A PROPORTIONAL STIMULATOR FOR FUNCTIONAL ELECTRICAL STIMULATION

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Summary

In order for functional electrical stimulation to achieve its full potential, some form of feedback is necessary to control the movement. A proportional stimulator is described which utilizes the integrated M response of the stimulated muscle to regulate contraction. Analysis of the system under closed loop conditions shows that the loop gain can be made high enough to ensure a fast response and yet not lead to instability.

Functional Electrical Stimulation

During the course of normal movement the body, the nervous system has very extensive feedback of information concerning the progess of the movement. Should a disturbance cause a movement to deviate its planned course, then feedback enables corrective action to be taken reflexly, or by intervention of conscious control.

In recent years, a number of laboratories have investigated the possibility of using electrical stimulation of motor nerves to provide movement in partially paralyzed patients with upper motoneurone lesions. Such stimulation has become called functional electrical stimulation (FES) because the object of stimulating is to produce a functional movement; by this we mean a movement which begins and ends at the appropriate times, which has sufficient force, is carried out at the appropriate speed and which achieves a goal desired by the patient without causing him discomfort. FES has been used successfully in the management of foot drop; the corrective functional movement in this disorder is a simple dorsiflexion of the ankle lasting from "heel off" until shortly after "heel strike" during the walking cycle. In this case the FES pulse train is present in amplitude, duration and frequency and any disturbance of the movement cannot be corrected.

Examination of the EMG of the anterior tibial muscles during walking shows that the muscles become active at heel off to dorsiflex the ankle but thereafter activity declines during the swing phase as less tension is needed to maintain dorsiflexion; finally activity increases at heel strike to prevent foot slap and to give lateral stability at the ankle joint. The present pulse train used in FES does not permit these fine gradations of muscular contraction but the tension produced is adequate to improve the gait.

If FES is to have applications in fields other than footdrop, a simple on-off control will not permit the full potential of the technique to be realised. For these applications, graded variations in tension will be necessary and in order to relate tension required to the desired goal, some form of feedback will be essential for a truly functional movement. This paper discusses those parameters of the stimulus which could be altered, the most satisfactory type of feedback and finally the performance of a feedback controlled proportional stimulator is described.

Stimulus Parameters

Those stimulus parameters which can be varied are: (1) Pulse frequency; (2) Pulse width; and (3) Pulse height. Variation of pulse frequency is not useful because at low rates the individual contractions are not fused into tetanus and tremor develops. At rates higher than that needed for fusion of contractions there is no significant increase in tension and the onset of fatigue is hastened.

Variation of pulse height or pulse width produces similar results but there is a definite range of pulse widths which can be used. Too long a pulse becomes very wasteful of energy, and a pulse which is too short and thus of too high a voltage can be painful because the fine nerve fibres in the skin beneath the electrode are stimulated. Within these constraints either can be used successfully. Of the two, pulse height modulation seemed to be the most suitable means of producing graded tension and a constant pulse width of 0.5 msec was found to be satisfactory.

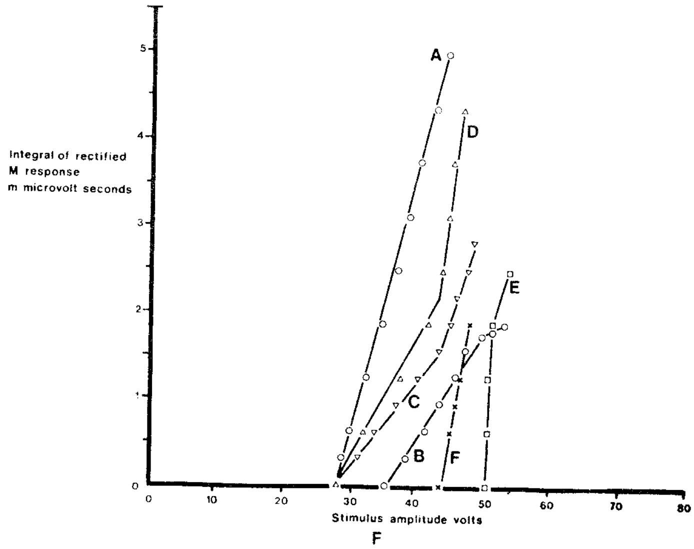
Feedback Signal

Three parameters could be used for feedback: force, movement and EMG. Force is not suitable as it can only be measured as force applied against an external load and therefore cannot be used in no load conditions. Movement, measured as joint angle, could be used but both movement and force measurements do not necessarily relate to the muscle being stimulated but are the resultant of activity of protagonist and antagonist muscles acting across a joint. Movement feedback has been used satisfactorily and is being used in functional electrical stimulation of the hand /1/. However, we decided to use the EMG of the stimulated muscle as it is technically easy to obtain.

As the muscle fibres are synchronously activated by the stimulus the EMG takes the form of a well defined wave, the M response, occurring at a set time after the stimulus but this electrical event precedes the development of tension. The electrical depolarisation of the muscle fibres is only one of a complex chain of events which culminate in the development of tension. Hill /2/ showed that the tension developed by a muscle fibre is not a constant but depends upon the length of the fibre and the velocity of shortening. Although there is no fundamental reason for expecting EMG and tension to be closely related, it is necessary to investigate the relationship between the M response and muscle tension to determine whether the M response can be used as a feedback signal.

A number of workers have studied the EMG/tension relationship under laboratory conditions. Bigland et al /3/ found a linear relationship between area of the M response and tension but not between amplitude and tension. Buchthal and Schmalbruch /4/ and Gottlib and Agarwal /5/ both described a linear relationship between amplitude and tension although both studied rather a limited range of parameters and the latter authors state that the relationship fails if the muscle was voluntarily contracted at the time the stimulus was applied. We therefore decided to use the area under the rectified M response as the feedback signal and to study relationship between stimulus amplitude and integrated M response when the lateral tibial nerve was stimulated at 40 H and the M response recorded from surface electrodes placed over the anterior tibial muscles.

