Med. & biol. Engng. Vol. 5, pp. 97–109. Pergamon Press, 1967. Printed in Great Britain

CONTROL OF A SKELETAL JOINT BY ELECTRICAL STIMULATION OF ANTAGONISTS*

L. VODOVNIK†, W. J. CROCHETIERE‡ and J. B. RESWICK§

Engineering Design Center, Case Institute of Technology, Cleveland, Ohio, U.S.A.

Abstract—The possibility of restoring controlled movements to a paralysed arm, by means of electrical stimulation of skeletal muscle, is investigated. The concept is applied to the elbow joint of the human arm, by surface stimulation of the biceps and triceps muscles, in a closed-loop position–control system. The results indicate that the method is feasible, but that more basic data is needed before an optimal control is attained.

1. INTRODUCTION

VICTIMS of traffic, diving, or other accidents, as well as polio patients, often continue their lives as partially- or totally-paralysed people. Paralysis of extremities is caused by the disconnection of the nerve paths between the muscles and the brain. These disconnections are classified into two groups.

1. Upper motor neuron lesions; when the disconnection occurs in the spinal cord.
2. Lower motor neuron lesions; when the disconnection occurs in the peripheral nerve between the spinal cord and the muscle.

In group one, the contractibility of the muscle is preserved for years after the injury; but due to disuse of the muscle, it weakens considerably. In the second group the contractibility of the muscle is also maintained, but for a shorter length of time, and the muscle gradually wastes away (atrophies).

To a large extent, paralysed patients are dependent on the help of others, which represents a heavy psychological (and many times physical) burden for both parties. Therefore, any attempt to make the patients as self-sufficient as possible is worth careful consideration.

Currently, the main rehabilitation means are splints powered by gas motors (usually liquid CO_2) or electric motors and activated by movements of normally-innervated parts of the body. In recent years a new approach developed, due to the fact that paralysed muscles are often still contractible. It was suggested that patients might use their own muscle power, instead of bulky externally-powered motors. Muscle fibers can be excited and thus contracted, not only by nerve-impulses, but also by electrical currents. The application of electricity to living tissue is called electrical stimulation. For many decades, physical medicine has used electrical stimulation as a therapeutic method. But not until recently was electrical stimulation applied to paralysed patients in order to restore at least partially the functions of their extremities in everyday living. The first attempts in this respect were made by LIBERSON *et al.* (1962) and LONG *et al.* (1963).

The present report deals with the application of controlled electrical stimulation to an extensor and a flexor muscle of the elbow joint, resulting in controlled movement of the human

* *Received* 30 *July*, 1966.

† University of Ljubljana, Ljubljana, Yugoslavia.
‡ Ph.D. Candidate.
§ Director, Engineering Design Center.

elbow joint. As mentioned previously, paralysis of the arm is caused by a disconnection of the nerve pathways between the muscles and the brain. It is believed that the feedback (afferent) signals from the muscle to the brain contain information relating to the angular position and velocity of the joint as well as the forces acting on the joint. In addition to these internal feedback paths, there is also a visual feedback path to the brain which may be required for the performance of a certain class of movement, e.g. object avoidance, writing, etc. Of course, there are other signals fed back to the brain which relate to tactile sensation, pain and temperature, which may have an influence on the movement of a joint, but we shall consider here only those signals which are the direct result of joint motion.

As a first attempt to duplicate the normal control of skeletal joint movement by means of functional electrical stimulation, only the angular position of the joint was fed back to the controller. External loading of the joint was minimized by supporting the arm in the horizontal plane with a splint structure. All movement of the elbow joint was then constrained to be in this horizontal plane. Figure 1 is an illustration of the control scheme used. In principle, the system is a closed-loop servomechanism which, by selective application of a stimulus to the flexor and extensor muscles, effects the desired position of the joint. Before dealing with the complete system, however, each component will be considered separately.

2. DESCRIPTION OF SYSTEM COMPONENTS

2.1. *Musculo-skeletal system*

A functional model of the torque characteristics for the elbow joint is shown in Fig. 2(a). The model is characterized by the moment of inertia (J) of the forearm and hand combination, and the viscous coefficient about the joint (K_5). The values of these two parameters were determined in a previous report (VODOVNIK, 1964).

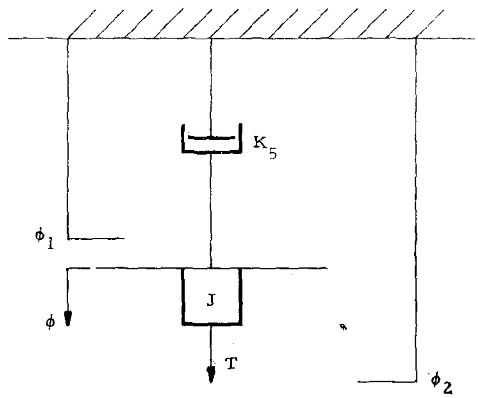

FIG. 2(a). Functional model.

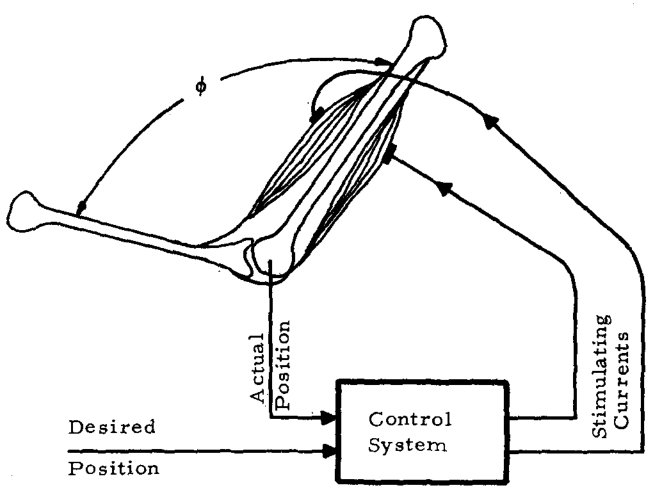

FIG. 1. Position control system for the elbow joint.

