A New Strategy for Controlling the Level of Activation in Artificially Stimulated Muscle

T. Ian H. Brown, Ying Huang, David L. Morgan, Uwe Proske, and Andrew Wise

***Abstract*—Distributed stimulation of slow skeletal muscle has previously been used to produce smooth tetanic contractions at low stimulus rates. This involved distributed or interleaved stimulation of portions of the muscle with near equal tension contributions. Extending this to fast and mixed muscle encounters difficulties in getting and maintaining equal twitch responses for the portions. This need has now been circumvented by using distributed stimulation with unequal interpulse intervals. Described here is a microprocessor-based eight channel distributed muscle stimulator that can adjust stimulation timing to produce an optimally smooth tension over a range of stimulus rates even when the portions are unequal. This design is based on modeling results. Distributed stimulation experiments performed on skeletal muscle show that this method can be used to achieve smooth tension at physiological stimulus rates, which should reduce fatigue. This has important implications in functional neuromuscular stimulation (FNS) as well as in enabling experiments to be conducted to characterize the biomechanical behavior of partially activated fast and mixed muscle.**

I. INTRODUCTION

MUSCLE is most frequently activated experimentally by synchronously stimulating all of the recruited fibers with each stimulus pulse. In contrast, natural muscle activation is characterized by large numbers of independently controlled motor units being asynchronously activated [1]. This enables naturally activated muscle to develop smooth tension even at quite low levels of activation by the action of statistical smoothing of many asynchronously activated motor units.

At low rates of synchronous stimulation, muscles produce what is known as an unfused contraction. In other words, the muscle responds mechanically to individual stimulus pulses, but these individual twitch responses do not fuse together to produce a smooth tension. In engineering terms there is a lot of stimulus frequency ripple in the resulting tension.

If instead of synchronously stimulating the muscle, we choose to independently stimulate say six separate, and equal, portions of muscle, and stimulate these six portions using distributed stimulation (Fig. 1), then the resulting tension will have its main ripple component at a frequency equal to the product of the number of inputs and the stimulus rate for each input. This produces an acceptable ripple for much lower stimulus rates and tensions. Such an approach, independently developed by Rack and Westbury at Birmingham, U.K., in 1969 [2] and Brown at Monash, Australia, in 1974 [3], was used to control the level of activation in cat soleus muscle. This technique has been used by several other groups since, but has not been extended to fast and mixed skeletal muscles. The main difficulty in using distributed stimulation to activate fast and mixed muscles is that the previous history of stimulation may change the twitch amplitude [4], so that the portions of muscle that are independently controllable may not be, and not remain, equal. If normal distributed stimulation is used, such inequalities will result in a component of tension ripple at the stimulus frequency.

A computer model presented here shows that the problem of unequal twitch tensions can be very effectively reduced by using optimized unequal interpulse intervals, and that this approach can reduce the stimulus frequency component of tension ripple to zero [5]. A stimulator based on this iterative model provides stimulation pulses that can be used to stimulate real muscle and similarly reduce the level of ripple in the resulting muscle tension [6]. This concept has been verified in cat soleus and gastrocnemius muscles.

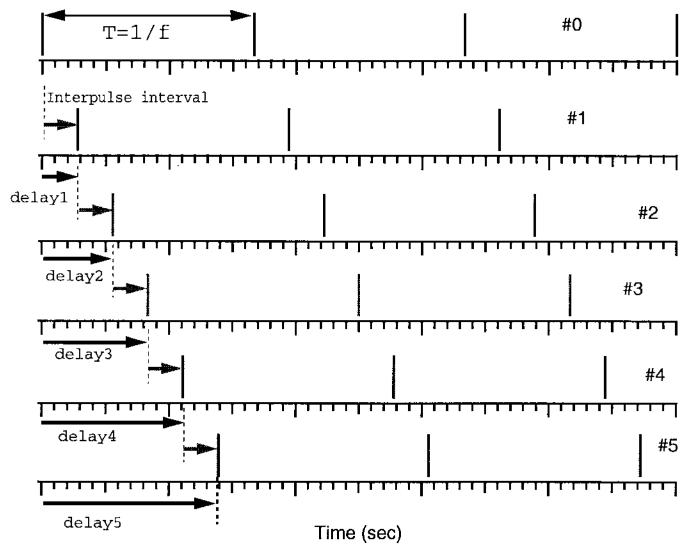


Fig. 1. Representation of pulses for six channel distributed muscle stimulation. Each portion of muscle receives the same stimulation rate, but the pulses are distributed or interleaved. By convention, the quoted stimulus rate is the rate to each portion [2].

Manuscript received July 14, 1997; revised April 3, 1998 and March 15, 1999. This work was supported by the Australian Research Commission and an Australian Postgraduate Research Award to Y. Huang.

T. I. H. Brown, Y. Huang, and D. L. Morgan are with the Biomedical Engineering Centre, the Department of Electrical and Computer Systems Engineering, Monash University, Clayton, Victoria 3168 Australia (e-mail: ian.brown@monash.edu.au; david.morgan@monash.edu.au; YH@basil.eng.monash.edu.au).