Figure 1 one shows six curves obtained from one normal subject, different curves resulting from slightly different placing of the stimulating electrode with respect to the nerve. The points to note are firstly, that the graphs are approximately straght lines. Some, such as D, seem to be two made up of two straght lines and this would be consistent with the M response being the summation of effects due to two separate muscles of groups of fibres within a muscle having different thresholds to stimulation, each group having a linear response. The shape of the M response suggested this interpretation for, at low values of stimulation, the response was centred around 8 msec after the stimulus pulse but for higher values a second response appeared centred at about 14 msec.



<u>Fig. 1.</u> shows the variation of the M response with stimulus amplitude when stimulating the lateral tibial nerve with rectangular pulses of 0.5 ms duration and at 40 Hz.

The second point is the great variation caused by quite a small movement of the stimulating electrode. All the results were obtained with movements of the electrode of around 1 mm except curves E and F when the electrodes were placed at a point about 2 cm further down the nerve. At this point the behaviour was so unstable that curves could only be obtained with the greatest difficulty. This shows the necessity for using some form of feedback to regulate the stimulator if controlled movement is to be achieved.

As the integrated M wave/tension relationship is linear and the stimulus amplitude is linearly related to the integrated M wave then the overall stimulus amplitude/tension relationship must also be linear.

<u>Proportional Stimulation</u>

Clearly there would be a considerable advantage in having closed loop control of muscle activity. One possible system has been constructed and is shown in block diagram in Figure 2.

The rectified M response is fed into an integrator which controls the level of stimulation yet allows the system to be independent of the absolute value of stimulus threshold. The EMG amplifier includes a gate which is open only for the duration of the M response so that the stimulus artefact and spurious signals occurring at other times are ignored.

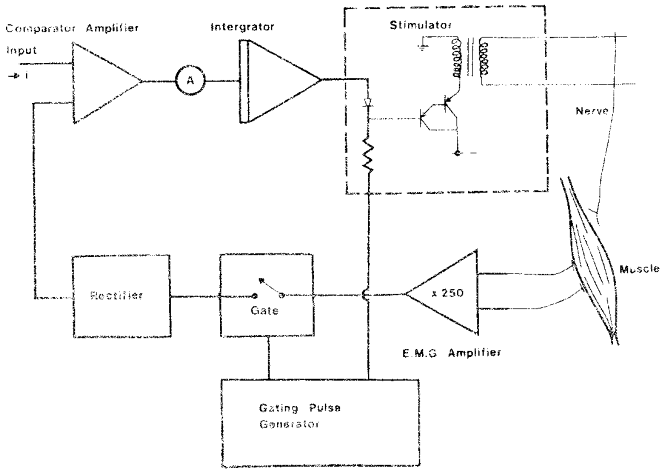


Fig. 2. shows the block diagram of the feedback stimulator system.

Equating currents into the summing junction of the comparator amplifier in the steady state gives

$$mgf = i$$

where m - is the integral of the rectified M response,

g - the gain of the EMG amplifier divided by the resistance (10 kΩ) from the rectifier to the summing junction (numerically this is 25 mA/V),

f - the stimulus frequency, and

i - the system input current.

It can also be shown that the step response should be an exponential rise with time constant $1/f \ln(-Agk)^{-1}$, k being the rate of increase of m with stimulus voltage.

Figure 3 shows typical experimental results. The steady state value of [illegible]. Estimate of Agk from the four step responses [illegible] 1,3. These would give values of [illegible]

0.27, 0.125, 0.125 and 0.133 μS. For comparison the values of k derived directly from the slopes of A and B in Figure 1 are 0.25 and 0.105 μS. These were taken with the electrode in roughly the same place as for Figure 3.

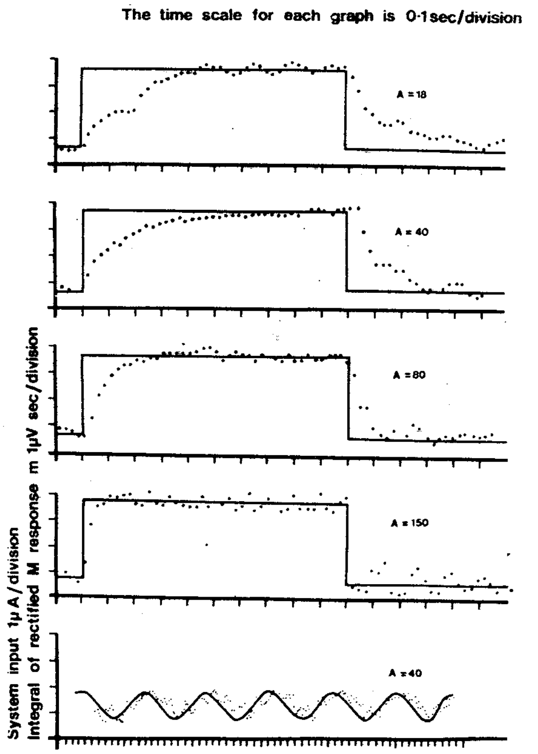


Fig. 3. shows one sine wave and four typical square wave responses of the system. The continuous line shows the system input, and the dots show the values of the integrals of each rectified M response. Stimulus parameters were the same as for Fig. 1.

Deviations from the perfect exponential can be explained by movement of electrodes caused by the muscle contraction, electronic noise, and beat frequency effects of residual mains hum.

However these results do show that the system does achieve the required aims and may be of great value in the design of stimulating systems for producing truly functional movements.

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