The Effect of Stimulus Parameters on the Recruitment Characteristics of Direct Nerve Stimulation

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Abstract—The effect of stimulus parameters on the recruitment characteristics of motor nerve was studied for regulated current monophasic and balanced charge biphasic stimuli. Results of a nerve model investigation indicated that the threshold difference between different diameter nerve fibers would be dependent on pulse width, the choice between monophasic and biphasic stimuli, and the delay between the primary cathodic and secondary anodic pulses. Threshold difference increased with decreasing pulse width, the greatest effects evident for pulses less than 100 μs. Biphasic stimulation with no delay between pulses provided greater threshold separation than monophasic stimulation or biphasic stimulation with delay. Animal experiments, in which recruitment in a nerve trunk composed of mixed diameter nerve fibers was examined, showed a decrease in recruitment slope with a decrease in pulse width and with the use of a biphasic, zero delay pulse. These results were examined through muscle force measurements using both a metal loop electrode encircling the nerve trunk and a nerve cuff electrode, i.e., a loop electrode in an insulating tube.

These results have applications in physiological systems in which modulated end organ response through direct nerve stimulation is desired, e.g., motor and auditory prostheses.

Introduction

Electrical stimulation of paralyzed muscle is a means to restore upper limb motor function in the quadriplegic patient [17]–[19]. Systems using this technique have employed coiled wire intramuscular electrodes [3] to activate skeletal muscle. Intramuscular (IM) electrodes have proven to be an effective tool in research, but present two major problems for clinical systems. First, IM electrodes are subject to repeated bending that can eventually result in electrode breakage. Second, Crago *et al.* [7] have shown that recruitment is dependent on muscle length with intramuscular stimulation, making control difficult. Direct nerve stimulation through a nerve cuff electrode is an alternative to IM stimulation and would not exhibit length dependent recruitment, nor would the electrode or its leads be subjected to the high degree of repeated bending experienced by the IM electrode.

A difficulty encountered with direct nerve stimulation through an electrode cuff is that the difference in current level between threshold excitation and maximal recruitment is not large. This type of stimulus versus response (recruitment) characteristic is not desirable in systems where gradation or modulation of the evoked response is needed, as in motor function restoration. A more gradual recruitment of nerve fibers during nerve trunk stimulation can be obtained if it is possible to increase the difference in threshold between different diameter nerve fibers.

In this paper, we report on the examination of stimulus parameters that effect an increase in the threshold difference between different diameter nerve fibers and, therefore, decrease the slope of the recruitment curve produced by direct nerve stimulation. Specifically, variations in pulse width and the use of a secondary stimulus pulse have been examined. Vodovnik *et al.* [21] have shown that pain sensation diminishes when narrow stimulus pulses are used with surface electrodes to produce muscle contraction. Since sensory afferent fibers that convey pain sensation are smaller in diameter than motor fibers [1], the reduction in pain sensation reported could be due to the selective excitation of larger nerve fibers without concomitant small fiber excitation. van den Honert and Mortimer [20] have found that the presence of the secondary anodic phase of the biphasic pulse could be used to abolish excitation initiated by the primary cathodic pulse. The ability of the anodic pulse to disrupt the regenerative processes in the nerve fiber may have a fiber diameter dependence, and may also be useful in separating the threshold levels of different diameter nerve fibers.

Studies performed on McNeal's simulated nerve model [14] are presented first. Results in animals with a metal loop electrode encircling the nerve trunk and a nerve cuff electrode are then reported as an extension of the model predictions.

Nerve Model

Analytical models have been developed to examine excitation phenomena in nervous tissue [4], [9], [12]–[14]. For the purposes of this study, the McNeal model [14] was used with minor changes (a predictor corrector method was used as the numerical integration procedure with maximum allowable error of 0.005 per iteration). The model, based on the Frankenhauser-Huxley equations [10] describing excitation of the myelinated nerve fiber, permits simulation of excitation by a monopolar point source electrode at a finite distance from a variable diameter nerve fiber. Extracellular electrical potentials along the fiber are the result of current flow in an infinite isotropic external medium.

Manuscript received January 8, 1982; revised November 23, 1982. This work was supported by the NIH-NINCDS Neural Prosthesis Program under Contract N01-NS-0-2330 and the NIH Systems Integrated Biology Training under Grant HL075-35.

