A NEW TECHNIQUE FOR FUNCTIONAL NEUROMUSCULAR STIMULATION

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SUMMARY

A technique was developed to allow proportional control of muscular contraction. It was shown that a drive stimulus placed proximally on a peripheral nerve delivering 60 pps at supramaximal levels in conjunction with a distal selective block stimulus at 600 pps at subthreshold levels provides means to regulate the contraction force. The induced graded contraction could be controlled over the range of 4 - 100% of the tetanic force by variations of the block stimulus pulse width or amplitude. Furthermore, the contraction was fused over the complete force range, linearly related to block pulse width and amplitude, and responding both statically and dynamically to such variations with an accuracy of ±2%.

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INTRODUCTION

Several functional electrical orthotic systems (FEOS) have been developed in recent years to compensate for loss of motor functions in stroke and spinal cord injured patients. These FEOS utilize electrical stimulation at suprathreshold levels and up to 60 pps applied at the afflicted nerve or muscle (Reswick, 1970; McNeal, 1973; Liberson, et al., 1961; University of Ljubljana, 1971; Committee on Prosthetics Research and Development, 1972).

Progress toward functional recovery for the disabled has been made through the various FEOS; however, limitation to the frequency range of 0 - 60 pps in the stimulation sources provided only ON-OFF control of the muscles. Proportional control was not achieved.

More recently, efforts were directed toward artificially producing proportional control of muscles (Peckham, 1970). In that approach, several electrodes are implanted in different areas of a muscle group and sequentially excited. Different levels of contractile force are produced, but the resultant effect is analogous to digital or discrete force production with continuous, smooth control still lacking.

The generally accepted explanation for the smoothly graded muscular control of normal intact persons is that a variable pulse rate in individual nerve axons occurs in combination with the recruitment of multiple axons innervating the fast and/or slow muscle groups (Mountcastle, 1974). Most of the FEOS cited above show that pulse rate control can be successfully induced artificially. What is lacking still is a method to artificially induce the recruitment process concomitantly with pulse rate control to obtain fine resolution of contractile force.

Tanner (1962) found that high frequency stimulation (20 kHz) applied to the middle portion of frog sciatic nerve would block stimulus impulse conduction through the nerve bundle in relation to the amplitude of the block stimulus. He suggested that the blocking effect occurred by local depolarization of fibers in relation to their size. Also, the larger alpha-fibers were attenuated prior to the smaller beta- and gamma-fibers which required the highest amplitude of the blocking stimulus. These findings were replicated by Woo and Campbell (1964) and Mendell and Wall (1964). Woo and Campbell also reported a more complex phenomena from their study of the block response of a small number of fiber units. McNeal (McNeal, et al., 1970), using rectangular pulses of 25 - 250 μs duration, reported that a block frequency of 600 pps at supramaximal levels produced optimal block. His findings were later applied clinically to alleviate spasticity and contracture in patients.

Based on the above findings, a scheme was devised to selectively apply electrical block in a peripheral nerve distal to a stimulus site. The objective: to artificially replicate the normal recruitment process and thereby demonstrate proportional control of contractile force.

PROPOSED CONTROL SCHEME

The scheme used to demonstrate the feasibility of proportional muscular contraction utilizing a pulse rate stimulus (hereafter called drive stimulus) in conjunction with a block stimulus is shown in Figure 1. The proximal drive stimulus is set in the range of 10 - 60 pps, and at a supramaximal level to elicit a fused level of contraction. The distal stimulus delivers the graded block signal. Variations in block pulse amplitude or pulse width are used to regulate the measured isometric force. An index of the control scheme's effectiveness is determined from the recorded force.

The hypothesis is that block pulse amplitude or width near supra threshold levels will permit only the smallest axons to conduct, thereby producing a low contractile force. Decreasing gradually either block pulse amplitude or width will allow larger fibers to conduct pulses and further increase the contractile force. Finally, at low or zero pulse amplitudes or widths, the largest axons conduct to produce the maximal force of contraction. Thus, one would predict that the ranges of block pulse amplitude or width variation lie between the threshold of the largest fibers (since these fire at lower levels) and the threshold of the smallest fibers (which require higher pulse amplitude and width levels to produce impulse activity). We expect this technique to provide a muscular control mechanism similar to the normal recruitment process.

