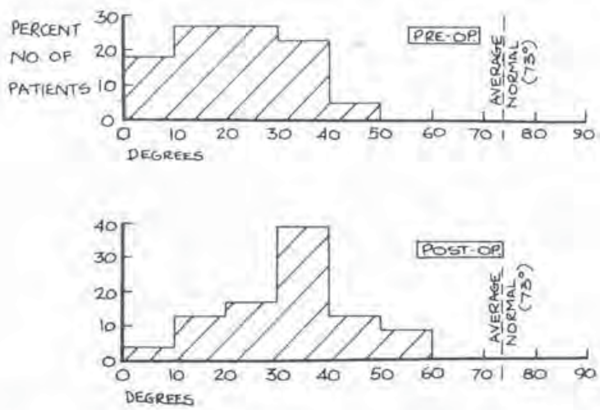


Figure 1 : Range of movement at the knee pre- and post-operation for 21 patients who underwent knee arthroplasty. The average range of movement for 40 age-matched normals was 73 degrees at the knee.

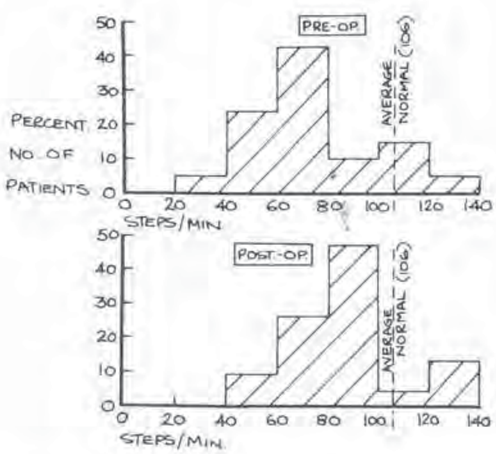


Figure 2 : Cadence pre- and post-operation for 23 patients who underwent knee arthroplasty. The average cadence of the age-matched normals was 106 steps/minute.

# FRACTIONATED KNEE EXTENSION RESPONSE TIME, REFLEX TIME, AND FIBER TYPE COMPOSITION IN RELATION TO AGE AND ACTIVITY LEVEL

Priscilla M. Clarkson, Ph.D., Walter Kroll, P.E.D., and Anthony Melchionda, M.D., Department of Exercise Science, University of Massachusetts, Amherst, MA. 01003

## INTRODUCTION

It has been clearly demonstrated that the speed of reaction time and movement time is considerably slower in aged individuals (Clarkson, 1978a;Kroll and Clarkson,1978). However, older persons who have been physically active throughout their lives did not exhibit the long response speeds found in inactive, sedentary individuals (Clarkson,1978a;Spirduso,1975). By fractionating reaction time into premotor time (corresponding to central processing time) and motor time (corresponding to peripheral processing time) components, it has been found that motor time is the least affected by age or activity level of the subjects (Clarkson,1978a). Contrary to reaction time and movement time findings, reflex time has been shown to be unrelated to the subject's age or level of activity (Clarkson,1978b). Since the above results suggest that the aging process predominantly affects central rather than peripheral processing, the present study was undertaken to examine the relationship of vastus lateralis fiber type composition and fractionated response and reflex time.

## PROCEDURE

Eight old active (57-75 yrs.), seven old inactive (57-67 yrs.), and six young (20-30 yrs.) male subjects volunteered to participate in the present study. Active older men were defined as those who maintained an exercise program at least three times per week for most of their adult lives. The young subjects were college students who were not involved in any sports training program.

Subjects reported to the laboratory on two separate days during which time 25 knee extension response measures were secured. For the response measures the subject was seated on a specially designed table with his heel resting upon a microswitch. The stimulus-response apparatus containing an NE-51 neon bulb and padded target was located directly in front of the subject. When the light stimulus was presented, the subject was instructed to kick the padded target which fronted a second microswitch. The target board was positioned in such a way that when the target was depressed, the subject's femoral-tibial angle would be at 120 degrees. The time from the flash of the light stimulus to release of the first microswitch was designated total reaction time. The time from the release of the first microswitch to the depression of the second microswitch was designated movement time.

Total reaction time was further fractionated into premotor time and motor time. When the stimulus light flashed it activated a light beam across a Teca oscilloscope screen. The location of the action potential on the oscilloscope time scale furnished a measure from the onset of the stimulus to arrival of the neural impulse to the motor point. This measure constituted the premotor time component. Motor time was derived be subtracting premotor time from total reaction time.

Ten patellar reflex trials were also taken on each testing day. The reflex trials were assessed using a Lafayette knee reflex apparatus: Complete methodology is described elsewhere (Clarkson,1978b). Using an electromyographical technique, total reflex time was fractionated into reflex latency and reflex motor time.

After the subjects had completed the response and reflex testing sessions, they reported to the University Health Service for a biopsy of the vastus lateralis muscle. The muscle sample was removed using a standard needle biopsy technique. After removal, the tissue was frozen in isopentane cooled in liquid $N_2$ and cut into 12 micron sections on a cryostat. The Myofibrillar ATPase method was used to classify fibers into slow twitch (ST) and fast twitch (FTa and FTb). The reactions were carried out at PH 9.4 following alkaline (PH 10.3) and acid (PH 4.6 and 4.3) pre-incubations. Fiber diameters were determined using the least diameter method which involved measurement of the greatest diameter across the lesser aspect of the fiber.

## RESULTS

The response time means are presented in Table 1. It can be observed that for total reaction time the Old Inactive group is 25.4 msec. slower that the Old Active group who in turn is 35.4 msec. slower than the Young group. An analysis of variance showed that this difference between groups was highly significant ($p < .01$). Examination of the reaction time components showed that the between groups differences were manifested predominantly in the premotor time component. No significant difference was found between groups for motor time. The movement time means reflected the findings for total reaction time; the Old Inactive group demonstrated the longest times followed by the Old Active group and the Young group. Upon analysis this between groups difference in movement time was found to be significant at the .01 level.