U. Proske and A. Wise are with the Department of Physiology, Monash University, Clayton, Victoria 3168 Australia.

Publisher Item Identifier S 1063-6528(99)04474-2.

1063–6528/99$10.00 © 1999 IEEE

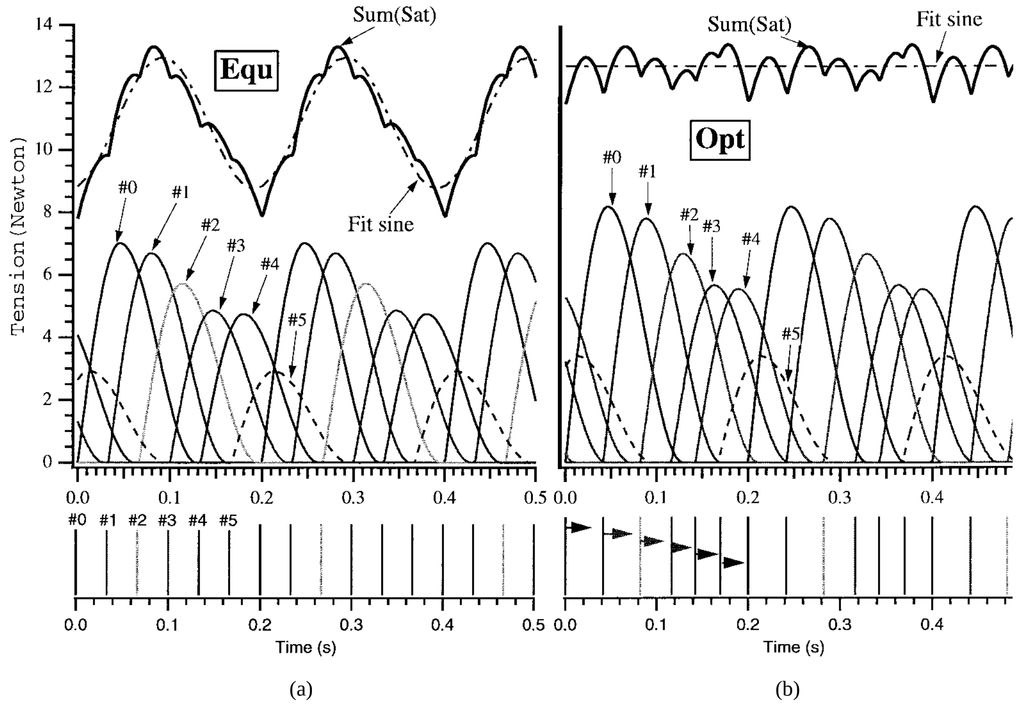


Fig. 2. Model of cat gastrocnemius muscle showing summation of tension from unequal twitches at (a) equal and (b) optimal time intervals. Twitch parameters were chosen to match a cat experiment. Time to peak tension $T_c = 43.5$ ms. Time for half relaxation $T_{1/2R} = 38.5$ ms. Twitch amplitudes are, respectively, 6.47N, 6.18N, 5.18N, 4.48N, 4.39g, and 2.59N. "Sum(Sat)" represents the total tension as described in the text. "Fit sin" represents the fitted sinusoid. The arrows on the lower part of (b) mean that the delays of all channels were increased for the next contraction, compared with the delays at equal interpulse intervals. This occurred in this case because the largest twitches came first.

II. Adjustable Interpulse Interval Model

A. The Model

A computer model was developed to investigate the effectiveness of the approach of adjusting interpulse intervals in a multichannel distributed muscle stimulator. Fig. 1 shows the pulses from a six channel distributed muscle stimulator where the interpulse intervals between the pulses from adjacent channels are equal. In the case of six channels, the interval (in seconds) between a pulse delivered to one channel and a pulse delivered to the next is $T/6$, where T is the period between subsequent pulses to the same channel, or the reciprocal of the basic stimulus pulse frequency (in Hz). These equal interpulse intervals are appropriate in a situation where the six muscle portions all generate equal tensions, but if they do not, as in Fig. 2, then unequal interpulse intervals might produce a smoother tension with less stimulus frequency ripple. In the adjustable interpulse interval stimulator, the interpulse intervals are not equal and are adjusted by an iterative algorithm to minimize the component of ripple at the fundamental frequency $(f = 1/T)$ in the resulting tension.

In the model the stimulus trains produce twitch trains [Fig. 2(a)] which are then summed. This assumes that each twitch train is unaffected by the others. A gently saturating nonlinearity after summation was used to prevent the simulated tension [Sum(sat) in Fig. 2] from rising indefinitely as the rate increased, but had negligible effect at low rates as shown here. For the model results shown, the rising phase of the twitch was represented by a quarter cycle sine wave and the falling phase was represented by a half cycle raised cosine wave. This formulation incorporates directly the simple measured twitch characteristics of time to peak and time for half relaxation as parameters. Other twitch shapes were tried with similar results.