The parameters in Fig. 2 are:
T = torque, kg m²/sec²
J = inertia of hand and forearm = 0·09 kg m²
K_5 = viscous coefficient = 1·2 kg m²/sec
φ = included angle of the elbow joint, radians
φ_1 = angle of maximum flexion, radians
φ_2 = angle of maximum extension, radians,

then $T(t) = J\ddot{\varphi}(t) + K_5\dot{\varphi}(t) \quad \varphi_1 < \varphi < \varphi_2$
$T(t) = \infty \quad \varphi < \varphi_1, \varphi > \varphi_2$

in Laplace transform notation,

$$T(s) = [J s^2 + K_5 s]\,\varphi(s) \quad \varphi_1 < \varphi < \varphi_2.$$

Use will be made of this expression and the block diagram of Fig. 2(b) in the model of the over-all system.

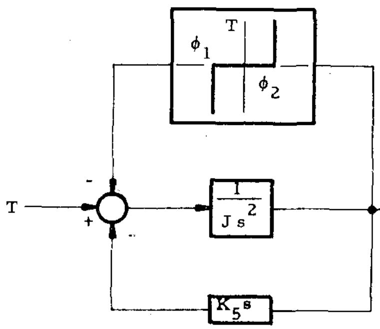

FIG. 2(b). Block diagram of musculo-skeletal dynamics of elbow joint.

Notice that there is no elastic element in the model. This does not imply that there is no elasticity about the elbow joint, merely that it is quite small within the range $\varphi_1 < \varphi < \varphi_2$ in relation to the inertia and viscosity and, therefore, is neglected.

2.2. *Characteristics of muscle stimulation*

When a proper electrical stimulus is applied to the skin overlying the general area of a muscle, the muscle will contract. There is one particular point on the skin above each superficial muscle which, when stimulated, will produce a greater contraction than at any other point over the muscle. This point is defined as the motor point of the muscle. In a previous report by VODOVNIK (1964) it was reported that the speed of this contraction was roughly proportional to the amplitude of the stimulus applied to the motor point. There are further complications, however, in that the magnitude of the stimulus must be above a threshold value and a delay time must elapse before any movement of the joint takes place. A block diagram describing this type of behavior is shown in Fig. 3. In all our experiments the stimulating waveform (V) was used, although there are variations which would also be satisfactory (VODOVNIK *et al.*, 1965). The threshold value (V_T) was found to be quite variable for different individuals, with an average value of about 20 V. If the stimulating voltage (V) then is greater than 20 V, the effective voltage ($V_e = V - V_T$) produces a torque ($K_4 V_e$) about the elbow joint after a delay of τ_0 sec. Although all the values shown in Fig. 3 were obtained from experiments with biceps muscles, it will be assumed for the purposes of analysis that the triceps muscle has similar characteristics.

2.3. *The controller*

A controller was developed to modulate the amplitude of stimulation to the biceps and

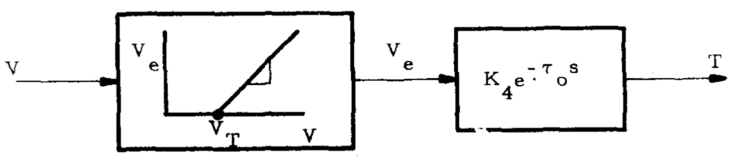

FIG. 3. Characteristics of muscle stimulation.
V = 0–70 V, 200μsec duration, 50 pulses/sec;
V_T = Threshold voltage ≃ 20V;
V_e = effective voltage;
K_4 = ·14 rad kg m²/V sec²;
τ_0 = delay = 0·05 sec;
T = torque, kg m²/sec².

triceps muscles in response to an error signal. This unit as shown in Fig. 4 has a three-channel capacity (possible control of three joints) although only one channel was fully implemented at this time. The command signal (input position) can be generated either internally with the dial (INTERNAL) or externally from a suitable voltage which may be a processed EMG signal (EXT-EMG). The position of the switch (REF) determines which signal is used INT or EXT. A receptacle (FEEDBACK) is also provided on the front panel for a position feedback signal (actual angle of the elbow joint). The adjustment of amplitude (AMPL), threshold (THRESH), and sensitivity (SENS) will be explained in the circuit discussion. The row of output jacks on the lower part of the panel provides a signal for a joint brake (BRAKE) which is proposed (not operational at this time). The remaining four jacks provide the stimulation signals for the biceps and triceps muscles (AGONIST, ANTAGONIST).

A schematic diagram of the controller circuitry is shown in Fig. 5. With switch s-1 in the INT position and the position feedback potentiometer connected, this potentiometer and potentiometer P-1 are in a Wheatstone bridge circuit. A voltage proportional to the error signal (0·30 V/rad) is then fed across the bases of dual transistor Q_1 which is in a differential amplifier circuit having a maximum gain of 200. This gain is adjustable by varying P-2 which is the AMPL adjustment described earlier. The amplified error signal then is fed to the bases of dual transistor Q_2 which modulates the amplitude of the stimulating voltage across the stimulating electrodes. In this stage there are two adjustments provided for each of the two separate modulators; potentiometers P-3 and P-4 adjust the biases on the transistor and are called the threshold adjustments (THRESH). Potentiometers P-5 and P-6 limit the maximum value of the stimulus applied to the muscle and also to a much lesser degree the gain of the modulator; it is called the sensitivity adjustment (SENS).

A block diagram of the controller is shown in Fig. 6. Φ_r represents the input (reference) angle to the controller while Φ_c is the actual (controlled) angle of the elbow joint. The differential amplifier is represented by the block K_2. The output of the amplifier (e') is then fed by parallel paths to the inputs of the modulator stages. A positive e' produces no output from the lower

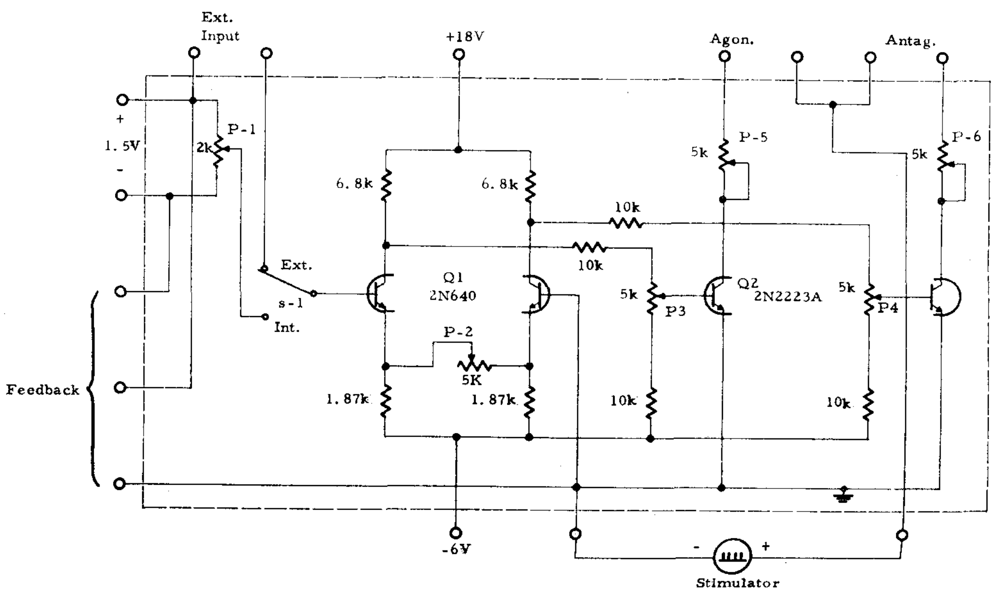

FIG. 5. Schematic diagram of controller.