In Table 2 can be found the mean values for the fractionated reflex time components. An analysis of variance of the reflex time components showed no significant between groups difference ($p > .05$). In fact, the groups do not differ by greater than 7 msec.

The results of the muscle biopsy are presented in Table 3. No significant difference was found between groups for ST, FTa, or FTb fiber percentages or diameters ($p > .05$).

## DISCUSSION

The detrimental effects of aging are demonstrated by the long reaction times and movement times in the old groups. The results of the present study confirm other studies which have reported that the difference in reaction time between old and young groups is predominantly the result of a lengthened premotor time component (Botwinick and Thompson,1966;Weiss, 1965;Clarkson,1978a). The longer movement times of the old groups compared to the young may be explained by a central processing dysfunction resulting in an inability to recruit the necessary motor units to carry out the response. Motor time, which is an index of muscle contraction time, seems to be unaffected by age since the mean values for reaction motor time are quite similar between old and young groups. Thus it can be suggested that the aging process exerts the greatest effect on the central processing system rather than the peripheral processing system.

The contention that aging preferentially affects central processing is further substantiated by the findings of no significant difference between groups in reflex time or in muscle fiber type composition. Age related atrophy of muscle fibers was not found in the present study; neither the percentages nor the diameters of the ST, FTa, or FTb fibers differed significantly between groups.

The benefit of a life style of physical activity was clearly shown by the findings of faster total reaction times and movement times in the Old Active group compared to the Old Inactive group. In fact, in movement time, the Old Active group was more similar to the Young group than to the Old Inactive group. Thus, an active life style may retard the age related slowing of reaction time and movement time exerting the greatest effect on the movement time component.

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TABLE 1. FRACTIONATED KNEE EXTENSION RESPONSE TIME MEANS (in milliseconds)

| | Total | Premotor | Motor | Movement |
|---|---|---|---|---|
| Old Inactive | 274.6 | 174.6 | 100.1 | 166.5 |
| Old Active | 249.2 | 155.7 | 93.5 | 117.4 |
| Young | 213.8 | 127.6 | 86.1 | 104.1 |

TABLE 2. FRACTIONATED PATELLAR REFLEX TIME MEANS (in milliseconds)

| | Total | Latency | Motor |
|---|---|---|---|
| Old Inactive | 89.9 | 19.3 | 70.5 |
| Old Active | 89.2 | 19.6 | 69.6 |
| Young | 82.4 | 19.4 | 63.0 |

TABLE 3. FIBER TYPE PERCENTAGES (%) AND DIAMETERS (Dia) (in microns)

| | Old Inactive | | Old Active | | Young | |
|---|---|---|---|---|---|---|
| | % | Dia | % | Dia | % | Dia |
| ST | 41.8 | 56.9 | 39.5 | 49.3 | 38.3 | 46.1 |
| FTa | 43.5 | 47.2 | 48.0 | 47.9 | 43.5 | 53.6 |
| FTb | 14.7 | 38.2 | 12.5 | 34.8 | 18.3 | 44.2 |


ACKNOWLEDGEMENT

This study was supported by a University Faculty Research Grant.

NOTES

# KINESIOLOGY IV

## Session 11B

HANDWRITING MOTOR LEARNING, DEVELOPMENT OF A NEW METHOD EVALUATED BY ELECTROMYOGRAPHY.

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and Department of Anatomy, Université de Montréal, Montreal, Quebec, Canada.

## INTRODUCTION

In occupational therapy, writing is one of the most important daily living activities in the rehabilitation of upper-arm amputated and hemiplegic patients. Jones (1) expressed three basic patterns in handwriting re-education, emphazing shoulder, forearm and hand movements. When re-educating the patient to write, the practice of gross movements must precede fine movements (1). In a different way, Gardner (2) presented a text-manual for remedial handwriting based on graphical practice. However, his exercises did not indicate a step-by-step procedure for individual needs. The aims of the present research is to develop a more precise methodology adapted to the physical and phychological needs of the right side hemiplegic patient.

## EXPERIMENT

Normal right-handed volunteers, male (n=1) and female (n=3), aged from 24 to 35 years participated in this study. They were assigned to a 23 day-training period, half an hour each day, including three electromyographic evaluations. Muscular activity of the left opponens pollicis, extensor carpi radialis brevis, brachio-radialis, flexor carpi radialis, trapezius (superior part) and triceps brachii (caput laterale) was evaluated. The Beckman's surface electrodes were connected to multichannel EMG amplifiers and monitored by a Tektronix four channel scope. Position of the electrodes was tested by requiring the highest possible isometric contraction related to the agonistic function of each muscle. The subject was seated in a predetermined upright position, 12-15 cm from the xyphoid process to the edge of the table. The left upper-arm was supported on the table at an abduction of 30 degrees and flexion of 45 degrees. An experimental orthesis was developed to stabilize the elbow while the forearm (in prono-supination) and hand were free to move in a standard writing position.

EMG evaluation. Within each of the three EMG sessions, the muscular activity of the above muscles was studied for each of 18 graphic signs. These signs, selected from Tremblay (3), reproduced the regular alphabet motor sequence. Efficiency of the left-handed writing performance was insured by positioning the paper reverse to the right-handed subjects. The experiment was modified throughout this study to establish a standardized posture and an experimental control within subjects. Therefore, only the third evaluation is reported here.

Training procedure. The subjects were trained five days a week, for half an hour, early in the morning. Each training period exercises were scheduled to fit into the prescribed time. Subjects performed under given instructions, at their own pace. A group of exercises was given and repeated each day. Exercises progressed from upper-limb gross movements, in standing position at a blackboard to fine hand movements. Thereafter, each subject completed his exercises on paper, but now sitting at a table. The first phase of exercises contained straight horizontal, oblique, vertical lines, circles and letters. The second phase began with alphabet while the third phase processed with the signature and the copy of common words. Constant attention was given to encourage success and hold motivation. The experimenter gave firm and simple commands in order to face the needs of hemiplegic patients.