METHODS

Six adult cats, anesthetized with a solution of chloralose and urethane (1% and 10% respectively), were used. This anesthetic agent was chosen as it does not appreciably affect the excitability of motor endplates, as may barbituate, steroid and some gaseous anesthetic agents. A lumbosacral laminectomy was performed, and the lumbar, hip and leg regions denervated except for the medial gastrocnemius or soleus muscle. The nerves to one of these muscles were dissected free over a distance of 2 to 4 cm, and the attempt made to free it from its sheath. At the heel, fascial extensions of the hamstring muscles and the plantar tendon were cut, and a fine steel cable with adjustable length was fixed into the severed tendon of the muscle understudy for later attachment to a strain gage. The animal was then firmly fixed

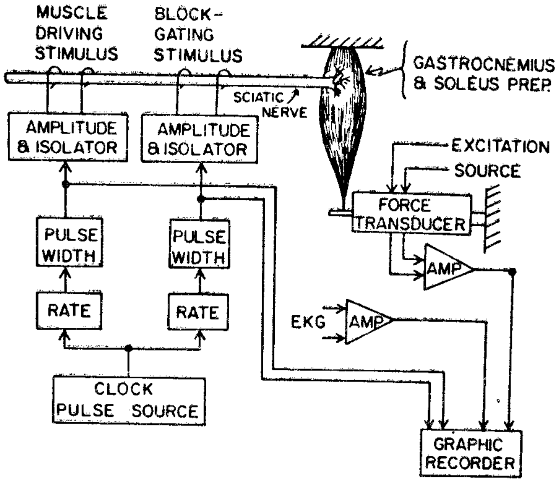


FIGURE 1 SCHEMATIC OF EXPERIMENT: FNS FEASIBILITY STUDY

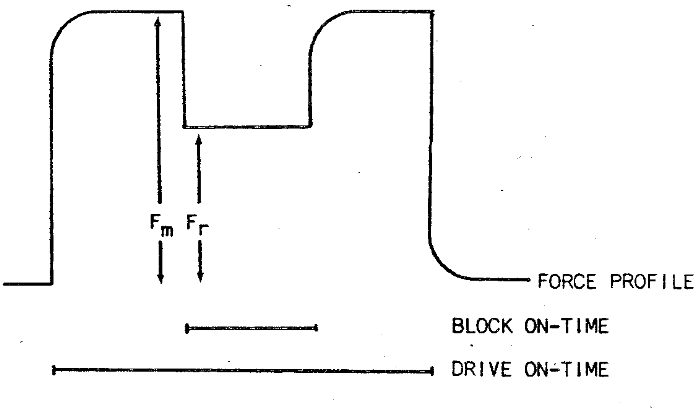
in a frame by: a pelvic support; a clamp on a spinous process; and pins through the femoral codyles, with the thigh and lower leg partially extended. Oil pools were formed at spinal, trochanteric and popliteal levels, which permitted placement of electrodes on the S_1 ventral root, the sciatic nerve at the trochanteric or mid-femoral level, and on the muscle nerve. Usually, bipolar hook Ag-AgCl electrodes of 5 mm interpolar distance were used, though wider spacings, tripolar electrodes and wrap around electrodes were also used. In early experiments, the stimuli were delivered by Grass model S4 and S8 stimulators, using capacity coupling at the Grass isolation units. In subsequent experiments, a Fired Haer stimulus generation system and Tektronix Type 2620 stimulus isolators were used. Rest periods were experimentally selected as needed to prevent evidence of fatigue on a subsequent trial. Muscle tension registered either by a Statham strain gage or Grass FT-10 force transducer was recorded on a polygraph. Temperature was monitored by a needle thermode inserted into the anterior compartment of the lower leg and the popliteal oil pool kept at about 37° C by radiant heat. Body temperature was separately maintained.

For the work reported here, the drive stimulus site was always proximally situated on the sciatic nerve. The blocking stimulus was applied at a nerve site interposed between the muscle under study and the drive stimulus site. Site spacings varied from about 5 cm maximum down to 1 cm, with typical spacings being 1 to 2 cm.

PROTOCOL

Stimulation parameters were as follows. Drive stimulus: monopolar rectangular pulses of supramaximal amplitude (usually 100 µs pulse duration and a 60 pps rate). Block stimulus: monopolar rectangular pulses at 600 pps, with variable pulse amplitude and width in the ranges of 0.1 - 3 mA and 10 - 700 µs, respectively.