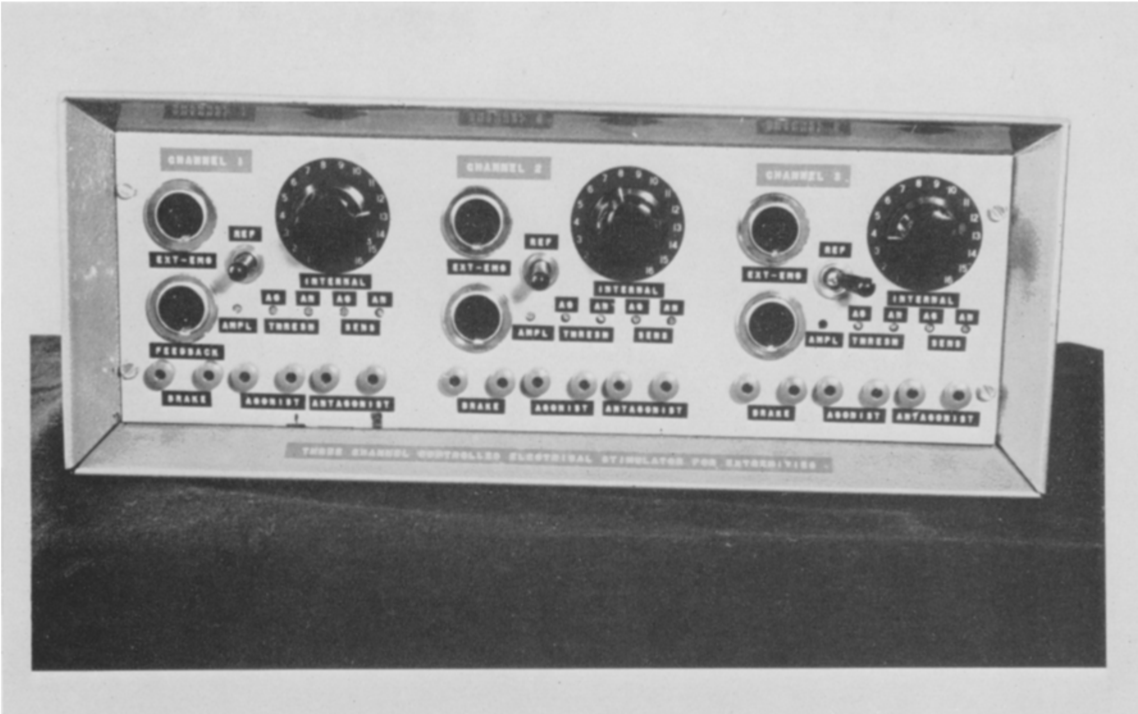

FIG. 4. Stimulation controller.

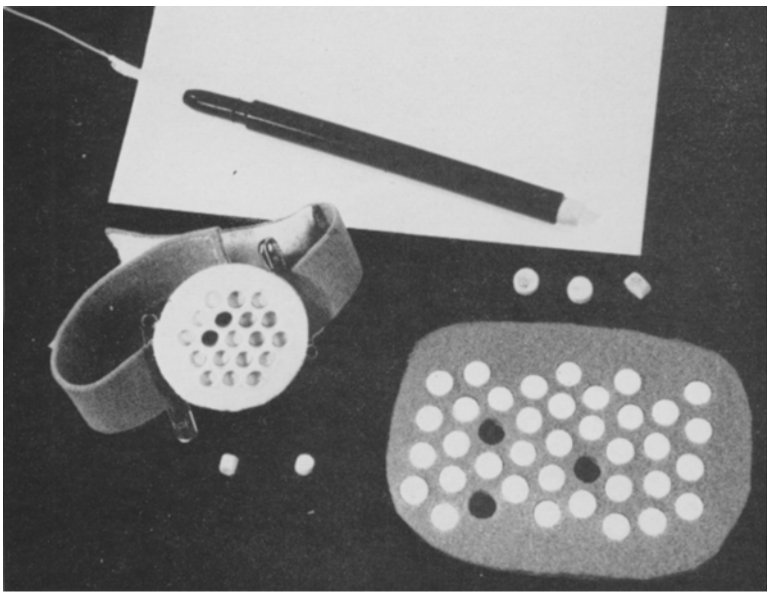

FIG. 9. Electrode matrix schemes.

(facing p. 100)

FIG. 11. Experimental set-up.