The algorithm for finding optimum intervals in both the model and the experiments worked as follows. First a smoothed version of the tension or simulated tension trace was found by box-car averaging over the stimulus interval T. This removes the frequency component of period T and all its harmonics. This smoothed trace was subtracted from the original trace to get a ripple signal. Then a sinusoid with a frequency equal to $1/T$ was fitted to the ripple by adjusting amplitude and phase. To get the timing for the next contraction, each stimulus train was delayed by an amount proportional to the amplitude of the fitted sinusoid at the phase of the sinusoid at which stimulus pulses in that train occurred. An arbitrary constant in this process was chosen to maximize the rate of convergence without producing oscillations. This method spreads stimuli when the total tension is rising, and brings them together when tension is falling. The iterative process converges quite rapidly to reduce the amplitude of the fitted sinusoid to near zero (Fig. 4). The fundamental component of ripple was chosen as an optimizing variable as it can be reduced to zero, avoiding problems of local minima that may be associated with other variables such as rms or peak to peak ripple which can only be reduced to a minimum.

All simulations were done on a Macintosh computer using the commercial application IGOR (Wavemetrics, Lake Oswego, OR). IGOR was chosen because it is an integrated program useful for visualizing, analyzing, transforming, and presenting numeric data. It has flexible waveform arithmetic,

TABLE I
MODEL RESULT FOR EQUAL AND OPTIMAL TIME INTERVALS AT 5 pps

Ripple	Equal	Optimal
Fundamental	20.51% DC	0.01% DC
RMS	15.22% DC	4.76% DC
Peak to Peak	54.76% DC	20.4% DC

curve fitting, fast Fourier transform (FFT) and other wave analysis functions.

B. Results

Fig. 2 illustrates stimulus pulse timings (lower parts of figure) and tension versus time records (upper parts of figure) produced by the model representing a fast muscle with six unequal components. The lower tension versus time records #0 to #5 represent the tension generated by the individual portions by stimulation at a stimulus rate of 5 pps. In Fig. 2(a) the interpulse intervals are equal, and in Fig. 2(b) the interpulse intervals have been adjusted to minimize the fundamental frequency ripple component of the resulting tension.

The bold traces, the simulated tetanic tension, clearly show that the fundamental component is effectively eliminated, and the root-mean-square (rms) ripple and peak to peak ripple are also significantly reduced by adjusting the interpulse intervals. These results are summarized in Table I.

As stated, these optimal intervals were computed at a stimulus frequency of 5 pps. However, when expressed as a fraction of T these intervals were found to be optimal for any stimulus rate. Once the optimal interpulse intervals have been found for a particular set of muscle portions, the stimulus pulse rate can be varied without disturbing the smooth tension response as represented by minimum fundamental ripple. This is illustrated in Fig. 3.

The model's convergence is illustrated in Fig. 4 where fundamental ripple is plotted prior to interval adjustment when the intervals are equal (iteration 0), and after each iteration. The fundamental ripple, expressed as a percentage of DC tension, fell to below 0.01% after six iterations in this case. The stability and rate of convergence did depend on the arbitrary constant used in adjusting the intervals, becoming slow if the constant was too small and unstable if it was too large. However the exact value was not critical, and satisfactory convergence was maintained when the constant was changed by a factor of more than five. An example of the effect of the constant is shown in Table II. Here the number of iterations to reduce fundamental ripple to 1% of dc for the model described in Fig. 2 is tabulated against the value of the constant, normalized to the value used in Fig 2.

Fig. 4 shows the effect of changing the order of stimulation. It suggests that there is no preferred order. If the twitches of various sizes are stimulated in descending or ascending size order instead of an alternating amplitude order, the fundamental ripple is larger for equal interval stimulation but reduces to the same low value after a few cycles of iteration.

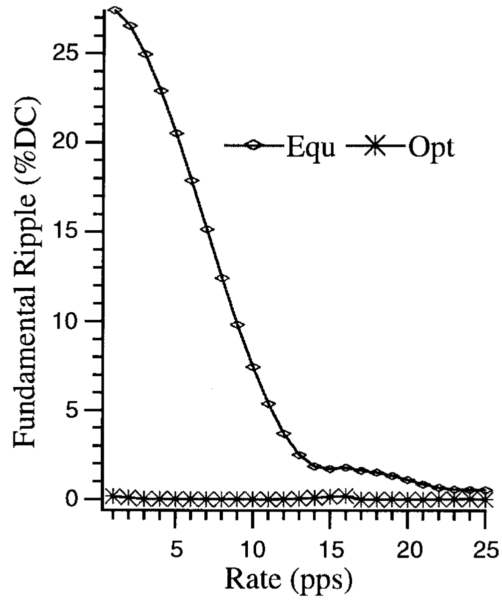


Fig. 3. Fundamental ripple as a function of rate of stimulation for the model in Fig. 2 with equal and optimized interpulse intervals. At high rates the ripple is small in either situation. At low rates, the optimized intervals give near zero fundamental component at all stimulation rates.