## RESULTS

EMG evaluation. Within the 18 signs evaluated, eight were chosen to illustrate the most representative criteria of muscular activity (Figure 1). The highest level of activity in at least one of the six muscles under study was observed in graphic test "five" (*ℓℓℓ*). The lowest level of activity was observed in at least one of the six muscles under study, in five of the eight graphic tests (tests 8,9,10, 12,18). A sequential pattern of muscle activity was observed in two tests for all the subjects (tests 7,18). Three out of the four subjects showed a sequential pattern of muscular activity in five tests (tests 3,5,9,10,12). A saccadic normal activity of rapid discharge (12-20/s) was seen in most of the subjects for all the chosen tests (Figures 1 and 2).

Training progress. In general, the main criteria of legibility was reached. The graphic structures, at the end of training, were more regular in form, size and downstrokes. Spacing, alignment and line quality were usually demonstrated.

## DISCUSSION

The present study brought in a new method for the standardization of a handwriting rehabilitation program. According to the literature referred to, such a methodology had never been applied. The EMG was necessary to select the most significant graphic tests illustrating normal particular features of muscular activity during left-handed writing in right-handed subjects. The opponens pollicis, known as an agonist and synergist in thumb function appeared to be more active in the dynamic part of the graphic test. Extensor carpi radialis brevis and flexor carpi radialis and the trapezius indicated precise synergistic functions.

The stable position assumed by the experimental orthesis promotes a more objective analysis of results in reducing the brachioradialis and triceps action.

The improved skill in writing quality is a difficult criteria to be judged by EMG. A skill exercise requires a fine coordination in the timing of the muscular contractions. The observed saccadic activity might be necessary for hand stability during precision tasks.

This preliminary study, done on normals, will give some control criteria which will facilitate the discussion of further results and its application to the hemiplegic patients.

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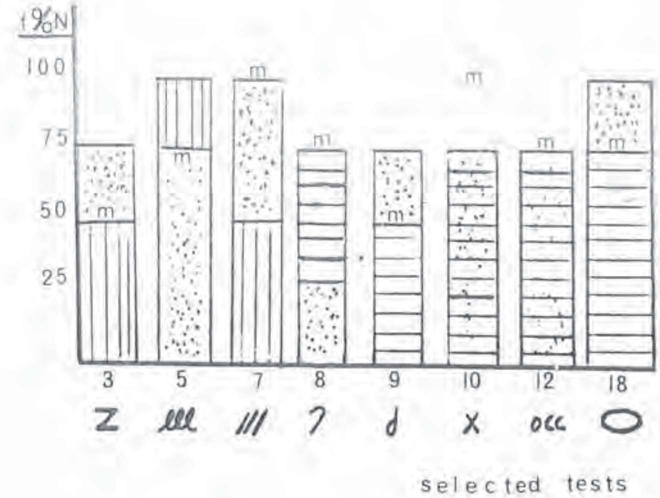


Figure 1. Tests selection in relation to frequency of specific observations in normal subjects.

Highest level of muscular activity (within the six muscles under study)
Lowest
Repetitive periodic observation EMG pattern
m Saccadic Muscular Activity (12-20/s)

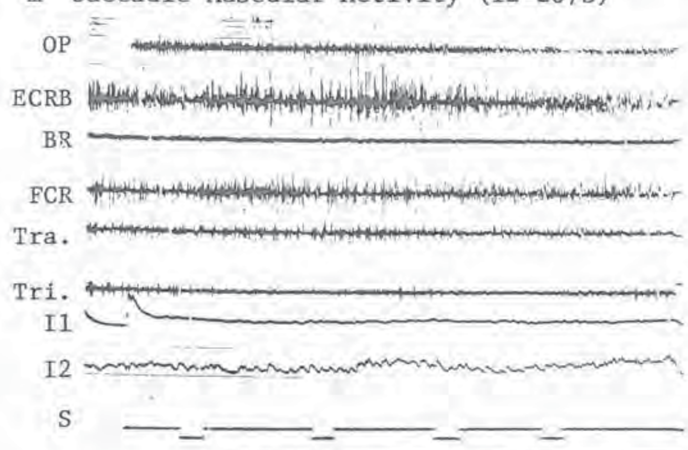


Figure 2. Example of EMG normal saccadic activity, generally observed during graphic tests.

OP: Opponens Pollicis, ECRB: Extensor Carpi Radialis Brevis, BR: Brachio Radialis, FCR: Flexor Carpi Radialis, Tra.: Trapezius upper part, Tri.: Triceps caput laterale, I1: Integration of OP, I2: Integration of ECRB, S: Sequence identification.

## ACKNOWLEDGMENT


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FREQUENCY SPECTRUM OF THE HUMAN MYOELECTRIC SIGNAL AS RELATED TO HANDEDNESS.

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## INTRODUCTION

Several investigations have demonstrated a difference in the frequency spectrum of the myoelectric signal obtained from various muscles (Larsson, 1975; Kadefors, Petersen and Broman, 1973; Herberts, 1969; and Gersten et al. 1965). These investigations were not concerned with a systematic description of normal muscles, but only with the possible difference from abnormal muscles. In these investigations the frequency spectrum was obtained by multiple filter bank measurements. A previous investigation by Lehr and Kasser (1978) demonstrated a difference in duration of task as related to handedness. It was suggested that the mechanism for this difference could be attributed to the increased capability of the dominant motoneuron pool to selectively rotate the motor units to obtain a higher degree of skill and thus sustain the duration of the task. Based on the assumption that spectral changes can be brought about by rotation of activity between motor units and/or the average potential duration of the acitve motor units (Kadefors, Petersen and Broman 1973) this investigation was initiated as a pilot for a larger task oriented investigation related to handedness.

## EXPERIMENT

Six young adults were selected from a group of volunteers. The selection was based on declaration of right handedness and apparent normal musculoskeletal system. Handedness was subsequently confirmed and ranked in completeness by the Edinburgh Handedness Inventory. A pair of surface electrodes were placed first over the first dorsal interosseous muscle of the right hand then over the same muscle of the left hand. The subjects were grounded, and a maximal contraction was elicited of ten seconds duration. The EMG signals were stored on FM tape for subsequent computer analysis. An analog to digital conversion was performed and the digitized signals were submitted to the Biomedical Data Package 03T time series program for a fast fourier transform. The parameters of the program were established to provide a continuous frequency spectrum from 0-500Hz.