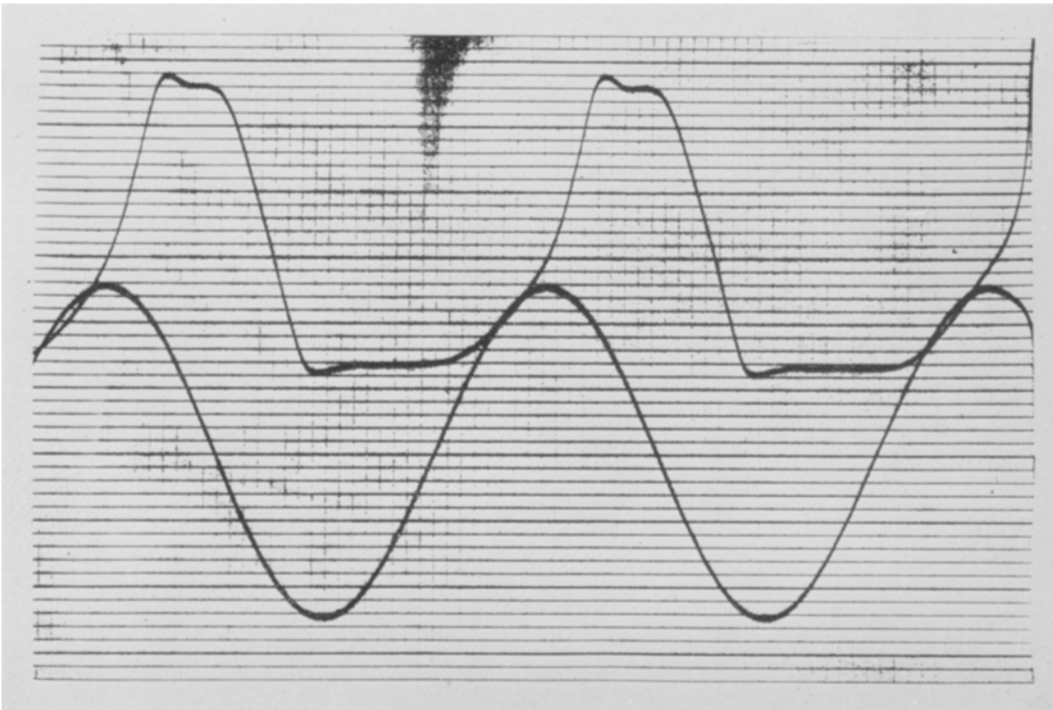

FIG. 12. Input voltage and output elbow angle vs. time, $f = 0{\cdot}3$ c/s.

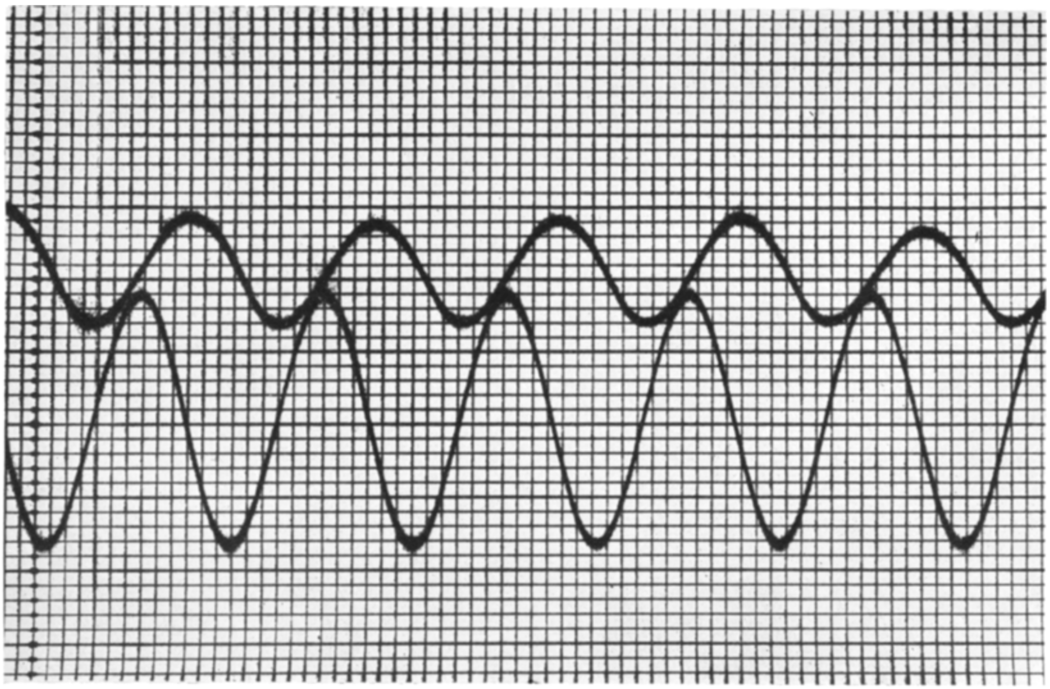

FIG. 13. Input voltage and output elbow angle vs. time, $f = 0{\cdot}8$ c/s.

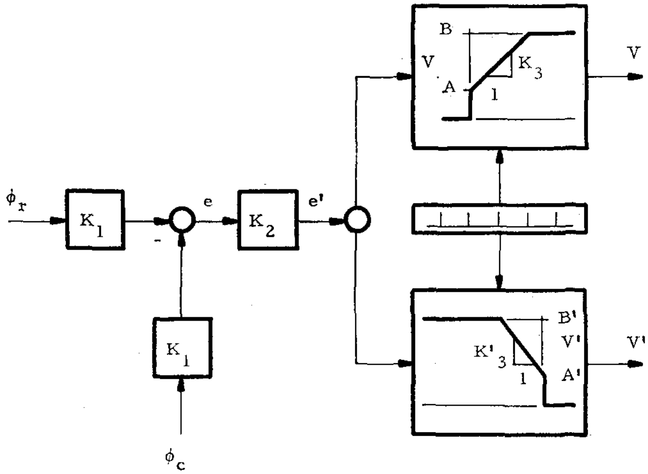

FIG. 6. Block diagram of controller.

modulator; however, the output of the upper modulator is dependent upon e', the threshold setting (A), the gain (K_3), and the saturation (B). The threshold (A = 0–28 V) is set directly by means of a potentiometer (P3) as mentioned in the circuit discussion. The saturation level (B) is also set directly by a potentiometer (P5) (assuming the electrical impedance of the electrode–skin–muscle path remains constant). The saturation level (B) is variable from 28V to 65V, while the gain (K_3) remains approx. constant at 15. A similar discussion applies for the other half of the controller.

2.4. *The over-all system*

A block diagram of the over-all system consisting of the controller, muscle stimulation characteristics, musculo–skeletal dynamics and position feedback loop is shown in Fig. 7. A theoretical analysis of the dynamics of the system may now be performed. If all the controller components are adjusted so that the system operates in a linear mode, the open-loop transfer function for a positive e' can be approximated as (see Appendix)

$$GH(s) = \frac{K_1 K_2 K_3 K_4 \, e^{-\tau_0 s}}{s\,(Js + K_5)}.$$

Using the mean values of the coefficients this becomes

$$GH(s) \cong \frac{70e^{-0 \cdot 05s}}{s\,(0 \cdot 075\, s + 1)}$$

A Nyquist plot of this transfer function is shown in Fig. 8. From control theory the system will be unstable, since the Nyquist plot encircles the singular point at -1, and a means of compensation will be required in the closed-loop operation.