The stimulation sequence is similar to the one utilized by McNeal (McNeal, et al., 1970), with the application of the drive stimulus first. After the muscular force reached its peak (about 2 seconds), the block stimulus was applied for 2 seconds and then removed. After the tetanic force recovered, about 2 seconds later, the drive stimulus was removed. Figure 2 shows this sequence in terms of the corresponding force response. An interstimulus interval of at least 60 seconds was allowed to prevent muscle fatigue.



$$BE = \frac{F_m - F_r}{F_m} \times 100\%$$

FIGURE 2 SCHEMATIC OF FORCE RESPONSE TO SEQUENCE OF DRIVE BLOCK STIMULI APPLIED TO SCIATIC NERVE

Four sets of measurements were made: force response as a function of block pulse amplitude for constant block pulse widths; force response as a function of block pulse width for constant block pulse amplitudes; force response for repetitive application of the same block stimulus, and force response to dynamic changes in block pulse amplitude or pulse width.

Variation of contractile force as a function of block pulse amplitude was recorded for constant pulse widths. The 600 pps block stimulus was first set at a pulse width of 10 µs and the stimulus-rest-stimulus sequence conducted while step-wise increasing the pulse amplitude from sub-threshold levels up to the level of block threshold (defined by a detectable reduction of force). Thereafter, the pulse amplitude was increased at increments of 0.02 mA and a recording of force response obtained for each amplitude level until no further block was evident. This procedure was repeated for block pulse widths ranging from 10 to 700 µs.

Variations of contractile force as a function of block pulse width were recorded for constant values of pulse amplitudes. A low level of block pulse amplitude (about 0.2 mA) was set, and the pulse width increased from 10 µs until block threshold was reached. Force response records were then taken for several pulse width increments up to the point where a pulse width increase produced no further reduction in force. This procedure was repeated for block pulse amplitudes ranging from 0.2 to 0.9 mA.

Force response to repeated application of a given block pulse width and amplitude setting was tested. At the end of each set of force versus pulse amplitude recordings at a constant pulse width, the amplitude corresponding to 50% block was reset and three stimulation sequences were repeated. The same procedure was conducted for force versus pulse width measurements for each pulse amplitude value.

Although detailed procedures to identify the dynamic response capabilities of the postulated muscular control scheme are beyond the scope of this study, it was deemed desirable to demonstrate its possibility. During a few measurement sessions, the block pulse amplitude and pulse width were adjusted to produce a maximal block (reduction of force). The drive stimulus was applied and maintained for an extended period of time. Two to three seconds after drive application, the block stimulus was applied. Then the pulse amplitude was manually decreased at the approximate rate of 0.1 mA/sec (until maximal recovery of the tetanic force) followed by an increase in pulse amplitude at the same rate to the point of maximal block. More rapid variations of pulse amplitude were also made. A similar procedure was used to explore dynamic variations in the block pulse width.

DATA

A typical set of stimulation sequences showing the effect on isometric force of various block pulse amplitudes is presented in Figure 3. The recordings of force response repeatability are shown at the right. Similarly, a typical set of force responses to various block pulse widths is shown in Figure 4. Again, the force response repeatability tests are shown at the right of the figure. Examples of the force response to dynamic changes of block pulse amplitude and block pulse width are presented in Figures 5 and 6, respectively. The initial three seconds demonstrates the force response to the supramaximal drive stimulus alone. Subsequent changes in the response follow application of the block stimulus with manual changes in the variable as described in the protocol.