TABLE II
THE NUMBER OF ITERATIONS TO REDUCE FUNDAMENTAL RIPPLE TO 1% OF DC FOR THE MODEL DESCRIBED IN FIG. 2

Arbitrary constant relative to value used in fig 2	.04	.2	.6	1	1.4	1.8
No. Iteration	>20	11	3	3	10	Unstable

The model thus suggests that the adjustment of interpulse intervals in a distributed multichannel muscle situation is likely to be very effective in reducing the fundamental component, and other measures of ripple, in the resulting tension.

III. STIMULATOR DESIGN

The model suggests the possibility of building a multi channel stimulator that can adapt the pattern of stimulation to provide a smooth submaximal activation of muscle despite unequal or even varying tension contributions from the muscle portions being stimulated. In practical terms, this involves stimulating the unequal portions of the muscle at equal interpulse intervals, measuring the tension, fitting a sinusoid to the fundamental tension ripple, and iteratively adjusting the interpulse intervals to minimize the fundamental ripple as described for the model. If the tension contributions change, a new optimum pattern can be found by repeating the process.

An eight channel stimulator was designed using CS-64180 miniature microprocessor and two CS-AIO16/4 analog I/O cards. This system is illustrated in Fig. 5. One analog input channel is used to measure tension from a force transducer. Eight analog output channels are used to provide the stimulus pulses. A Macintosh computer is used to collect a record

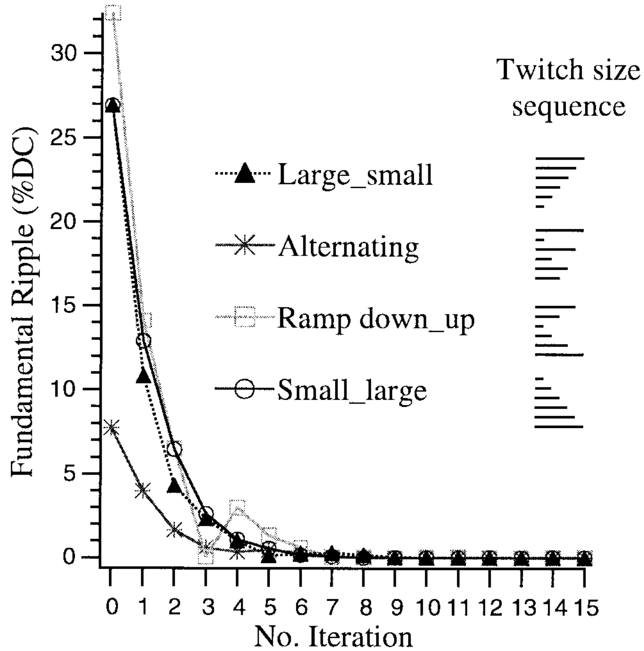


Fig. 4. Reduction of ripple during iteration in the model. The different traces are for different orders of stimulation. Near-zero fundamental ripple is attained rapidly for all orders. Twitch amplitudes are 5.85N, 5.17N, 4.48N, 3.80N, 3.12N, and 2.44N.

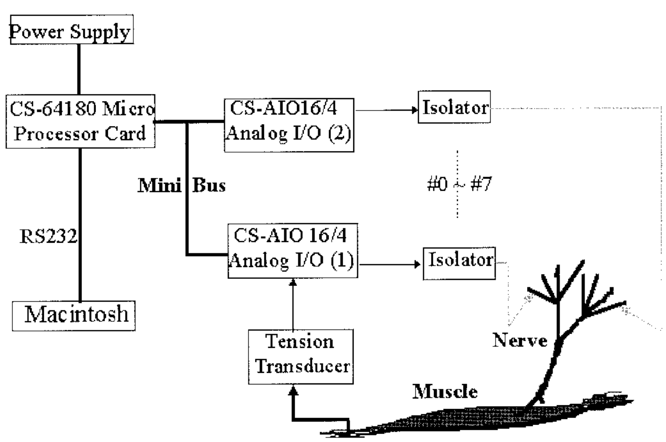


Fig. 5. Block diagram of stimulation system.

of tension, to compute interpulse intervals, and then send these intervals to the microprocessor card via the RS232 serial link. On command, the processor card then generates stimulus pulses in the eight stimulus channels.