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0018-9294/83/0700-0407$01.00 © 1983 IEEE

The effects of short duration monophasic and biphasic stimulation on threshold levels for excitation of various diameter fibers were examined. The previous finding of McNeal indicated that current thresholds varied inversely with fiber diameter for a given pulse width. By examining the differences in threshold for different diameter nerve fibers, it would be possible to estimate the stimulus parameters that would provide the greatest excitability differences between different diameter nerve fibers. It would then be possible to provide a more gradual recruitment of nerve fibers with increasing stimulus intensity. The stimulus parameters examined were pulse width, monophasic versus biphasic, and delay between the two phases of the biphasic pulse.

In the first study, the effects of pulse width and delay time between cathodic and anodic pulses were examined. The stimulus waveforms and variables used are shown in the top two traces of Fig. 1.

Fibers of 10 and 20 μm external myelin diameter (axon diameter being 70 percent of total diameter [12]) were placed 1 mm from a monopolar stimulating electrode. These diameter values were chosen to represent the ends of a distribution of motor nerve diameters [1]. Thresholds were determined for each fiber, and the difference in threshold between the 10 and 20 μm fiber (I_{10}-I_{20}) was calculated for a series of symmetric rectangular biphasic pulses. Pulse widths of 25, 50, 100, and 200 μs and delay times of 0, 20, 40, and 80 μs were used. Results plotted in Fig. 2 show threshold difference versus delay time for a family of pulse widths. It is evident that the use of narrow pulses and the elimination of delay between cathodic and anodic pulses provide the best enhancement of threshold separation. Similar results were obtained with fibers placed at 2 mm away from the stimulating electrode.

In the second study, biphasic stimulation was compared directly with monophasic stimulation. Fibers with total diameters of 10, 12, 14, 16, and 20 μm were placed 1 mm from the stimulating electrode. The pulse width was held constant at 25 μs, and the delay time for the biphasic pulse was zero. These values were chosen on the basis of the results of the first study to yield large threshold differences. A graph of threshold current versus fiber diameter is given in Fig. 3 (solid lines). The slope of the biphasic curve is greater than that of the monophasic curve, indicating a greater difference in threshold between different diameter nerve fibers.

In an analytical model, threshold levels remain stable over time. This may not be the case in a living system because of fluctuations in membrane properties, perturbations in the medium between electrode and nerve, and electrode movement. Computer simulation of these effects was performed by moving the stimulating electrode 5 percent in both directions from the original position 1 mm away from the nerve fiber. This was the most direct method of simulating the physiological fluctuations since there was no facility in the model for manipulation of the electrical potentials along the fiber. The dashed lines in Fig. 3 show the threshold ranges created by this procedure. If a stimulation current equal to just threshold level for a 16 μm fiber is considered, then comparison can be made between the range of fibers that would be recruited by the monophasic and biphasic waveforms. The dotted lines in the figure indicate that the biphasic stimulation would recruit a narrower range of fibers by approximately 33 percent, and therefore be more selective than monophasic stimulation.

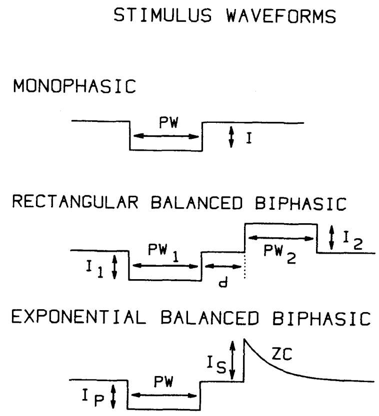

Fig. 1. Stimulus waveforms used in this study. Monophasic and rectangular wave biphasic stimuli were used in the model work. All three stimuli were used in the experimental work. Negative pulses are cathodic, positive pulses are anodic. For rectangular wave biphasic, $PW_1 = PW_2$ and $I_1 = I_2$, producing the balanced waveform. For exponential wave biphasic, charge injection in the primary and secondary pulse are equivalent, but I_p and I_s are not necessarily equal. Z = electrode preparation impedance, C = discharge capacitor value.

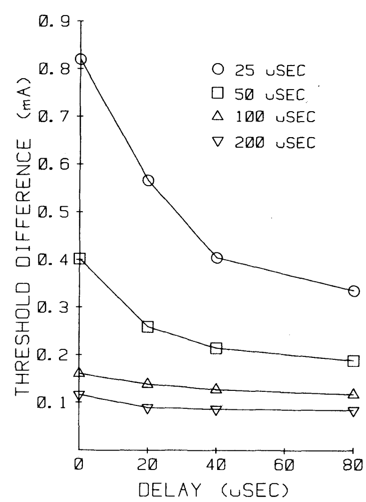

Fig. 2. Nerve model determination of the difference in threshold between 10 and 20 μm fibers (I_{10}-I_{20}) for various pulse widths and delay times. All stimuli are rectangular biphasic pulses and the minimum separation between nerve and electrode is 1 mm.