## DATA ANALYSIS

The resulting frequency spectra were subjected to multiple analytical manipulations. The primary and secondary frequency peaks (two highest values) of each spectrum were selected, averaged and then analyzed by t-test were normalized so that an analysis of the patterns above 50% of the full graphed power (amount) of the signal component were demonstrated. The spectrum was divided into three regions: a lower third from 0-156Hz, a middle third from 171-328Hz, and an upper third from 343-500Hz. Perusal of the pattern obtained by this normalization process formed the basis for the results of the investigation.

## RESULTS

The test for significance of the primary and secondary frequency peaks was non-productive with regard to the separate thirds of the frequency spectrum (Table I). There was a distinct peak pattern in the dominant hand (right-Fig I) when compared with the non-dominant hand (left-Fig 2). This pattern was characterized by fewer peaks above 50% of the full graphed power of the frequency spectrum. A predominant frequency was not evident across subjects but each subject revealed fewer peaks of frequency for the dominant hand. Subject number 6 revealed a reverse pattern of activity and the possibility exist that the handedness inventory was not definitive for this subject. If the data is regrouped with this in mind the t-test is significant ($t_{10}=3.57$, $p=0.005$; right first dorsal interosseous M 2.16, SD .9, and left first dorsal interosseus, M 5.00, SD 1.6) for the number of frequency peaks.

## DISCUSSION

The difference in the pattern of the frequency peaks for the right hand (Fig I) and left hand (Fig 2) can be attributed to the skill level of the most frequently used (dominant) hand. This greater degree of skill is reflected in the limitation of the number of peak frequencies utilized by the motoneuron pool of the dominant hand. This limitation of frequencies is reflective of the possible high level of rotation of separate motor units utilized to obtain a few specific frequencies which maintain a maximum control over the skill level. On the other hand, the left, the pattern of multiple frequency peaks is indicative of the low level of motoneuron pool control over the rotation of the motor units. This would account for the greater number of frequency peaks observed in the non-dominant hand.

It can be noted that the specific frequencies of the dominant hand were not consistant between subjects. This observation may indicate that for the task selected the frequencies obtained were not specific for the particular task. It is of course possible that each specific pattern of frequency is an individual characteristic of the training of the motoneuron pool.

## CONCLUSION

These results suggest an emerging measureable distinct peripheral physiological motor parameter which is reflected in the motoneuron pool and is directly related to handedness. The enlarged investigation will expand the types of task involved and electrode sites to provide a more definitive statement on these findings Should the larger investigation support these findings then this technique offers the possibility of insight into sub-human handedness, lateralization and its role in human evolution.

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Table 1: Frequency Peaks above 50% Full Graphed Power in Thirds of Spectrum

| Subject Rt Hand | 0-156Hz | 171-328Hz | 343-500Hz | Peaks |
|---|---|---|---|---|
| 1 | 93* | | | 1 |
| 2 | | | 359* | 1 |
| 3 | 93* | 312** | | 2 |
| 4 | | 328** | 375**,468 | 3 |
| 5 | 78 | 281** | 375* | 3 |
| 6 | 31 | 203*,250 281**,328 | 375 | 6 |
| Lf Hand | | | | |
| 1 | 31*,140 | 218** | 359 | 4 |
| 2 | | 218,296* | 359**,406 | 4 |
| 3 | 62* | 218,312** | 390 | 4 |
| 4 | 83,156 | 187*,203** | | 4 |
| 5 | 46 | 171*,203 250,312 328 | 390**,437 | 8 |
| 6 | 31** | | 359*,468 | 3 |

*primary frequency **secondary frequency

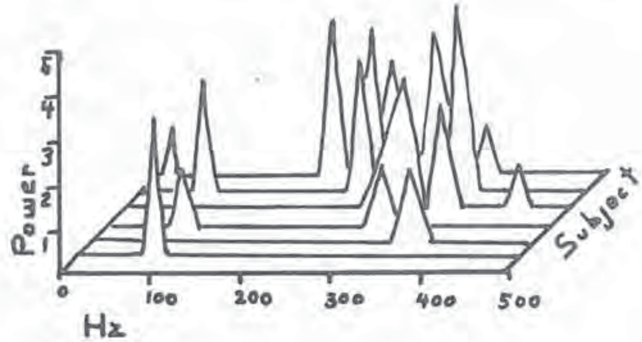


Fig. 1

Right hand pattern of frequency peaks above 50% of full graphed power by subject.

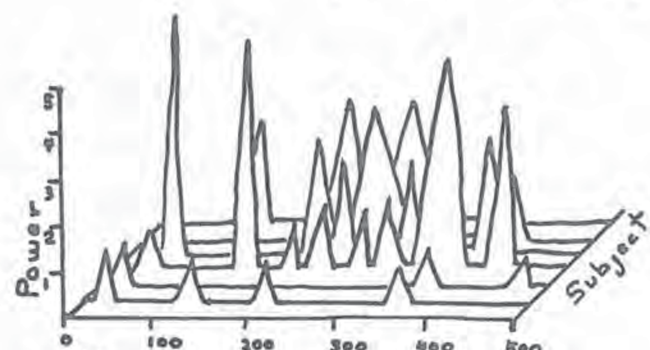


Fig. 2

Left hand pattern of frequency peaks above 50% of full graphed power by subject.

# EFFECTS OF FORCED EXERCISE ON NORMAL AND MYOPATHIC HAMSTERS: AN ELECTROMYOGRAPHIC STUDY

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## INTRODUCTION

For a number of years, experiments have been conducted to determine the effects of chronic exercise programs on myopathic animals (1,2,3,4,5). To date however, the validity of the use of exercise as a therapeutic tool in the management of various myopathies is yet to be determined. While histologic, histochemical, and biochemical alterations mediated by specific exercise programs have been identified, no myoelectric studies have been conducted to parallel the aforementioned studies. The present experiment was undertaken to investigate and compare the effects that chronic exercise has upon myoelectric parameters for affected and non-affected animals.