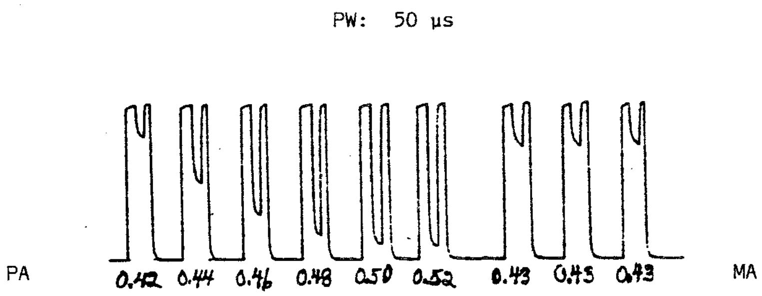


FIGURE 3 FORCE RESPONSE FOR DIFFERENT BLOCK PULSE AMPLITUDES AT A CONSTANT BLOCK PULSE WIDTH OF 50 µS

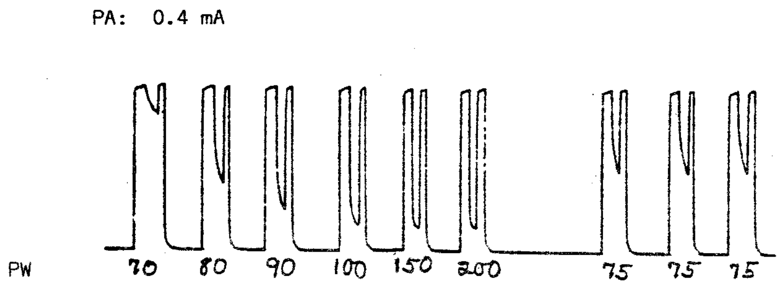


FIGURE 4 FORCE RESPONSE FOR DIFFERENT BLOCK PULSE WIDTHS AT A CONSTANT BLOCK PULSE AMPLITUDE OF 0.4 mA

DRIVE STIMULUS: 60 PPS
1 mA
100 µs

BLOCK STIMULUS: 600 PPS
VARIABLE (0.2 - 0.8 mA)
100 µs

FIGURE 5 FORCE RESPONSE TO DYNAMIC VARIATION OF BLOCK PULSE AMPLITUDE

DRIVE STIMULUS: 60 PPS
1 mA
100 µs

BLOCK STIMULUS: 600 PPS
0.4 mA
VARIABLE (130 - 230 µs)

FIGURE 6 FORCE REPONSE TO DYNAMIC VARIATION OF BLOCK PULSE WIDTH

RESULTS

To quantify the data, a modified version of the blocking effectiveness (BE) utilized by McNeal (McNeal, et al., 1970) is employed. The BE is defined by

$$BE = \frac{F_m - F_r}{F_m} \text{ X } 100\%$$

where F_m - maximal tetanic force and F_r is defined here as the residual force at the onset of block (see Figure 2). BE = 100% indicates complete block with no residual force and BE = 0% corresponds to no block or tetanic force.

Blocking Effectiveness as a Function of Block Pulse Amplitude

Utilizing the BE index, the typical stimulation sequences for block pulse amplitude variations (Figure 3) are represented in Figure 7. BE is plotted versus block pulse amplitude for constant pulse width values. The curve consists of two linear portions. For BE ranging from 0% to about 80%, the BE is linearly related to relatively small changes in amplitude for each of the fixed pulse widths. For a BE greater than about 85%, a large change in pulse amplitude is required to achieve a slightly higher BE. In this block saturation range, however, the BE is still linear with pulse amplitude changes. Total block (BE = 100%) was not achieved at stimulus values believed to be tolerated by the nerve, that is, without apparent nerve damage. However, BE values of 90 - 95% are obtainable. Stated another way, the isometric force can be controlled from 5% to 100% of the tetanic force by block pulse amplitude variation for given pulse widths.

Another aspect of these results is that a pulse width in the 100 to 700 µs range requires the least pulse amplitude for given levels of BE. Pulse widths shorter than that range require higher pulse amplitudes to achieve the same level of BE. Of the two pulse width extremes, the shorter pulse widths place higher demands on pulse amplitude for a given BE than do the longer pulse widths. Also, the BE saturation levels for shorter pulse widths is slightly less than the BE saturation levels for the longer pulse widths.

An additional observation from these results may be made. Examine the slopes of the curves in the BE range at 85% or greater. The incremental change in BE per increment change of block pulse amplitude is virtually constant for block pulse widths greater than 100 µs. For pulse widths less than 100 µs, however, the increment of pulse amplitude required to produce the same incremental change in BE continues to increase as the pulse width beomces narrower. This factor has important implications on the instrumentation to accomplish dynamic control of

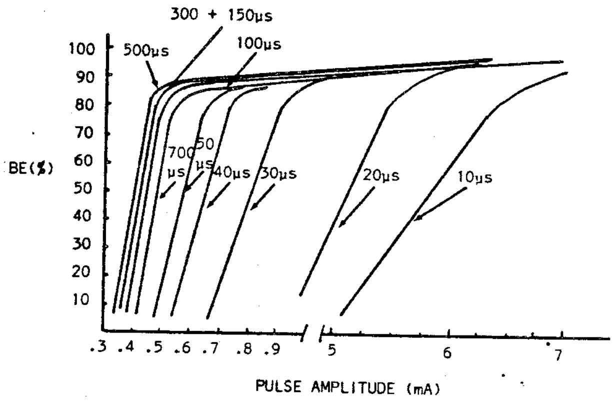


FIGURE 7 EFFECT OF BLOCK PULSE AMPLITUDE ON FORCE REDUCTION FOR CONSTANT VALUES OF BLOCK PULSE WIDTH

muscle force. For example, dynamic control of BE through pulse amplitude variation is easier to instrument if the pulse widths are less than 100 μs.