This system can deliver either synchronous pulses to each of the eight channels, or distributed stimulation. In the distributed mode the iterative process begins with equal interval distributed pulses being delivered to the muscle to generate a brief contraction. Through a tension transducer attached to the muscle tendon, the tension is recorded by the microprocessor and transferred to the computer. An algorithm in the computer then fits a sinusoid of the fundamental stimulus frequency and computes an adjusted set of interpulse intervals. The computer then sends these new interpulse intervals to the microprocessor which delivers the same period of stimulation with these new intervals. Tension is once more recorded and new intervals

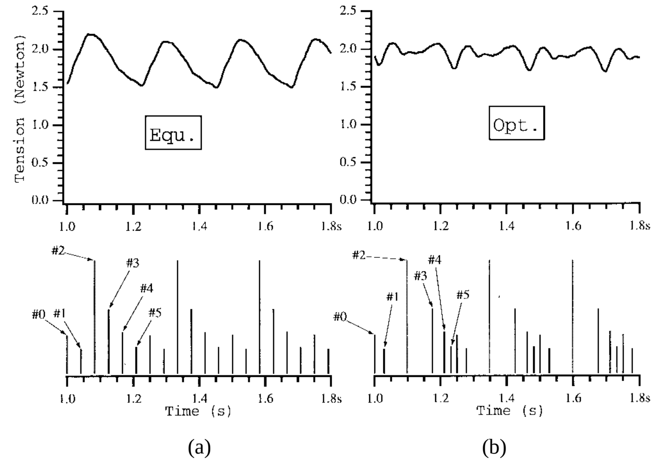


Fig. 6. Tension records from a cat soleus muscle stimulated by six ventral root portions. The stimulation rate was 4 pps and the length was 8 mm shorter than maximum physiological length ($L_{\max}$—8). The upper tension versus time records show the tension recorded during a period of stimulation with (a) equal or (b) optimal interpulse intervals. The lower parts of the figure show the timing of the stimuli with the previously measured amplitudes of the twitches shown by the height of the pulses.

computed in a convergent iterative process that quickly reduces the fundamental ripple component to near zero.

The program of the microprocessor was written using the hi-tech C compiler for the Z80/Z180/64180 (hi-tech software). There are two timers on the CS-64 180. One timer is used to produce stimulus pulses and the other is used to control the sampling of tension. Parameters for pulse generation and tension recording are entered into an IGOR control panel. Pulse parameters include the number of channels, pulse rate, pulse width, stimulation duration, and intervals. Tension recording parameters include data sampling rate and number of samples. The pulse width and stimulation duration are controlled by software loops.

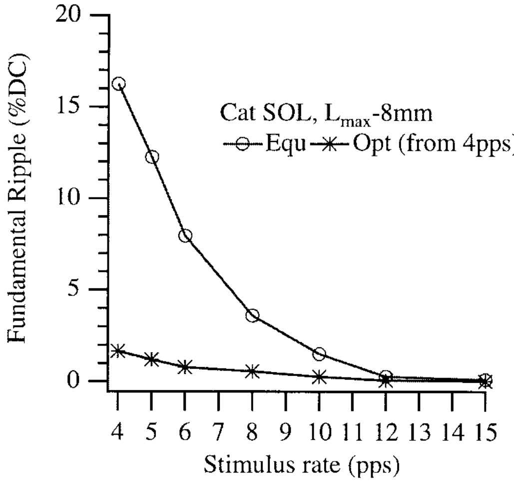
IV. Experimental Evaluation

Preliminary experiments were conducted using isometric contractions of cat soleus and gastrocnemius muscles and divided ventral roots to define the portions of muscle being stimulated. The general preparation was as described Rack and Westbury [2]. Three cats, weighing between 2.1 and 4.2 kg, were used. The animals were anesthetized with 40 mg/kg body weight intraperitoneal sodium pentobarbital and the depth of anesthesia was maintained with supplementary intravenous doses. The muscle was activated via cut ventral roots that were subdivided into six portions which produced a range of twitch tensions. Muscle length changes and tension were recorded. The soleus muscle was initially stimulated at 4 pps and gastrocnemius at 10 pps, and the muscle length was initially set at 8 mm shorter than the maximum physiological length for soleus and 2 mm for gastrocnemius.

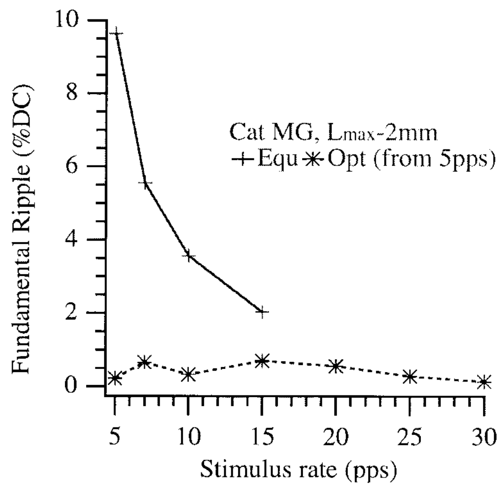
Fig. 6 illustrates stimulus pulse timings (lower parts of figure) and tension versus time records (upper parts of the figure) from cat soleus muscle. As in the case of the model, the fundamental ripple measured in soleus muscle at a stimulus rate of 4 pps was significantly reduced by adjusting the interpulse intervals. With equal interpulse intervals the fundamental

TABLE III
CAT'S RESULTS FOR EQUAL AND OPTIMAL TIME INTERVALS AT 4 pps

Ripple	Equal	Optimal
Fundamental	16.3% DC	1.7% DC
RMS	11.7% DC	4.1% DC
Peak to Peak	38.8% DC	18.3% DC



(a)



(b)

Fig. 7. Fundamental ripple as a function of rate of stimulation for (a) cat soleus muscle and (b) gastrocnemius muscle. Interpulse intervals were optimized at a stimulus pulse rate of 4 pps for soleus and 5 pps for gastrocnemius. At increased rates, the train delays were kept at a constant proportion of the time between successive stimuli in the same train.

ripple during the plateau of tension was 16.3% of dc, whereas after optimizing the interpulse intervals the fundamental ripple was reduced to 1.7% of dc. The results are summarized in Table III.