## EXPERIMENT

Eight male Syrian hamsters, Mesocricetus auratus (inbred line, autosomal recessive, B10 R 53.58) were used as experimental animals. Eight normal male Syrian hamsters from the same supplier served as controls. The study was designed to include four treatment groups: (1) myopathic (unexercised) animals housed in sedentary cages throughout the experiment, (2) myopathic (exercised) animals housed in sedentary cages while being exercised, (3) normal (unexercised) animals housed in sedentary cages throughout the experiment, and (4) normal (exercised) animals housed in sedentary cages, but subjected to the forced exercise program. Exercise was conducted five days per week and the program consisted of progressively increased swimming time and resistance in the form of a percentage of daily body weight attached to the animals while swimming.

The exercise program progressed from day one (animal age - 30 days) with 15 minutes of swimming to days 27 and 28 of the program where each subject group was able to swim for 60 minutes with the addition of 3% of body weight attached. Day 27 was used to gather pretest myoelectric data for the normal hamsters (66 days of age) while the myopathic hamsters were examined on day 28 (67 days of age). Attempts to gather data prior to this time were unsuccessful as the animals were too small for thorough examination. Both groups were reexamined after 40 days of resisted swimming. At that time, the myopathic animals were 87 days old and the normal animals were 88 days old.

For each electromyography examination, the animals were anesthetized with an intraperitoneal injection of sodium pentobarbital (65 mg/kg). Recordings were performed with two monopolar needle electrodes (Grass E2 platinum alloy electrodes - length 11 mm, diameter, .30 mm). One needle was inserted into the muscle belly while the other was placed subcutaneously so as to act as the indifferent electrode. The same electrodes were used throughout the study. The signals were processed by a Grass Model 7 Series Polygraph. The polygraph was equipped with Grass Model 7P3 preamplifiers and Grass Model 7 driver amplifiers. The high frequency response of the Model 7P3 is down 10% at 10 kHz. and down 50% at 20 kHz. while the low frequency response is 50% at 10 Hz. The high frequency response of the driver amplifier is down 50% at 80 KC while the low frequency response is 50% at 10 Hz. Myoelectric signals were monitored by an audio speaker and a Textronix 564 storage oscilloscope. All data were stored on magnetic tape by a Philips N4504 reel to reel tape recorder.

Nine anatomic sites were monitored unilaterally on each animal including: biceps brachii muscle, triceps brachii muscle, flexor forearm mass, pectoral mass, tibialis anterior muscle, posterior leg mass, anterior thigh mass, medial thigh mass, and gluteal mass. After insertion of the electrodes and with the muscle at rest, recordings were made to observe the presence or absence of spontaneous activity.

Single motor unit activity was elicited by invoking a light myotatic reflex. Muscle activity needed for the production of an interference pattern was also elicited via myotatic reflex. For this, the segment or joint was manipulated until maximal response was achieved. This was followed by recordings occurring from needle movement and muscle percussion.

Data reduction consisted of measuring the duration of the potentials, that is, the time between the point where the deflection left the baseline and the point where the deflection returned to the flat baseline. Amplitudes were measured from peak to peak. Polyphasic potentials were taken as those with more than four phases. These were included in the duration time measurement. Interference patterns were filtered, digitized and submitted to a spectral analysis to determine the frequency components.

This work was supported by an NIH grant (NS10716-03, ORD 20267).

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A MODEL FOR THE ANALYSIS OF SKELETAL MUSCLE FUNCTIONS

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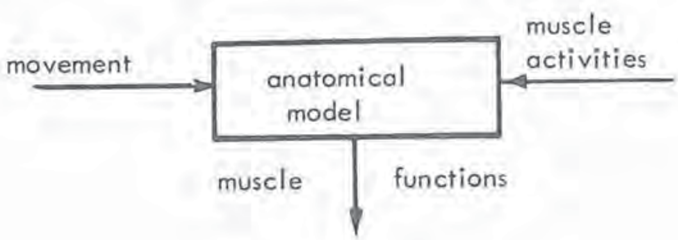
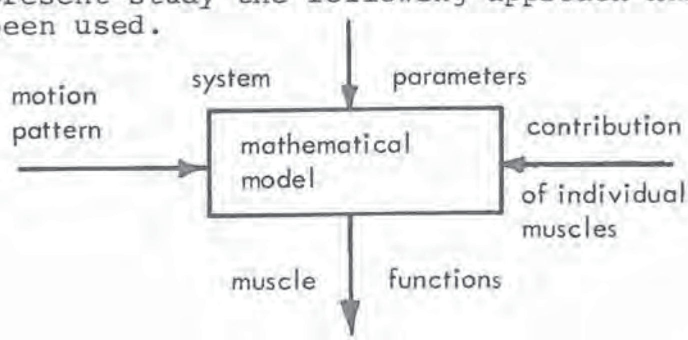
When recording an abnormal locomotion pattern of a patient it is essential for a correct diagnosis and appropriate treatment to have adequate understanding of the functions of the muscles. For studying these functions the approach most commonly used is to prescribe a specific movement and to record the myoelectric activities. (Hirschberg and Nathanson-1952; Greenlaw and Basmajian-1968; Janda and Stará, Wheatly and Jahnke-1951; Greenlaw-1973; Basmajian-1974).
Combining these data with anatomical observations muscle functions are then deduced. This is shown in the next scheme.




However, more quantitative methods are needed to answer the question whether a muscle actually produces the movement, contributes to motion coordination or, in cases of abnormal gait, derives its activity from the disorder of the neuromuscular system. To obtain more detailed information of the roles the individual muscles play in movements, in the present study the following approach has been used.




Important system parameters are:
1. topography of the muscles and of their attachments in relation to the skeletal elements.
2. fiber arrangement and mechanical properties of the muscles.
3. number and type of motor units.