In summary, the 100 μs block pulse width separates the BE response versus block pulse amplitude into two categories: BE versus pulse amplitude for pulse widths larger than 100 μs and shorter than 100 μs The family of curves for widths larger than 100 μs are parallel, consist of two piecewise linear segments, possess a constant slope and reach saturation at near 95% BE. Pulse widths less than 100 μs, however, have varying slopes, require larger pulse amplitudes and the BE saturation level is slightly lower for the shorter pulse widths. The results demonstrate that block pulse amplitude variation is a feasible parameter for proportional control of isometric force. A proper pulse amplitude dynamic range can be chosen to proportionally control the isometric force over a 5 - 100% range.

Blocking Effectiveness as a Function of Block Pulse Width

The typical stimulation sequence for constant pulse amplitudes of Figure 4 are reduced to BE as a function of block pulse width in Figure 8.

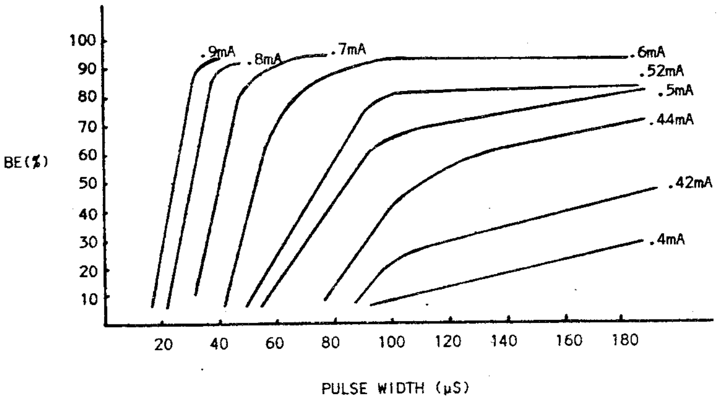


FIGURE 8 EFFECT OF BLOCK PULSE WIDTH ON FORCE REDUCTION FOR CONSTANT VALUES OF BLOCK PULSE AMPLITUDE

The most striking feature observed is the varying slopes of the family of constant amplitude curves. This feature is attribted to the large pulse widths necessary to excite the small fibers for a given pulse amplitude as compared to that required for larger fibers.

The family of curves for BE versus block pulse width at various constant values of block pulse amplitude is divided into two areas for discussion: pulse amplitudes greater than 0.5 mA regardless of pulse width, and pulse widths equal to or greater than 100 µs for constant pulse amplitudes less than 0.5 mA. For block pulse widths less than 100 µs at constant block pulse amplitudes greater than 0.5 mA, the curves consist of two piecewise linear segments. The steep-slope portion indicates that large changes in BE can be achieved with only small changes in pulse width. This group of segments includes a range of BE from about 4% up to 65% for the 0.5 mA pulse amplitude, or up to about 95% for a 1 mA pulse amplitude. The second group of piecewise linear segments is interpreted to be a 'saturation' level. That is, for block pulse amplitudes greater than about 0.5 mA, there are regions where a large increase in block pulse width produces only a slight increase in BE. These two segments are connected by a nonlinear "knee."

For block pulse widths greater than about 100 µs and block pulse amplitudes less than 0.5 mA, the nonlinear "knee" portion of the family of curves disappears. The saturation portion of the curves passes through a nonlinear transition toward a condition of less saturation. That is, at 0.4 mA pulse amplitude, pulse widths greater than 100 µs are linearly related to BE but limited to BE values less than 50%. In this low pulse amplitude region, a large incremental change in block pulse width will make a modest change in BE at a low level of BE. A similar change in pulse width for a pulse amplitude of 0.5 mA produces little change in BE which is already at a high level.