The results of the simulation predict that narrow biphasic stimuli without delay between cathodic and anodic pulses will yield the largest threshold differences between different diameter nerve fibers. Furthermore, by going to higher stimulus amplitudes, either by decreasing pulse width or by using biphasic stimuli, the recruitment of nerve fibers will become less vulnerable to system variability, thereby allowing for a more gradual recruitment of fibers during nerve stimulation.

Animal Experiments

Insight provided by the model was used to investigate the recruitment characteristics of whole nerve excitation in animals. Stimulation through both a metal loop electrode and a nerve cuff electrode (a loop with an insulating cuff) was examined. The loop electrode preparation provided an exper-

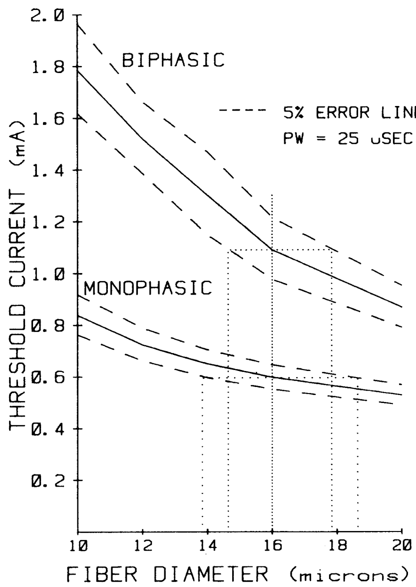

Fig. 3. Nerve model comparison between biphasic and monophasic stimulation of the same pulse width. The 5 percent error lines indicate the simulated effect of physiological variation on the recruitment properties. (See text for details.)

imental progression between the monopolar electrode isotropic field characteristics, seen previously in the nerve model study, and the nerve cuff electrode, a more appropriate scheme for an implant application.

15 adult cats (2.5-5.2 kg) have been studied. The animals were anesthetized initially with intramuscular injection of ketamine hydrochloride (25 mg/kg) accompanied by atropine sulfate (0.03 mg/kg) to reduce salivation. The cephalic vein was cannulated, and anaesthesia was maintained with sodium pentobarbital (IV 10 mg bolus injections as required). The cats were intubated and maintained on a respirator (Harvard Apparatus Company) throughout the data collection procedure at rates and volumes indicated by the manufacturer to be appropriate for their respective body weights.

Surgical isolation of the medial gastrocnemius muscle and approximately 2.5 cm of its innervating nerve branch (diameter range 0.8-1.2 mm) was approached through incision into the popliteal fossa, care being taken not to interfere with their blood supplies. The medial gastrocnemius (MG) and its nerve was chosen for this study because of their distribution of muscle fiber types and nerve fiber diameters, respectively [8], and because of the relative ease of their isolation. Markers were placed in the tibia and in the Achilles tendon to indicate maximal physiological length with approximately 45° of knee flexion.

When metal loop electrode stimulation was examined, a stainless steel loop was placed and closed around the nerve approximately 1 cm away from its point of entry into the muscle. The electrode formed a ring centered on the nerve trunk with a diameter approximately twice that of the trunk. The trunk was supported by glass hooks placed on both sides of the electrode. A 37°C saline pool was formed in the leg to bathe the nerve and muscle.

For nerve cuff stimulation, the MG nerve was cut proximally leaving approximately a 2.5 cm length of nerve. The nerve was then threaded through the 1 cm long silicone rubber nerve cuff electrode. In the initial experiments, stainless steel multistranded wire was used as the electrode in the cuff. Later on, electrodes were made from 26 gauge metal tubing. In both cases the electrode was centered between the ends of the silicone rubber cuff. Cuffs with bore sizes of 1.0 mm and 1.4 mm were used to accommodate nerves from different preparations without compressing them. A 5 cm hypodermic needle (22 gauge) placed subcutaneously in the animal's back was used as a stimulation reference electrode. The leg was immobilized in a frame by clamps holding the ankle and the bony protuberances of the knee. The MG was fixed at the measured maximal physiological length.