Fig. 7 illustrates the effect of changing the stimulus pulse rate. The level of fundamental ripple established at one stimulus rate remained reduced when the muscle was stimulated over a range of other stimulus rates, without readjusting the intervals, for Fig. 7(a) slow and Fig. 7(b) fast muscles. Similarly when the muscle length was varied over its physiological range, the optimal interval settings still delivered a tension with much reduced fundamental ripple (Fig. 8). At these optimal intervals, rms and peak to peak ripple were also reduced compared to values obtained with equal interval distributed stimulation (Fig. 9). This lack of dependence of the optimal intervals on length greatly simplifies control when length changes are included.

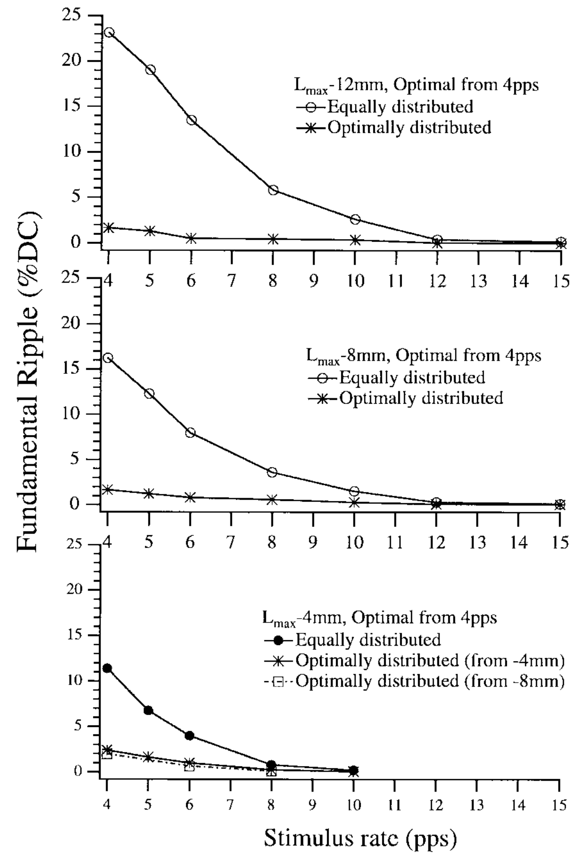


Fig. 8. Fundamental ripple as a function of rate of stimulation for cat soleus muscle recorded at three different muscle lengths. The optimum intervals determined at one length also produced a low ripple at other lengths.

V. IMPLICATION FOR FNS

Functional neuromuscular stimulation (FNS) is the restoration of function to paralyzed muscles by electrical stimulation. This was first introduced by Liberson *et al.* in 1961 [7] when he developed a peroneal nerve stimulator for treating foot drop in hemipalegic patients. Since then, FNS has been used in many other applications [8]–[10]. One major obstacle in applying such a method clinically is fatigue, which during prolonged stimulation at sufficient rate to generate smooth contraction, causes muscle to rapidly lose its contractible strength. Although the causes and mechanisms of fatigue are not well understood, it is clear that high rates of stimulation lead to more rapid fatigue [11], [12]. In most FNS applications to paralyzed limbs, there is usually only one electrode per muscle and tension is graded by varying the amplitude and/or width of the stimulation pulses. All the fibers that are stimulated are stimulated together, known as