3. EXPERIMENTAL TECHNIQUE

3.1. *Subjects*

All the subjects used in the following experiments were normal (unparalysed) individuals. The subjects were instructed to remain in a passive state during the experiment to simulate a paralysed condition. The reasons for using normal subjects were primarily the unavailability of newly-paralysed patients whose muscles were not degenerated, and secondly because, at least for an upper motor neuron lesion, there is no significant difference between the stimulatability of these muscles and normal muscle. This was confirmed experimentally on two different quadriplegic patients (see Appendix). If the results of the experiments with the normal subjects are favorable, a necessary condition for the application of controlled electrical stimulation on paralysed patients will be that the tone of their muscles be maintained probably by a regular program of electrical stimulation initi-

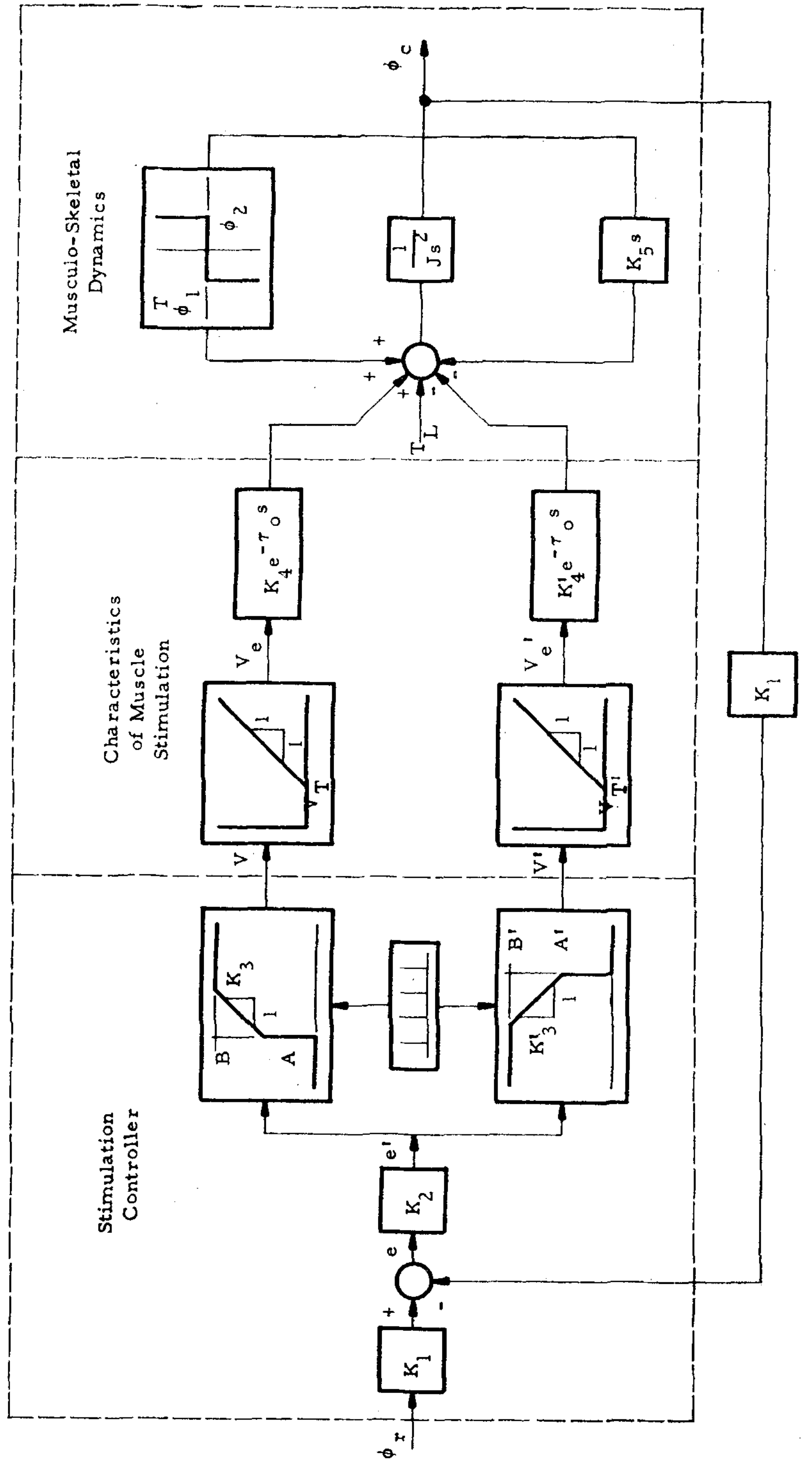

FIG. 7. Block diagram of over-all system.

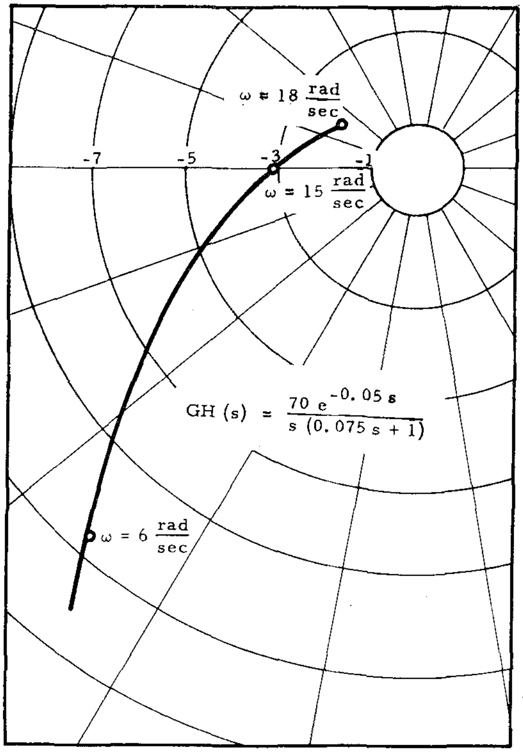

FIG. 8. Theoretical Nyquist diagram.

ated as soon as possible after the accident. Although the following experiments were performed on normal subjects, there seems to be no reason why the results could not be generalized to apply to the paralysed patient.

3.2. *Extensor and flexor muscles of the elbow*

The principal flexor of the elbow joint is the brachialis muscle. Although this muscle possesses a motor point, the resulting contraction due to an external stimulus is less than the contraction of the biceps muscle stimulated with the same current. For this reason, the biceps was always used as the elbow flexor in our experiments. There is a complication, however, in that the action of the biceps during flexion also supinates the pronated forearm. This problem was overcome by allowing the forearm to remain in a position which was mid-way between supination and pronation. In this position no rotary movement of the forearm was observed when the biceps was stimulated and the elbow flexed.