The contractile response of the muscle was evaluated for both single pulses and 30 Hz, 1/2 s pulse trains. With the pulse waveform parameters set at chosen values, the stimulus amplitude was decreased in steps from the level that gave the maximal force response (100 percent recruitment) to that which gave a threshold response. Step size was chosen small enough to obtain a relatively smooth recruitment curve throughout its range. The muscle was potentiated (i.e., twitched repeatedly at maximal stimulus level) before each run through the stimulus range to ensure a stable force output. When single pulses were used, five separate twitches were performed on the muscle at 1 s intervals for each stimulus intensity. The ensemble average of these twitches was then calculated and used as the average twitch response.

Data collection and stimulator gating of the animal experiments were controlled by a software routine operating on a PDP 11/23 microcomputer. A battery powered regulated current stimulator developed in our laboratory, and a Nuclear-Chicago transformer isolated line powered stimulator were used. Choice of stimulator depended on the secondary pulse waveform desired (see next section). Muscle force was measured by a rigid strain gage force transducer attached to the Achilles tendon. The output signal from the transducer was amplified by a Tektronix 3C66 carrier amplifier and digitally converted by a successive approximation A/D converter. A sampling frequency of 1 kHz was used for data collection.

Stimulus Waveforms

Monophasic and biphasic waveforms were both examined. Monophasic pulses were rectangular with widths ranging from 10 to 300 μs. The nerve electrode served as the cathode, while the reference electrode was the anode. Biphasic pulses were of two types: rectangular wave (using the Nuclear-Chicago stimulator) and exponential decay [16] (using the battery powered stimulator developed in our laboratory). In both cases the amount of charge injected in the secondary anodic pulse was equal in magnitude but opposite in sign to the amount of charge injected in the primary cathodic pulse. No special significance is attributed to the exponentially decaying pulse, except that balanced charge can be achieved with relative ease, and that the relative magnitudes of the current levels in the primary and secondary pulse can be manipulated by varying the series capacitor.

The parameters that varied during this study were pulse width (10-300 μs), delay between cathodic and anodic pulses (0-100 μs), and the secondary pulse discharge time constant, dependent on the electrode load impedance and the value of

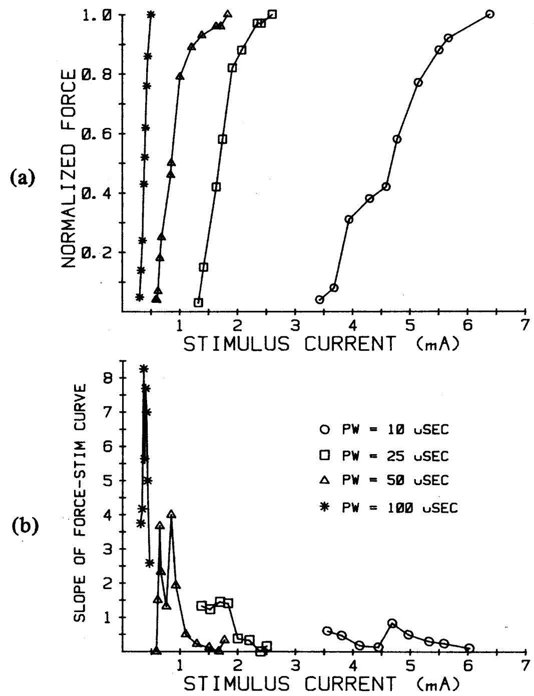

Fig. 4. Effect of pulse width on recruitment curve. Biphasic, zero delay rectangular pulse stimulation through a loop electrode in a saline bath. (a) Actual recruitment curves. Force normalized to 8 N in all cases. (b) Slope of the recruitment curves along their length (in units of normalized force/mA).

the series discharge capacitor. Graphs of the waveforms examined with indications of the parameters evaluated are shown in Fig. 1.

Secondary Pulse Peak Current Measurements

When capacitive discharge biphasic current pulse were used, the magnitude of charge injected in the primary and secondary pulse was equivalent. The peak of the secondary pulse was varied by changing the value of the series coupling capacitor. Peak current in the secondary pulse was measured directly.

Biphasic stimulus pulses of various primary pulse widths and discharge time constants were monitored by measuring the voltage drop across a 10 Ω resistor connected in series with the preparation. Measurements were made using a Gould 0S4020 digital storage oscilloscope (sample rate 2 MHz). The ratio of peak secondary current to primary current (I_s/I_p) was then calculated for all stimulus waveforms examined so that the effects of various secondary pulse magnitudes could be compared.