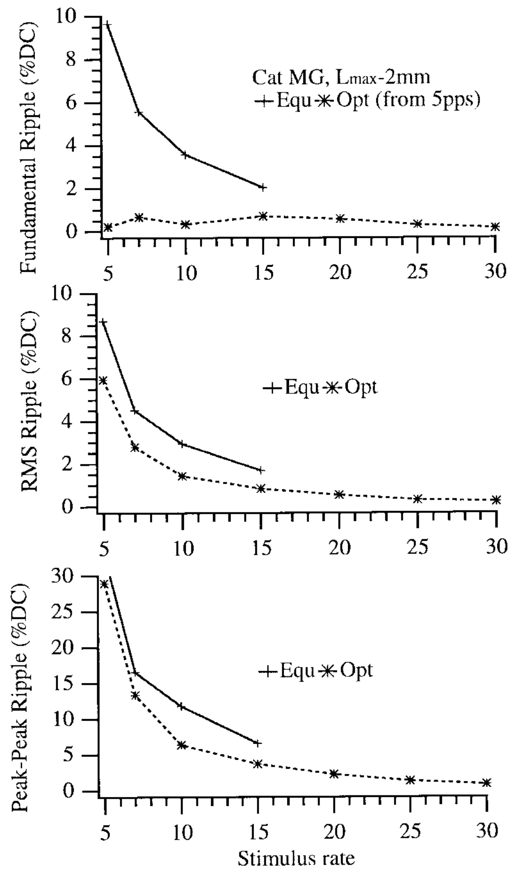


Fig. 9. Fundamental ripple, rms, and peak–peak ripple as a function of rate of stimulation for cat gastrocnemius muscle. Reduction of fundamental ripple always reduced other measures as well. The improvement produced by optimizing intervals for this experiment is less than for the experiment using soleus because the portions were more nearly equal. Compare the lower parts of Fig. 6 with Fig. 2 which models the portions used in this experiment.

synchronous stimulation. Stimulus pulse rates are kept high to insure smooth contraction, with the result that the muscle fibers fatigue rapidly. Distributed stimulation of portions of a single muscle or muscle group would lower stimulus rates for each fiber, helping to reduce fatigue, improve the control of muscle activation, and allow muscles to be smoothly, partially activated. Distributed stimulation has not been applied in FNS because of the difficulties of obtaining equal portions of muscle to independently control, and keeping them equal if fatigue or potentiation occurs more rapidly in some portions than others. The method proposed here only requires that tension measurement presents amplitude and phase information on the fundamental ripple component of tension. High-frequency response and linearity are not essential, reducing the difficulties of applying this approach in an implanted FNS application. Possible sources include measurements of tendon stretch and externally mounted accelerometers.

If muscle portions are obtained by multiple surface or intramuscular electrodes instead of divided ventral roots, some "overlap," i.e., inclusion of some fibers in two portions, becomes likely. Such fibers would be expected to fatigue more rapidly, leading to changes in twitch contributions, and a need to readjust the interpulse intervals. We have not yet investigated the effects of overlap experimentally, but small amounts of overlap should not disturb our conclusions, other than possibly decreasing the time between reoptimization runs.

VI. Implication for Investigating Muscle Biomechanics

Muscle biomechanics is mostly understood from studies undertaken on fully tetanized muscle. It is however of considerable interest to have an understanding of the biomechanics of partially activated muscle. These measurements are difficult to make because, at low rates of synchronous stimulation, the tension developed in usually very poorly fused. Thus, for example, there have been few studies done for the force velocity relationship of partially activated fast or mixed muscles. The biomechanics of partially activated muscle is of significant relevance to the understanding of muscle function in normal use, and in a control situation where the mechanical impedance of the mammalian skeletal system may need to be understood. Where the movements being imposed contain large components at the stimulation frequency, the optimization process can be suspended.

The distributed stimulation approach described in this paper therefore suggests the possibility of conducting a wide range of physiological studies on slow, fast, and mixed muscles at different levels of activation. Similarly, it enables the behavior of these muscles to be examined under conditions where the activation is changing in time.

VII. Conclusion

A method of multichannel muscle stimulation has been investigated in a computer model and in an animal model. The approach uses unequal time intervals between the pulses delivered sequentially to six or so stimulus channels in order to minimize the fundamental component of tension ripple. The model suggested an optimizing stimulator which was built and tested on cat soleus and gastrocnemius muscles. The model and animal experiments confirm the following:

- shifting the stimulus times can substantially reduce the fundamental ripple;
- the required shifts needed to minimize fundamental ripple can be found by a simple iterative algorithm;
- rms and peak to peak ripple values are also decreased along with the fundamental ripple component;
- once the optimal intervals have been established, changing the stimulus rate does not change these intervals;
- optimized interpulse intervals can be obtained for any given muscle length;
- although the technique can be used with three or more inputs, and the benefits increase with more inputs, the practical range is probably four to eight.

References

[1] H. S. Milner-Brown, R. B. Stein, and R. Yemm, "The orderly recruitment of human motor units during voluntary isometric contractions," *J. Physiol.*, vol. 230, no. 2, pp. 359–370, 1973.

[2] P. M. H. Rack and D. R. Westbury, "The effects of length and stimulus rate on tension in the isometric cat soleus muscle," *J. Physiol. (London, U.K.)*, vol. 203, pp. 443–460, 1969.

[3] T. I. H. Brown, "The nonlinear and dynamic behavior of cat soleus muscle," Ph.D. dissertation, Monash University, Australia, 1974.

[4] R. Close, "Dynamic properties of mammalian skeletal muscles," *Physiologic. Rev.*, vol. 52, pp. 129–197, 1972.

[5] D. L. Morgan, T. I. H. Brown, and Y. Huang, "A model of distributed muscle stimulation using unequal input," in *Proc. Australian Physiol. Pharma. Soc.*, vol. 27, no. 1, p. 20P, 1996.

[6] T. I. H. Brown, Y. Huang, and D. L. Morgan, "A programmable stimulator for distributed muscle stimulation with unequal inputs," in *Proc. Eng. Physic. Sci. Med. Health Conf.*, Canberra, Australia, Oct. 1996, p. 192.