The principal extensor of the elbow joint is the triceps muscle which has three heads; the long head, the lateral head, and the medial head. There is a motor point associated with each of the heads; therefore a choice had to be made among them. The long head is the least powerful in elbow extension and was discounted. In general, the lateral head produced the greatest contraction and was used in all the experiments. In the following discussion, whenever the term flexor is used, we will be referring to the biceps muscle; and whenever the term extensor is used, we will be referring to the lateral head of the triceps muscle.

3.3. *Motor point*

The motor point was previously defined as that point on the skin overlying a muscle at which, if a stimulus is applied, that stimulus will cause that muscle to contract to a greater degree than at any other point. This point is quite easily definable for any one angular position of a joint (or length of the muscle); but during movement of the joint, the muscle shifts in relation to the skin and the motor point likewise moves. That the motor point actually moves was shown experimentally by means of an electrode matrix placed on the skin over the general area of the motor point. Two matrices that were used and the stimulating probe are shown in Fig. 9. The first matrix, which was molded from silastic rubber, was 2 in. in dia. and contained a pattern of closely spaced $\frac{1}{4}$-in. dia. holes. The matrix was attached to the arm by means of a strap with "Velcro" fasteners, then $\frac{1}{4}$-in. dia. saline-saturated felt plugs were inserted into the holes and made contact with the skin. Leakage of the saline solution between felt plugs was kept to a minimum by firmly fastening the silastic matrix to the arm. A large reference electrode (4 in. $\times$ 4 in.) (not shown) was then placed at a remote point on the body (the leg); then the stimulus was applied between each of the felt electrodes and the reference by connecting the anode to the reference electrode and touching the cathode (probe) to the felt electrodes. It is known that the cathode is much more effective than the anode for producing muscular contraction; also since the current density at the anode is very small due to the large size of the reference electrode, there were no visible effects at the anode. At the cathode, the motor point seemed to move quite a bit in relation to the individual electrodes of the matrix for various angular positions of the elbow joint. No conclusions, however, were drawn from this experiment since there was some movement observed between the silastic matrix and the skin. Applying the stimulus between two of the felt electrodes produced no notably different muscular contraction and introduced another variable, so the remote reference electrode was used for all experiments.

The second version of the electrode matrix used (Fig. 9) has an adhesive backing so that it can be affixed directly to the skin, thereby allowing no relative movement between the skin and the matrix. Further modifications include a larger over-all size (4 in. $\times$ 5 in.) and a large electrode size ($\frac{3}{8}$ in.-dia.). The matrix was used on one subject in the same manner as described previously and the motor points of the extensor and flexor were found to lie within an area of approx. 1 in. dia. from full extension to full flexion of the elbow joint. The electrodes used in the experiment were, therefore, made to cover an area of 1 in. dia.

3.4. *Electrodes*

The first electrodes used on the project were the band-aid-snap type. This electrode consists of a band-aid, a piece of aluminum foil, and a snap which joins the two and provides an electrical connector for application of a stimulus.

From experience it was found that this electrode could not be used directly on the skin for fear of burning the skin. That is, unless the aluminum foil is pressed firmly on the skin maintaining a relatively constant impedance between the foil and the skin over the area of the foil, large current densities in isolated areas are possible which can actually burn the skin. An interface was provided between the foil and the skin by inserting a saline-soaked gauze pad of the same size as the foil. This worked well, except that sometimes, during movement, the gauze pad would slip and cause the foil to come in contact with the skin and produce a burn. Also, the band-aid proved to be less than ideal for keeping a constant pressure between electrode and skin. A better electrode, and that which was used in the following experiments, is shown in Fig. 10. The electrical lead is soldered to a 1 in. dia. copper disc and a 1 in. dia. felt pad (saturated in saline) is placed between the copper disc and the skin. The electrode is then attached to the arm over the motor point with adhesive tape (Johnson and Johnson "Elastikon") in a criss-cross fashion. This scheme worked well during all experiments, although the application of electrodes was time-consuming.

3.5. *Subject placement*

The arm of the subject with an extensor electrode and a flexor electrode attached to it was placed in a horizontal splint structure at the shoulder level of the seated subject. A photograph of the experimental set-up is shown in Fig. 11.

3.6. *Controller adjustments*

With the subject set up as above and connected into the system, the controller adjustments are made to minimize the nonlinearities in the input–output relations. That is, we would like the relation between e and T to be $K_2K_3K_4\, e^{-\tau_0 s}$ (Fig. 7) and likewise between e and T' to be $K_2K_3'K_4'\, e^{-\tau_0 s}$. This is accomplished, in stages, by first of all setting K_2 to an intermediate value. (The procedure for linearizing the upper path will be described but the same method applies to the lower path.) The threshold potentiometer (P3) is then adjusted controlling the value of A which we set equal to V_T so that for $e > 0, V_e > 0$. That is, any error signal will cause an above-threshold value of the stimulation voltage to be

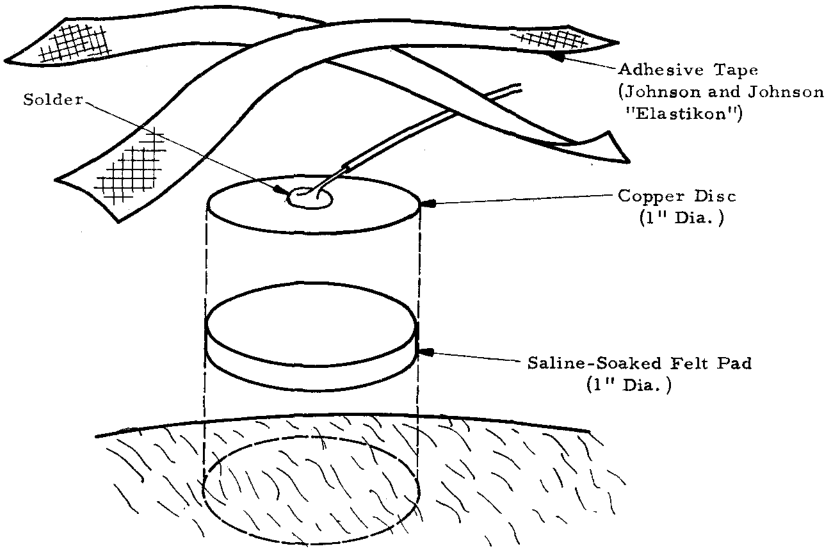

FIG. 10. Stimulating electrode.

applied across the stimulating electrodes. This is done by setting e to zero, then varying P3 until movement of the arm is impending. The saturation level (B) and to some extent the value of K_3 is then adjusted by varying the sensitivity potentiometer (P5). This adjustment is made by setting e to its maximum value, then P5 is adjusted so that V does not saturate. If this condition is not possible to arrive at by varying P5, then K_2 must be reduced and the threshold and sensitivity adjustments are repeated until a satisfactory combination is achieved.