Results

A reduction in the slope of the recruitment curve using both loop and cuff electrodes was achieved by 1) decreasing pulse width, 2) providing a biphasic pulse, and 3) limiting the delay time in the biphasic pulse. Single pulse and pulse train (30 Hz) stimulation paradigms provided recruitment curves that differed only be a scaling factor. These two sets of results are, therefore, presented in composite. Differences did exist between the properties of the loop and the cuff electrode. These two preparations are, therefore, discussed separately.

Loop Electrode

Decreasing pulse width with loop electrode stimulation caused a reduction in the slope of the stimulus versus force recruitment curve. In addition, as pulse width decreased, the recruitment curves shifted towards greater stimulus amplitudes. Results shown in Fig. 4 are for one of seven preparations studied, and are consistent with the total data pool. Slopes of the normalized curves for each stimulus pulse width are plotted in the lower part of the figure. These results were obtained with zero delay rectangular biphasic pulses, but similar results have been obtained with monophasic pulses.

A bell shaped distribution of motor nerve fiber diameters exists in the medial gastrocnemius nerve [1], [8]. Recruitment from large to small diameter fibers would, therefore, yield an accumulative histogram that is sigmoidal in shape. Because the experimental recruitment curves are of necessity composed of discrete points, the curves in the upper half of Fig. 4 are somewhat discontinuous. Nonetheless, they do appear sigmoidal. This is consistent with the contention that the mechanism by which recruitment is being made more gradual is an increase in the threshold difference between different diameter nerve fibers.

Inclusion of the secondary anodic pulse, i.e., use of the biphasic waveform, shifts the recruitment curve to higher stimulus amplitudes than seen with the monophasic stimulus. Furthermore, when the capacitor value is small, i.e., when the peak secondary current to primary current ratio I_s/I_p is large, the secondary pulse has a greater effect, and the recruitment curves fall farther towards the right. In addition, use of the large I_s/I_p biphasic pulse provided some improvement in re-

TABLE I
I_s/I_p VALUES

Pulse Width (μs)	2.0	0.47	0.22	Capacitance (μF) 0.1	0.047	0.0047	0.001
Loop Electrode							
10	–	0.15	0.22	0.35 ± 0.05	0.57 ± 0.05	2.0 ± 0.3	2.9 ± 0.3
Cuff Electrode							
100	0.13 ± 0.10	0.24 ± 0.11	0.39 ± 0.15	0.51 ± 0.15	0.86 ± 0.20	4.9 ± 1.5	10.0 ± 0.7
25	0.08 ± 0.04	0.13 ± 0.09	0.13 ± 0.07	0.18 ± 0.03	0.28 ± 0.05	1.4 ± 0.3	2.3 ± 0.6
10	0.10 ± 0.01	0.10 ± 0.00	0.12 ± 0.05	0.15 ± 0.03	0.18 ± 0.04	0.73 ± 0.33	1.0 ± 0.04

Ratio of peak secondary pulse current to primary current as seen across the electrode-cat preparation. Loop electrode measurements were done on one preparation at several primary current amplitudes. Cuff electrode measurements were done on four separate preparations. Average values ± one standard deviation are presented here. When no deviation is indicated, only one measurement at those parameter settings was performed.

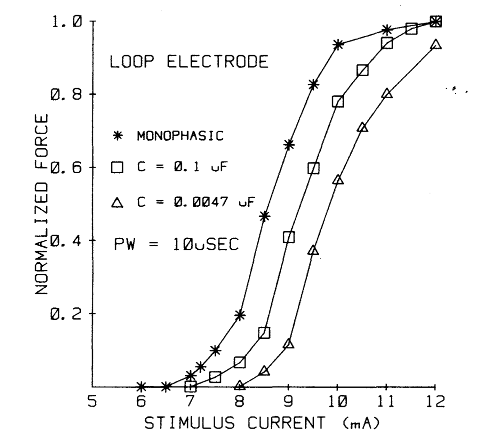

Fig. 5. Effect of inclusion of the secondary anodic pulse with loop electrode stimulation. Capacitor values for the exponentially decaying pulses are indicated. Force normalized to 25 N in all cases.