[7] W. T. Liberson, H. J. Holmquest, D. Scot, and M. Dow, "Functional electrotherapy: Stimulation of the peroneal nerve synchronized with the swing phase of the gait of hemipalegic patients," *Arch. Physic. Med. Rehab.*, vol. 42, pp. 101–105, 1961.

[8] Y. Handa, T. Handa, M. Ichie, H. Murakami, N. Hoshimiya, S. Ishikawa, and K. Ohkubo, "Functional electrical stimulation (FES) systems for restoration of motor function of paralyzed muscles-versatile systems and a portable system," *Frontiers Med. Biol. Eng.*, vol. 4, pp. 241–255, 1992.

[9] A. R. Kralj and T. Bajd, *Functional Electrical Stimulation: Standing and Walking after Spinal Cord Injury*. Boca Raton, FL: CRC Press, 1989.

[10] P. H. Peckham, J. T. Mortimer, and E. B. Marsolais, "Controlled prehension and release in the C5 quadriplegic elicited by functional electrical stimulation of the paralyzed forearm musculature," *Ann. Biomed. Eng.*, vol. 8, pp. 369–388, 1980.

[11] B. Bigland-Ritchie, "EMG and fatigue of human voluntary and stimulated contractions," in *Proc. Human Muscle Fatigue: Physiologic. Mechanisms, Ciba Foundation Symp. '82*. London, U.K., Pitman Medical, 1981, pp. 130–156.

[12] C. D. Marsden, J. C. Meadows, and P. A. Merton, "'Muscular wisdom' that minimizes fatigue during prolonged effort in man: Peak rates of motoneuron discharge and slowing of discharge during fatigue," *Adv. Neurol.*, vol. 39, pp. 169–211, 1983.

T. Ian H. Brown received the Diploma degree in electrical engineering from the Caulfield Institute of Technology, in 1965 and the Bachelor's of Engineering degree in electrical engineering from Monash University, Australia, in 1974. He received the Ph.D. degree from Monash University, studying the nonlinear and dynamic behavior of cat soleus muscle.

He then received a Postdoctoral Fellowship which enabled him to undertake research overseas at the Department of Physiology, Birmingham University, Birmingham, U.K., on human motor systems' stability. He then spent a further six years in the United Kingdom with research positions at the Biophysics Department of University College London, The Department of Medical Physics, London Hospital Medical College, and Kings College Hospital Department of Medical Physics before returning to Australia to work for the Australian Department of Health. He is currently the Director of the Monash University Centre for Biomedical Engineering, Australia, with research interests in human–motor systems control and prosthetics control.

Ying Huang received the B.E. and M.E. degrees in biomedical engineering at Zhejiang University, Hangzhou, China, in 1987 and 1992, respectively. She began the Ph.D. degree in biomedical engineering at Monash University, Melbourne, Australia, in 1994, studying the smooth submaximal activation of mammalian skeletal muscle. Her dissertation was submitted in January 1999.

Her research interests include function electrical stimulation (FES) of skeletal muscle and biomedical instrumentation. She is now working at Monash Medical Centre as a Biomedical Engineer.

David L. Morgan received the B.E. degree from Adelaide University, Australia, in 1970, the Ph.D. degree from Monash University, Melbourne, Australia, in the Department of Electrical and Computer Systems Engineering in 1977, and D.Sc. degree in the Department of Physiology at Monash University in 1992.

He is currently a Reader in Electrical and Computer Systems Engineering, and Associate of the Department of Physiology, and Research Director of the Monash University Centre for Biomedical Engineering. His has received experience with appointments as a Visiting Professor at Harvard University, Cambridge, MA, where his work has been centered on muscle. His research has concentrated on the mechanical properties of muscle, with submaximal activation and sarcomere nonuniformities.

Uwe Proske received the undergraduate degree and training from Adelaide University, Auastralia, in 1964 and the Ph.D. and D.Sc. degrees from Monash University, Melbourne, Australia, in 1968 and 1990, respectively.

In 1994, he was appointed to a Personal Chair in the Department of Physiology, Monash University. He has spent extended periods studying abroad at Washington University, St Louis, MO, in 1977, Wrzburg, Germany, 1986, and Collge de France, Paris, France, 1981 and 1989. His interests are wide-ranging and include the study of muscle and skin sense organs, in particular, the mammalian muscle spindles, their role in reflex action, and in proprioception. He has studied the mechanical property of muscle, the effects of exercise, and how this may alter the responses of muscle sense organs and their central actions.

Andrew Wise received the B.Sc. (Hons.) degree from Monash University, Melbourne, Australia, in 1995. He is currently working towards the Ph.D. degree in the Department of Physiology at Monash University.

His research interests include electrical stimulation of mammalian muscle as well as the role of sensory information in the perception of limb movement in humans. His long-term goals are to make contributions to the scientific basis of rehabilitation and prosthetics.