It is again observed that complete block was not obtained. BE values near 95%, however, are readily available as long as block pulse amplitude is above a critical level. The results demonstrate that graded proportional control over the contractile force can be obtained by varying the block pulse width. A block stimulus using large pulse amplitudes results in a short pulse width dynamic range. Smaller pulse amplitudes provide a larger, more practical, dynamic range of 100 to 300 µs pulse widths for 4 - 95% BE. Pulse width control of contractile force appears feasible.

Repeatability

BE accuracy and deviation range for three sequential runs at the same block pulse amplitude and width were analyzed. Table 1 summarizes the average deviation and deviation range for sequential runs.

TABLE 1

Average Deviation (%)		Average Deviation Range (%)	
PA	PW	PA	PW
.29%	1.74%	.29%	4.29%

Note:
Largest deviation range was 8.7%, and smallest was 0%.

The accuracy criterion of obtaining a prescribed force indicates block pulse amplitude control to be superior to block pulse width control. Furthermore, the proposed force control scheme is capable of providing a repeatable contractile force for a prescribed block pulse amplitude and width at an accuracy better than 2%.

CONCLUSIONS

Stimulation of a peripheral nerve by a pulse rate stimulus of 60 pps and supramaximal threshold in conjunction with a distal block stimulus of varying pulse amplitude or width allows proportional control over the contractile force. It is believed that the artificial signals induce some form of the normal neuromuscular recruitment process. Preliminary assertions regarding the response of muscular contraction to drive-block stimulation are:

1. Proportional control of artificially induced muscle force is accomplished by:
    a. Variations of block pulse width
    b. Variations of block pulse amplitude.
2. The contractile force is controlled over the range 4 - 100% of the tetanic force.
3. The force generated is always fused, even at 4% of the tetanic force.
4. The force varies in a linear fashion with block pulse width or pulse amplitude.
5. The range of block pulse amplitude control for 4 - 100% force variation is the shortest at pulse widths greater than 100 µs, but largest for pulse widths less than 100 µs.
6. The 4 - 100% range of block pulse width control of regulated force is least for pulse amplitudes near suprathreshold levels and greatest near subthreshold levels.
7. The maximal saturation level that is reached by block pulse width variation is dependent on the constant pulse amplitude applied.
8. The muscular contractile force is controlled in either dynamic or static fashion by block pulse width or pulse amplitude.
8. Short-term repeatability of a prescribed block level is better than ±2%.

The described technique can be applied toward the rehabilitation of stroke and spinal cord patients suffering from lack of motor control of peripheral muscles. Further studies are necessary in order to determine and optimize this technique for implant purposes.

BIBLIOGRAPHY

Casey, K. L. and M. Blick, "Observation on Anodal Polarization of Cutaneous Nerves," Brain Research 13:155-167, 1969

Committee on Prosthetics Research and Development, National Academy of Science, Evaluation of the Ljubljuna Functional Electrical Ankle-Foot Orthosis, National Research Council, 1972

Crago, P. E., P. H. Peckham, and J. P. Van der Meulen, "The Choice of Pulse Duration for Chronic Electrical Stimulation Via Surface Nerve and Intramuscular Electrodes," Annals of Biomedical Engineering 2:252-264, 1974

Crago, P. E., J. P. Van der Meulen, and J. Reswick, "A Pilot Study of the Effect of Functional Electrical Stimulation on the Recovery of Function Following Stroke in Men," in Wulfsohn and Sances (Eds.), The Nervous System and Electric Currents, Plenum Press, New York, 1970

Crochetiere, W., "On the Use of Electrically Stimulated Muscle as a Controlled Activator of a Limb," Ph.D. Thesis, Case Western Reserve University, June, 1967

Friedman, H., "Energy Consideration in the Design of Implantable Electrodes for Peripheral Nerve Stimulation," in M. Gavrilovic and A. B. Wilson (Eds.), Advances in the External Control of Human Extremities, pp. 603-614, Belgrade, Yugoslavia, 1975

Gracanin, F. and L. Marincek, "Development of New Systems for Functional Electrical Stimulation," in M. Gavrilovic and A. B. Wilson (Eds.), Advances in the External Control of Human Extremities, pp. 529-657, Belgrade, Yugoslavia, 1973

Gracanin, F., A. Vavken, A. Jeglic, and M. Benedik, "An Outline of Principles and Methods Utilized in Evaluating Implanted Systems for Functional Electrical Stimulation," in M. Gavrilovic and A. B. Wilson (Eds.), Advances in External Control of Human Extremities, pp. 529-657, Belgrade, Yugoslavia, 1975