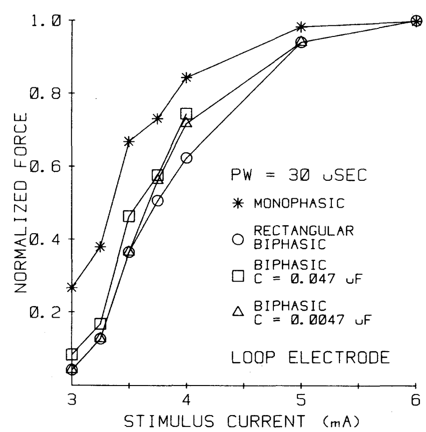

Fig. 6. Comparison of the rectangular and the exponentially decaying secondary pulse with loop electrode stimulation. All curves are normalized to a maximum force of 11.8 N.

cruitment characteristics over that seen with a monophasic pulse, although the decrease in slope was more modest than that obtained through pulse width manipulation. Values of the I_s/I_p ratio are given in Table I for various pulses widths, discharge capacitors, and for both loop and cuff electrode stimulation. These trends were seen in nine animals, and representative results are shown in Fig. 5 for a 10 μs primary pulse width. This result is in agreement with the findings of van den Honert and Mortimer [20].

Direct comparison of the effects of the rectangular secondary pulse with the effects of the exponentially decaying secondary pulse was made in one preparation (Fig. 6). As seen before, the biphasic curves fall at higher stimulus amplitudes than the monophasic curve. Rectangular biphasic stimulation moved the recruitment curve farthest to the right. This is in spite of the fact that the secondary peak current of the capacitive discharge pulses was much higher than the secondary current of the rectangular pulse. Measured I_s/I_p ratios for the 0.047 and the 0.0047 μF capacitors in this preparation are 1.74 and 6.90, respectively. Smaller capacitors, i.e., ones that would produce higher I_s/I_p ratios, were not used because of the power supply limitations of the available stimulators.

The addition of delay to the biphasic waveform shifted the biphasic curve back towards the monophasic result. Delay also produces a slightly steeper recruitment curve. These effects were observed in five cats.

Cuff Electrode

As seen with the loop electrode, a decrease in pulse width shifted the stimulus amplitude versus force curve towards greater stimulus amplitudes, while it also decreased the slope. The stimulus levels with the cuff electrodes were much lower than those for loop electrode stimulation. This is attributed to the limit on current spread produced by the silicone rubber insulating tube. These effects have been shown with the cuff electrode in eight cats. Results from one animal but typical of all animals studied using monophasic stimuli are shown in Fig. 7.

The effect of inclusion of the secondary pulse using the cuff electrode was found to be dependent on the primary pulse width. For small primary pulse width stimulation (10 μs), the biphasic curves shifted towards higher stimulus amplitudes as has been previously described for the loop electrode, and is shown in Fig. 8. Slopes were also reduced, although not to the extent seen for changes in pulse width. Contrary to the small primary pulse results, for large primary pulse width stimulation, the biphasic curve shifted to lower stimulus amplitudes. Fig. 9 shows this effect at a pulse width of 300 μs.

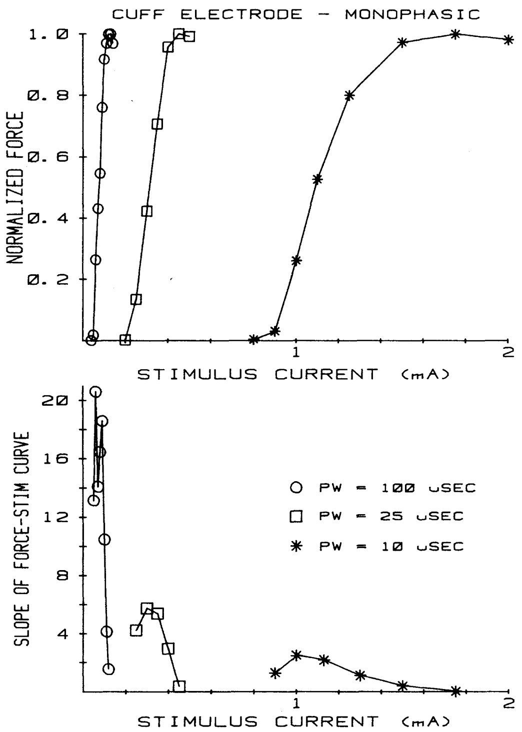

Fig. 7. Effect of pulse width on recruitment curve with nerve cuff electrode stimulation. Monophasic, single pulses used here. Graphs are in same format as Fig. 4. Note the low-stimulus current values relative to loop electrode stimulation. 10, 25, and 100 μs curves normalized to 18.3, 27.1, and 33.0 N, respectively.