A computerized mathematical model has been developed in order to study the influences of these parameters on muscle functions. In this report only one of the parameters will be dealt with, viz, the skeleton-muscle geometry of a single-headed bi-articular muscle. Two situations will be analysed. In the first one the muscle is represented as a single line element. In the other one the muscle is segmented into four straight line elements simulating the actual position of the muscle in relation to the skeleton. The muscle chosen is the biceps femoris because of its importance in posture and locomotion.

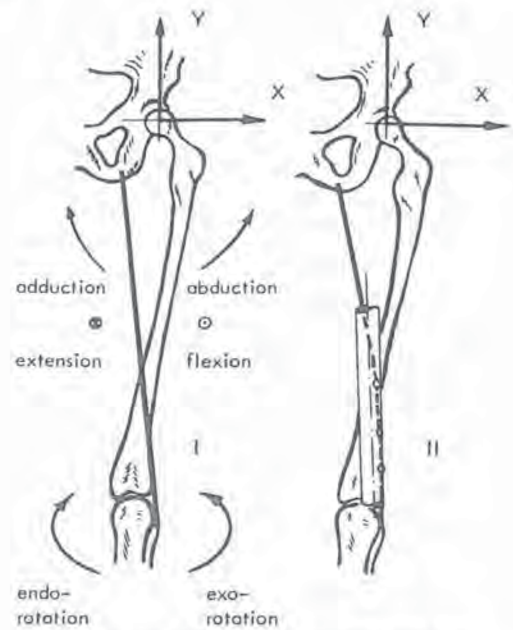



Fig. 1 biceps femoris, represented by one line element (I) and by 4 line elements (II)

The model calculates the resistance exerted by the biceps femoris during flexion and extension in the hip joint. The mechanical properties of the muscle are assumed to be as shown in fig. 2.

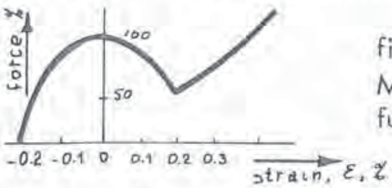



fig.2
Muscular force as a function of the strain

Results of the calculations are given in fig. 3. The resistance against flexion is indicated as the extension moment, against endorotation as exo-rotation moment and against abduction as the adduction moment. The solid lines refer to the single line simulation (I), the dashed lines to the segmented muscle representation (II).

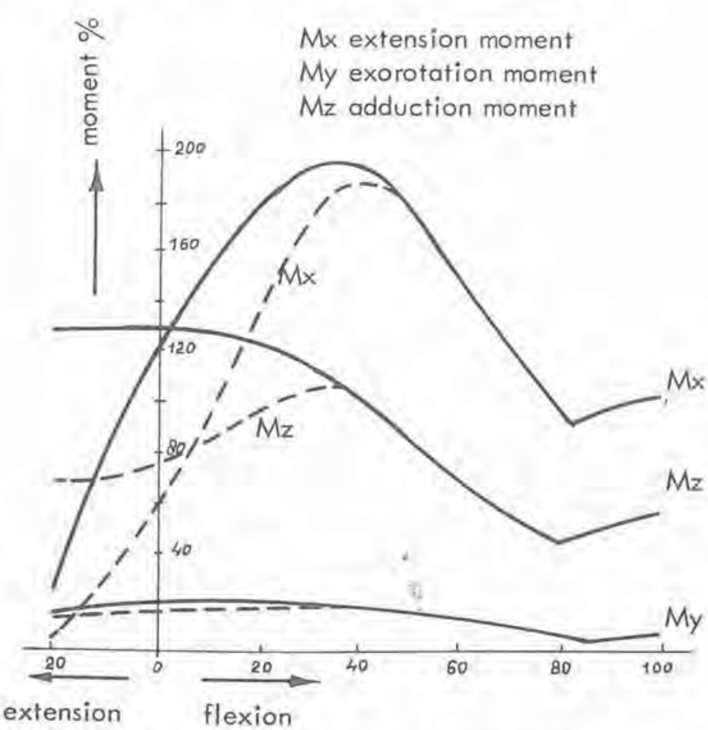



Fig.3 Moment at the hip joint in relation to flexion and extension.

Assuming the muscular force-strain relation to be correct, the biceps femoris will become insufficient at a flexion angle of 80 degrees.
The maximal flexural resistance of the biceps femoris occurs at about 35 degrees flexion.
During flexion the femur tends to abduct. This action will decrease by increasing the flexion.
In both situations the resistance against endorotation can be neglected with respect to the other resistances.
The difference between the solid line and the dashed line shows that the influence of the geometry cannot be neglected. In situation II with regard to situation I the extension- and adduction moments will be halved at zero degrees of flexion.

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# MULTIDIMENSIONAL CORRELATION OF MUSCLE PROPERTIES

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A multidimensional analysis of certain aspects of isometric contraction of muscle, as measured by many workers, may explain some scatter of data and suggest implications for prosthetics and orthotics.

## FORCE-LENGTH (F-L)

Blix (1895) measured isometric F-L curves by electrical stimulation of isolated muscle. Inman and Ralston (1954) studied voluntary contraction by arm amputees with cineplasty tunnels. Maximal available force decreases rapidly at shorter muscle lengths, becoming negligible at about 0.55 Rest Length (RL). The portion beyond (RL) depends on relative strength of the particular muscle. (The net useful force also depends on the passive stretch curve, leading to serious consequences of a contracture.)

## FORCE-LENGTH-TIME (F-L-T)

Monod and Scherrer (1957), as well as Rohmert (1960), studied the maximal time during which isometric muscle forces could be sustained, plotting holding time versus fractions of a "maximal" force which can be sustained in specific position for 0.1 min. A force of 0.5 maximal can be held about 1.1 min.; at 0.2, about 6.5 min., and below 0.15, indefinitely.