4. EXPERIMENTAL FINDINGS

4.1. *Open-loop test*

In this first series of tests the feedback loop was opened and a Hewlett Packard low-frequency sinusoidal oscillator (Model 202A) was connected across the bases of transistor Q_1. Continuous recordings of the input signal voltage and the angle of the elbow joint (output) were made on a Visicorder recorder. During an experiment, the frequency was varied in steps from 0·1 to 2 c/s for specific values of input amplitude.

The results of the four experiments conducted were quite similar. For frequencies below 0·6 c/s, the records were characteristically as shown in Fig. 12. That is, the elbow moves at a speed of about 2 rad/sec in this case to its extremes of flexion and extension (saturates) independent of the input frequency. When the frequency approaches 0·6 c/s, however, the output follows an approximately sinusoidal oscillation. Figure 13 shows a typical record obtained at a frequency of 0·8 c/s.

It was found that for each of the four experiments conducted that the output lagged the input by 180° at a frequency which was invariably between 0·8 and 1·0 c/s. Also, the open loop gain at this frequency was greater than 1, indicating, as suspected from the theoretical analysis, that the closed-loop system will be unstable unless some means of compensation is provided.

For frequencies of about 1·3 c/s the output followed an erratic variation of low amplitude which at times followed a subharmonic of the input frequency.

4.2. *Compensation*

As a means of compensation for the closed-loop operation, the twin-T network shown in Figs. 14 and 15 was designed and constructed. The null frequency, although shown as 0·87 c/s,

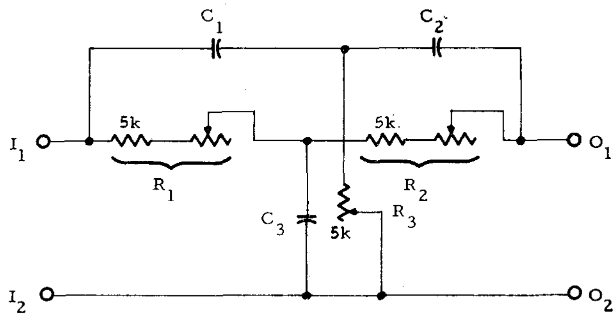

FIG. 14. Twin-T compensator, schematic diagram.
$R_1 = 10{\cdot}4$k;
$R_2 = 5{\cdot}17$k;
$R_3 = 3{\cdot}45$k;
$C_1 = C_2 = 25\ \mu$F;
$C_3 = 50\ \beta$F.

is adjustable by means of varying the potentiometers P1 and P2. The compensator was inserted into the system between the error-detection bridge and the differential amplifier, and has the effect of decreasing the open-loop gain to less than 1 at the critical frequency (0·8–1·0 c/s) with minimum attenuation for the other frequencies, thereby making the closed-loop operation stable.

4.3. *Closed-loop test*

As a further confirmation that compensation was necessary, the loop was closed and the compensator was omitted. An input signal was then applied to the controller and the arm began to oscillate violently at about 1 c/s. The twin-T network was then inserted and the oscillations ceased. The compensator introduced a disadvantage, however, in that the dynamic response of the closed-loop system was affected.

Because of the attenuation of the filter, the frequency response on either side of 0·87 c/s was also affected. At frequencies below 0·4 c/s and to some extent above 1·3 c/s, however, the response was quite good. In the static case the angular position of the elbow joint was controllable within about ± 5°.

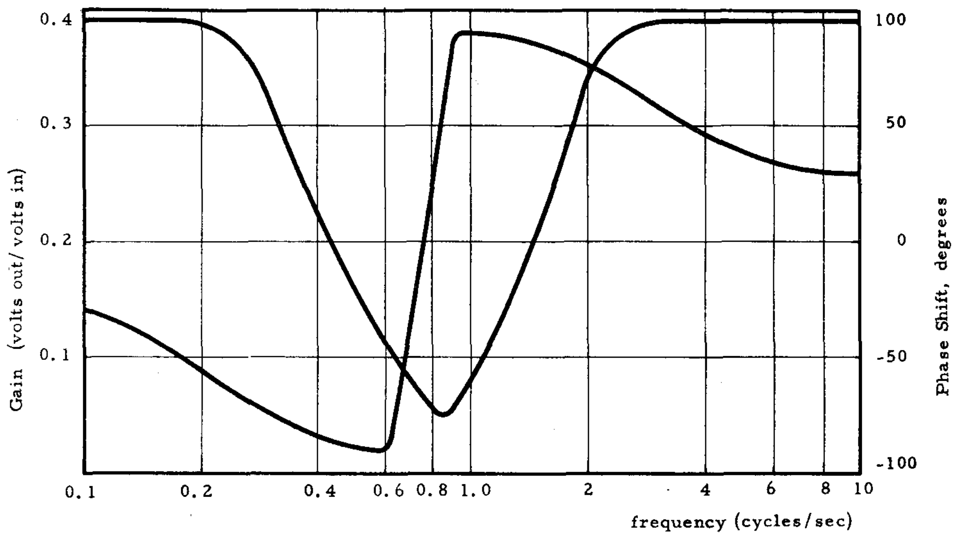

FIG. 15. Twin-T compensator characteristics.

4.4. *EMG-controlled closed-loop*

As a concluding experiment, the amplifier of the electronic bypass described by RESWICK (1964) was used to provide an external signal to the controller. EMG pickup electrodes were placed over the extensor digitorum communis (EDC) of the left arm; then they were connected through the amplifier to the external (EXT) input jack of the controller. When the wrist was relaxed, the flexor received the maximum stimulus, whereas when the EDC was contracted maximally the extensor received the maximum stimulus. The subject was then able to control the position of his elbow joint by varying the contraction of the EDC of his left arm. No quantitative measurements were made at this time but the subject was asked to point to the experimenter's finger as the experimenter moved his finger in a random fashion in the plane of rotation of the arm. With very little training, the subject was able to perform this function quite well for speeds within the normal range of movement. In effect, this experiment showed that it is possible to control movements of both flexion and extension of a joint by contracting a single muscle.