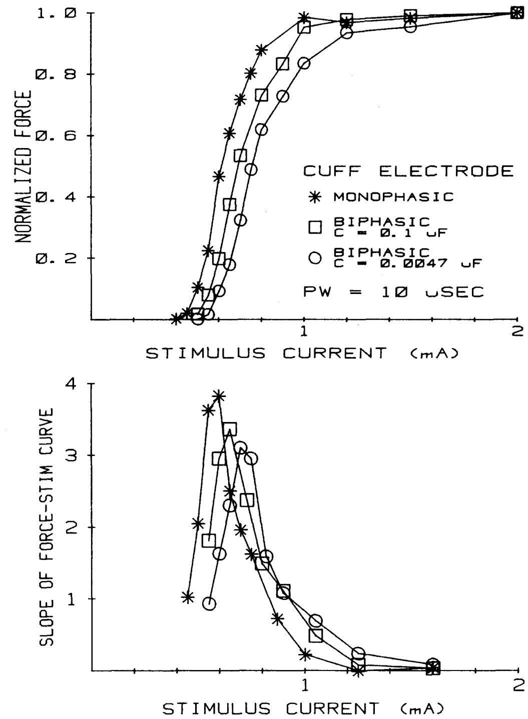

Fig. 8. Effect of inclusion of the secondary pulse with nerve cuff electrode stimulation at narrow (10 μs) primary pulse widths. Biphasic curves are shifted to the right of the monophasic curve. Large I_s/I_p values (small capacitance values) produce greater effects. Monophasic, 0.1 μF, and 0.0047 μF curves normalized to 17.4, 16.8, and 18.3 N, respectively.

The I_s/I_p ratio for the biphasic pulse was 125, a value chosen to ensure a strong secondary pulse effect. When the stimulus was switched to the monophasic from the biphasic waveform at constant amplitude (i.e., when the secondary pulse was removed), the force produced by the muscle *decreased*, indicating that the secondary anodic pulse provided the stimulatory effect. These results were seen in three preparations. The stimulatory effect of the secondary anodic pulse was also produced during small pulse width stimulation provided the I_s/I_p ratio was made sufficiently large (Fig. 10). The curve with the highest I_s/I_p ratio shifted towards the monophasic curve, in contrast to the trend exhibited by the other curves with lower I_s/I_p ratios in Fig. 10 and by all the curves in Fig. 8.

The effects of changes in delay between cathodic and anodic pulses with the nerve cuff electrode were similar to that seen with the loop electrode. Increasing delay shifted the recruitment curve towards lower stimulus amplitudes. This effect was more pronounced at shorter primary pulse widths when the relative magnitude of the delay was large. In addition, the slope of the recruitment curves was less for shorter delays.

Discussion

Threshold differences between nerve fibers of different diameters can be increased by decreasing the width of the excitatory (cathodic) pulse. The addition of a secondary anodic pulse to effect balanced charge biphasic stimuli can enhance this threshold difference. The effect becomes greatest as the delay between secondary and primary pulses approaches zero. The existence of these phenomena was speculated on the basis

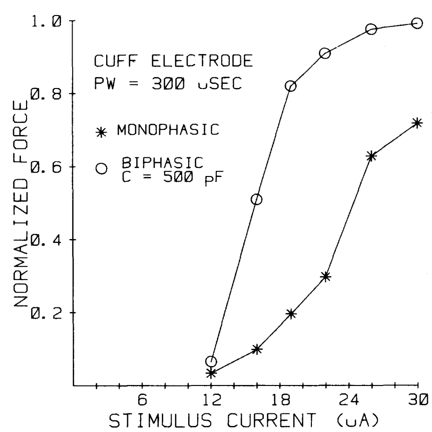

Fig. 9. Effect of inclusion of the secondary pulse with nerve cuff electrode stimulation at wide (300 μs) primary pulse widths. Both curves normalized to 9.9 N.

of results reported for experimental work in human [21] and in animal [20]. A mathematical model for excitation of myelinated nerve developed by McNeal [14] was used to corroborate the speculation and to provide insight into the nature of the phenomena. With this insight and with the knowledge that the efferent nerve supply of muscle contains a distribution of fiber diameters [1], we postulated that the relationship between stimulus magnitude and evoked muscle response (i.e., slope of the recruitment curve) could be altered by adjustment of the width of the primary cathodic pulse and the addition of a secondary anodic pulse. The postulate was tested in cat