Suppose that the maximal voluntary force at each point in the idealized F-L diagram is Rohmert's maximal. Three orthogonal axes, Fig. 1, may represent force F, muscle length L, and time T. Then the F-L curve is plotted on one "wall," at time 0.1 and the F-T curve (rotated 90 degrees) can be plotted on a perpendicular vertical plane through rest length. This F-T curve descends so sharply after piercing the F-L plane at a time corresponding to 0.1 min. that any variation in holding time could result in substantial variation in measured forces. Similarly F-T curves could be drawn on parallel planes for other lengths.

Rest Allowance

Rohmert (1960) reported studies of fatigue and recovery from repeated isometric contractions and calculated rest allowances. A force of 0.50 maximal exerted for 0.5 min. (nearly half of its maximal holding time, 1.1 min.) requires approximately 2 minutes, or 4 times the exertion time, rest allowance.

## FORCE-MYOELECTRIC ACTIVITY (F-EA)

Disagreements among investigators who have studied the relationship between muscle force F and myoelectric activity EA may partially be due to different techniques, instruments, or even muscles and locations of electrodes. Inman, et al. (1951) and many later investigators used electronic integrators though time constants varied. Lippold (1952) recorded directly the amplified and rectified myoelectric activity, then integrated the graphical record. This activity was always directly linearly related to muscle force. Kuroda, Klissouras, and Milsum (1969) used an analog computer. Relative distances between electrodes and active muscle fibers might be important. Conventionally integrated myoelectric activity EA from surface electrodes typically will be less than proportional to F, both normalized, from the origin to about 0.75 maximal F; at high F, EA dramatically increases, Fig. 1.

Below 0.15 maximal F, EA for a given F does not increase with T. At 0.30, EA shows little scatter and very slow increase whereas at 0.80, there is broad scatter and steep increase in electrical activity during the short holding time available (Laurig and Rohmert, 1968).

The F-T curve implies the need for locks in body-powered artificial arms to resist large loads for long periods. If conventional EA is used to control prostheses, low fractions of available maximal forces should be used. Many myoelectrically controlled prostheses have been binary (off-on) or three-state (off-on-reverse), requiring brief bursts. With increasing efforts directed toward continuous proportional controls, the highest muscle force required probably be close to 0.15 maximal, yet risks of inadvertent operation must be low. Other processing methods and electronic circuits, e.g., Isidori & Nicolo (1966) and Childress, et al. (1973), can change this shape. It would seem desirable to give more EA reliably at low F. Amputees might exercise to maintain a standard strength.

In the rotated axes of Fig. 1, F versus EA (alternate dot and dash) lies above a diagonal between the "origin" and 1.00 F at Rest Length--1.00 EA. Radial lines represent constant force ratio on the "floor." New (and lower) curves of F versus EA could then be drawn for other muscle lengths, with new 1.00 F corresponding to any absolute value chosen from the F-L curve for a shorter muscle length.

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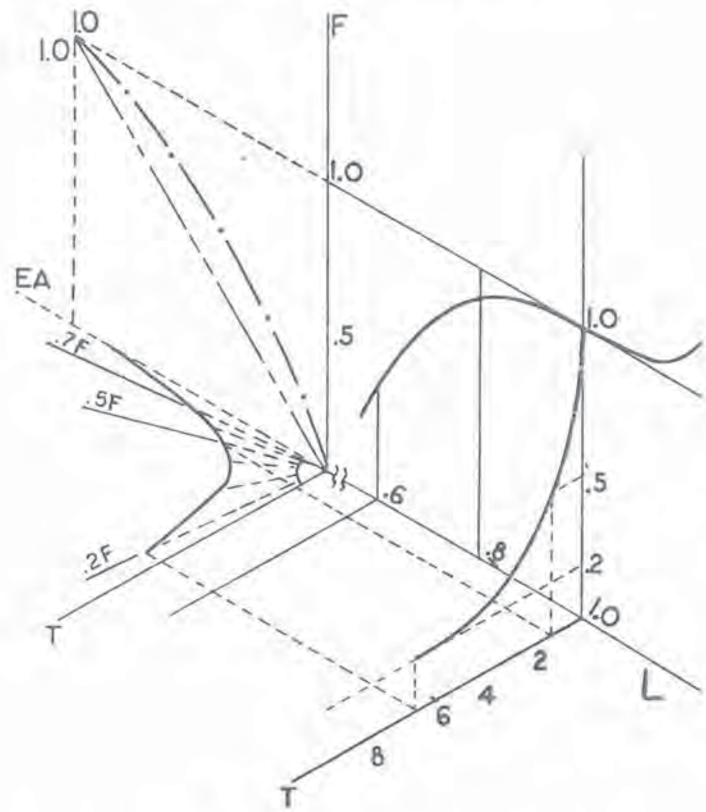


Fig. 1. Multidimensional correlations of force length, holding time, and integrated myoelectrical activity during voluntary isometric contraction.

# ELECTROMYOGRAPHIC EVALUATION OF MOPPING AND SWABBING IN CLEANING WORK

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## INTRODUCTION

Many Swedish investigators have among cleaners found high frequencies of disease and discomfort, originating from the locomotor system. In a questionaire answered by 279 cleaners, 40 per cent reported complaints from the shoulders, 33 per cent from the neck. The general work task that was considered to be the heaviest physically was floor cleaning. This is usually performed either by mopping or swabbing. In mopping, a moist short-threaded cloth is wiped in S-formed curves on the floor, similarly in swabbing, a long-threaded wet cloth is used. The cleaners experience a heavier physical load on the shoulders when swabbing compared to when mopping. The aim of the present study was to investigate whether swabbing gives a heavier muscular load on the shoulders and which shoulder muscles are under strain during the work procedure.

## EXPERIMENT

Six experienced (employed > 1 year) healthy female cleaners age 20 to 41 years were studied. A few days before the experiments the subjects were examined and found to have ordinary body size and normal muscle strength in shoulder elevation, shoulder abduction and power grip.

The experimental task consisted of continuous swabbing and mopping of a 16 $m^2$ area for one hour each. The subjects were to maintain their normal work rate for the sheduled hour. Between the mopping and the swabbing one hour of rest was given. The myoelectric activity was recorded on tape by bipolar surface electrodes from the right and left descending part of the trapezius muscle and at the left middle part of the trapezius muscle, and by bipolar wire electrodes from the right supraspinate muscle.