Gugic, P., U. Stanic and A. Trnkoczy, "Hybrid Control of the Ankle Joint," in M. Gavrilovic and A. B. Wilson (Eds.), Advances in External Control of Human Extremities, pp. 617-630, Belgrade, Yugoslavia, 1975

Hambrecht, F. and J. Reswick (Eds.), Functional Electrical Stimulation, M. Dekker, Inc., New York, 1977

Kralj, A., S. Grobelink, and L. Vodovnik, "Functional Electrical Stimulation of Paraplegic Patients," in M. Gavrilovic and A. B. Wilson (Eds.), Advances in External Control of Human Extremities, pp. 561-574 Belgrade, Yugoslavia, 1975

Liberson, W. T., H. J. Holmguest, D. Scott, and M. Dow, "Electrotherapy: Stimulation of the Peroneal Nerve Synchronized with the Serving Phase of the Gait of Paraplegics," Archives of Physical Medicine 42: 101-105, 1961

Magdaleno, R., D. McRuer, and G. P. Moore, "Small Perturbation Dynamics of the Neuromuscular System in Tracking Tasks," NASA Report CR-121, 1968

McNeal, D. R., "Annual Report of Progress, 1973," Rehabilitation Engineering Center at Rancho Los Amigos Hospital, pp. 1-22, Downey, California, 1973

McNeal, D. R., B. R. Bowman and W. L. Momsen, "Peripheral Block of Motor Activity," in M. Gavrilovic and A. B. Wilson (Eds.), Advances in External Control of Human Extremities, pp. 473-519, Belgrade, Yugoslavia, 1970

Mendell, L. B. and P. H. Wall, "Presynaptic Hyperpolarization: A Role for Fine Afferent Fibers," Journal of Physiology 172:274-294, 1964

Mortimer, J. and P. H. Peckham, "Electrical Excitation of Paralyzed Muscle," in M. Gavrilovic and A. B. Wilson (Eds.), Advances in External Control of Human Extremities, pp. 575-583, Belgrade, Yugoslavia, 1973

Mountcastle, V. B., Medical Physiology, 13th Edition, Volume 1, C. V. Mosby Company, St. Louis, Missouri, 1974

Paeniezek, R., J. Kiwereki, J. Wirski, and H. Borowski, "Some Problems of Implant Stimulation Applied to Grasp Movements," in M. Gavrilovic and A. B. Wilson (Eds.), Advances in External Control of Human Extremities, pp. 568-602, Belgrade, Yugoslavia, 1975

Peckham, P. H., "Design Consdierations in Electrical Stimulation of Skeletal Muscle," Case Western University EDC Report, EDC 4-28-23, 1968

Peckham, P. H., B. A. Romich and J. B. Reswick, "Sequential Electrical Stimulation of Skeletal Muscle," 8th International Conference on Medical and Biological Engineering, Chicago, Illinois, July 20-25, 1969; also in Wulfohn and Sances (Eds.), The Nervous System and Electric Currents, Plenum Press, New York, 1970

Ranck, J. B., "Which Elements are Excited in Electrical Stimulation of Mammalian Central Nervous System?" Brain Research 98:417-440, 1975

Reswick, J. B., "Final Report, Biomedical Research Program on Cybernetics Systems for the Disabled," EDC Report No. 4-70-29, SRS Grant No. RD-1814-M, Case Western Reserve University, Cleveland, Ohio, 1970

Schuck, E., H. Friedman, W. Wilemon, and D. McNeal, "Developing Clinical Devices for Hemiplegic Stroke Patients," in M. Gavrilovic and A. B. Wilson (Eds.), Advances in External Control of Human Extremities, pp. 535-546, Belgrade, Yugoslavia, 1975

Stein, R. B., "Peripheral Control of Movement," Physiological Review 54:215-143, 1974

Tanner, J. A., "Reversible Blocking of Nerve Conduction by Alternating Current Excitation," Nature 195:712-713, 1962

University of Ljubljana Faculty for Electrical Engineering, Development of Orthotic Systems Using Functional Electrical Stimulation and Myoelectric Control, Final Report, Ljubljana, Yugoslavia, December, 1971

Woo, N. Y. and B. Campbell, "Asynchronous Firing and Block of Peripheral Nerve Conduction by 20 kc Alternating Current," Bulletin of the Los Angeles Neurological Society 29 (2):87-94, 1964