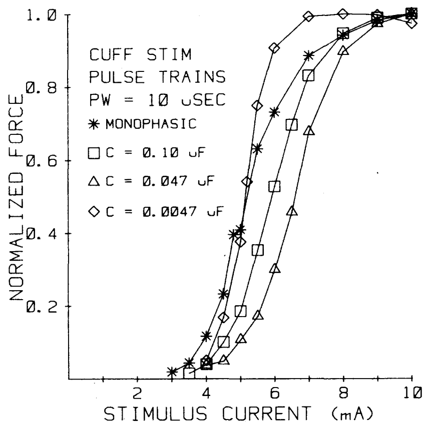

Fig. 10. Special case of the effect of the secondary pulse on the recruitment curves using nerve cuff stimulation. The aberration lies in the fact that the 0.0047 μF curve (diamonds) lies close to the monophasic curve. This is in contrast to the usual result as shown in Fig. 8.

medial gastrocnemius using a loop electrode around the motor nerve and a cuff electrode, which is a loop inside a nonconducting tube. We found that the slope of the recruitment curve could be reduced by reducing pulse width, with the strongest effects observed for pulse widths below 100 μs. (The smallest pulse width tested was 10 μs.) We expect that further decreases in recruitment slope could be obtained at shorter pulse widths. For the present, however, 10 μs is a reasonable limit for most neural prostheses.

Knowledge about the effect of including a secondary pulse to effect balanced charge stimulation is valuable since balanced charge stimulation is recommended to avoid tissue damage [2], [16], and it too could alter the recruitment curve. We found in animal experiments using rectangular biphasic stimuli that the slope of the recruitment curve was reduced by the inclusion of the secondary pulse. The greatest effect was observed when the delay between the end of the primary pulse and the beginning of the secondary pulse was zero. Little difference existed in the recruitment curves for monophasic and biphasic stimuli for delays greater than 80 μs.

The exponentially decaying secondary pulse stimulus used in the animal experiments provided an opportunity to evaluate further the effects of the secondary pulse. The results for narrow primary pulses (e.g., 10 μs) showed that large secondary peak currents relative to the primary pulse current (i.e., large I_s/I_p values) offered no increase in threshold separation over the rectangular biphasic waveform. Furthermore, as the relative magnitude of the peak secondary current was decreased by using larger discharge capacitors, the threshold separation effect decreased. The results for wide primary pulse stimuli (e.g., 300 μs) showed that a steeper slope in the recruitment curve and a shift to lower stimulus amplitudes was obtained when the peak secondary current was large relative to the primary pulse amplitude. These results are a consequence of two phenomena, one relating threshold charge injection to stimulus pulse width, and the other concerned with the existence of virtual anodes or cathodes at the cuff ends. A narrow, high-amplitude stimulus pulse requires less charge to excite nerve membrane than does a wide, low-amplitude stimulus [6], [15]. During the anodic phase of the stimulus current, a point along the nerve some distance from the stimulating electrode will experience a virtual cathode due to current return [15]. When the secondary pulse is large relative to the primary pulse, excitation can occur at the virtual cathode even if only a portion of the stimulus current is flowing. This is because the narrow, high-amplitude pulse is more efficient in excitation, i.e., it requires less charge. The difference in excitation efficiency for stimuli of different duration increases for pulse widths greater than 100 μs and decreases for pulse widths less than 100 μs, approaching zero as the pulse width approaches zero [15].

The reader should be aware that applications using pulse width modulation with balanced charge biphasic stimulation could exhibit difficulties in a transition between the two aforementioned phenomena. A change from primary pulse excitation to secondary pulse excitation in the course of changing muscle force could introduce a discontinuity in the recruitment curve. In systems using closed-loop control [5], this could produce instability.

The results of this investigation show that the recruitment gain (slope of the evoked response versus stimulus amplitude curve) for direct nerve stimulation can be decreased by using narrow stimulus pulses. Furthermore, if biphasic pulses are desired in order to reduce tissue damage at the electrode surface, then a pulse without delay and with a high I_s/I_p ratio would provide some further enhancement of the recruitment characteristics so long as the primary pulse amplitude exceeds the amplitude of the secondary pulse. When the secondary amplitude exceeds that of the primary pulse, excitation may occur at a point distant to the stimulating electrode, introducing a discontinuity in the recruitment curve. The findings we report here have application in physiological systems where a nerve fiber diameter distribution exists. Examples of appropriate application include motor nerve and auditory nerve stimulation [11].

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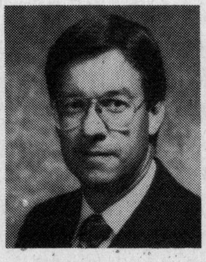

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