The experiments started with a series of test contractions for the investigated muscles in order to obtain the EMG-force relationship. This was done by a simultaneous recording of myoelectric activity and force during a slowly increased submaximal contraction for shoulder elevation, retraction and abduction.

## DATA ANALYSIS

For muscular load evaluation, vocational electromyography offers the possibility to estimate the contraction levels of a muscle. However, in an occupational situation the contraction levels are rapidly fluctuating and for ergonomic evaluation it is necessary to get a measure of the distribution of contraction levels over a certain period of time. By estimating the amplitude probability distribution function (APDF), such a measure is offered, exposing the static, the median, and the maximum contraction level for the time studied (Hagberg, 1979).

The myoelectric and force signals were determined as root-mean square (RMS) values by computer aided analysis from a tape recorder (Ericson and Hagberg, 1978). By power function regression analysis of the EMG levels versus force levels during the test contractions, the EMG-force relationship was established. Regression in reverse procedure of this relationship transformed the APDF of the EMG-signals during work to an APDF of contraction levels for the different muscles. The RMS versus time regression (exponential) was estimated for the first 15 minutes and the significance of the slope estimators (increase in RMS) was tested by the Student's t-test ($p < 0.05$).

## RESULTS

The distribution of load levels was evaluated for the first five minutes during swabbing and mopping to avoid influence of fatigue on myoelectric signal amplitudes. High static load was found for the right upper part of the trapezius muscle and for the right supraspinate muscle (figure 1) during both mopping and swabbing. Significant differences in static load between the two cleaning methods could only be found for the middle part of the trapezius muscle, although the static load levels for this muscle were low. However for the maximal contraction levels (figure 2) there was a significant difference between the two cleaning methods for three muscles. The maximum load level ratio (average for the six cleaners) swabbing/mopping was for the upper part of the right trapezius muscle: 1.30 (sign. $p < 0.10$), for the upper part of the left trapezius muscle: 1.49 (sign. $p < 0.05$), for the middle part of the trapezius muscle: 1.78 (sign. $p < 0.05$) and for the supraspinate muscle: 1.16 (non-sign.).

Significant increase ($p < 0.05$) in RMS values for the first 15 minutes of work occurred for the right and left upper part of the trapezius muscle in four respectively six work tasks, for the middle part of the left trapezius and supraspinate muscle in one respectively six work tasks. No difference in RMS-value change over time was found between the two cleaning methods for the first 15 minutes. The time for each 16 $m^2$ cleaning was 51 per cent longer in swabbing compared to mopping. The work performance velocity was approximately constant during the first 15 minutes of swabbing and mopping.

## DISCUSSION

For both mopping and swabbing the mean static load (mean for six cleaners) for the upper part of the right trapezius muscle and the right supraspinate muscle exceeds suggested limits (2-5 per cent MVC) for continuous long term contractions (Jonsson, 1978). This may be due to the equal stabilizing engagement by the right trapezius and supraspinate muscles in the shoulder. Preliminary clinical examinations of cleaners with cervico-brachial disorders show a dominance of complaints from the muscles concerned.

The higher maximum contraction levels found for the swabbing procedure were not high enough to promote local muscle fatigue to a higher extent compared to mopping, measured as an RMS-increase the first 15 minutes.

## CONCLUSIONS

Both mopping and swabbing should probably not be performed continuously for a long time without intermittent rests in order to avoid fatiguing processes in the shoulder muscles.

## ACKNOWLEDGMENT

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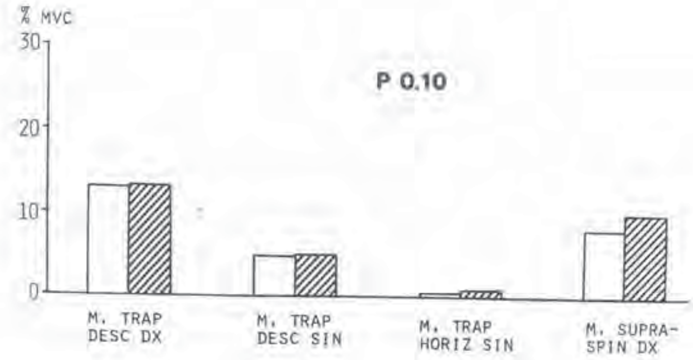


Fig 1 The static (p 0.10) load levels (mean for six subjects) for the fist five minutes for the muscles during mopping (blank area) and swabbing (shaded area)

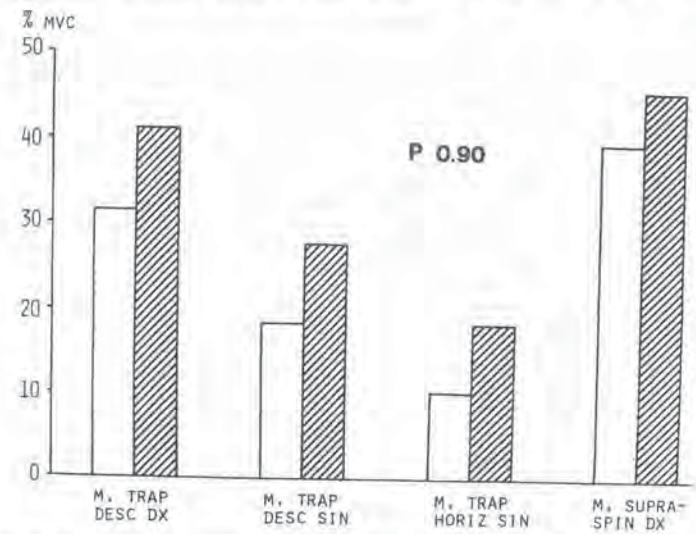


Fig 2 The maximum (p 0.90) load levels (mean for six subjects) for the first five minutes for the muscles during mopping (blank area) and swabbing (shaded area).

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