5. CONCLUSIONS AND SUGGESTIONS FOR FUTURE RESEARCH

It has been shown that a reasonable degree of control over the angular position of the elbow joint is possible by modulating the stimulus voltage amplitude applied to the motor points of a flexor and extensor of the joint. This does not mean to imply that this is the best method of stimulating muscle to control movement. Perhaps the modulation of stimulus current would produce a more acceptable response. Stimulus pulse-width modulation or pulse repetition-rate modulation have also been considered as control signals, but seem to offer less promise at this time. In any case, a more basic study of the dynamic properties of electrically stimulated muscle is necessary before a refined controlled system can be built.

Acknowledgements—The authors are indebted to Dr. CHARLES LONG, Chief of Physical Medicine and Rehabilitation at Highland View Hospital for his interest in the project and the laboratory space which he provided. We thank also the subjects who volunteered for the experiments, in particular Dr. WOITEK WOZNY who was most willing and patient. Mr. E. PRENTKE is also to be commended for the construction of the electrical equipment used in the experiments. This research was sponsored by the Vocational Rehabilitation Administration, U.S. Dept. of Health, Education and Welfare, Grant No. RD-1672-M-64.

REFERENCES

LIBERSON, W. T., HOLMQUEST, H. J., SCOT, D. and DOW, M. (1962) Functional electrotherapy: stimulation of the peroneal nerve synchronized the swing phase of the gait of hemiplegic patients. *Proc. of the Third International Congress of Physical Medicine, Washington, D.C.*, 1960, pp. 705–710. Westlake Press, Chicago.

LONG, C. II, and MASCIARELLI, V. D. (1963) An electrophysiologic splint for the hand, *Arch. phys. Med. Rehabil.*, **44**, 499.

VODOVNIK, L. (1964) The dynamic response of a musculo–skeletal system due to electrical stimulations. Engineering Design Center Report No. EDC 4-64-10, Case Institute of Technology.

VODOVNIK, L., LONG, C. II, REGENOS, E. M. and LIPPAY, A. (1965) Pain response for different tetanizing currents. *Arch. phys. Med. Rehabil.*, **46**, 187.

RESWICK, J. B., VODOVNIK, L., LONG, C. II, LIPPAY, A. and STARBUCK, D. (1964) An electronic bypass for motor neuron lesions. *Proceedings 17th Annual Conference on Engineering in Medicine and Biology*, p. 15, Cleveland.

VODOVNIK, L., LONG, C. II, RESWICK, J. B., LIPPAY, A. and STARBUCK, D. (1965) Myo-electric control of paralyzed muscles. *Trans. I.E.E.E.*, **BME-12**, 169.

VODOVNIK, L., LIPPAY, A., STARBUCK, D. and TROMBLY, C. A. (1964) A single channel myo-electric stimulator. Engineering Design Center Report No. EDC 4-64-9, Case Institute of Technology.

CROCHETIERE, W. J. (1964) A preliminary analysis of the dynamics of the human finger. Thesis for the M.S., Case Institute of Technology, Cleveland, Ohio.

KANTROWITZ, A. (1963) Electronic physiologic aids. *Report Maimonides Hosp.*, Brooklyn, New York.

RUCH, T. C. and FULTON, J. F. (1960) *Medical Physiology and Biophysics*, W. B. Saunders Co., Philadelphia and London.

GIBSON, J. E. (1963) *Nonlinear Automatic Control*. McGraw-Hill, New York.

APPENDIX

Theoretical open-loop transfer function

From Fig. 9

$$V_e = V - V_T$$

$$V = e'K_3 + A$$

$$\therefore V_e = e'K_3 + A - V_T$$

let $A = V_T$ then

$$V_e = e'K_3 = eK_2K_3$$

$$\varphi_c(s) = \frac{V_e K_4 e^{-\tau_o s}}{s(Js + K_5)} = \frac{eK_2K_3K_4 e^{-\tau_o s}}{s(Js + K_5)}$$

$$G(s) = \frac{\varphi_c}{e} = \frac{K_2K_3K_4 e^{-\tau_o s}}{s(Js + K_5)}$$

$H = K_1$

$$GH(s) = \frac{K_1K_2K_3K_4 e^{-\tau_o s}}{s(Js + K_5)}$$

$K_1 = 0{\cdot}3$ V/rad
$K_2 = 135$
$K_3 = 15$
$K_4 = {\cdot}14$ rad kg m^2/V per sec^2
$K_5 = 1{\cdot}2$ kg m^2/sec
$J = 0{\cdot}09$ kg m^2.

Substituting:

$$GH(s) = \frac{70\, e^{-0{\cdot}05s}}{s(0{\cdot}075s + 1)}$$

$$GH'(j\omega) = \frac{70}{j\omega({\cdot}075\, j\omega + 1)}$$

$$\mathrm{Re}\,[GH'(j\omega)] = -\frac{5{\cdot}25}{1 + {\cdot}0056\,\omega^2}$$

$$\mathrm{Im}\,[GH'(j\omega)] = -\frac{70}{\omega(1 + {\cdot}0056\,\omega^2)}.$$

The lag term $e^{-0{\cdot}05j\omega}$ contributes a phase angle of $-0{\cdot}05\omega$ rad to the plot of $GH'(j\omega)$ to produce the $GH(j\omega)$ curve which is plotted in Fig. 10.

Stimulation of paralysed muscle

A quantitative measure of the stimulatability of a muscle can be obtained from an experimentally determined strength–duration curve. This curve expresses the relation between least amplitude (strength) of an applied stimulus and least time during which it must flow in order to reach threshold. There is a minimal stimulus amplitude below which excitation of the muscle does not occur. Figure 16 shows a set of strength duration curves plotted for a number of muscles of a quadriplegic patient. The curves were obtained by placing a stimulating electrode over the motor point of the muscle and measuring for each pulse duration the minimum stimulus intensity (current in this case) which will cause the muscle to contract visibly. Since the curve approaches its minimum value (rheobase) quite gradually, the duration at which this occurs would provide a very inaccurate measure of the stimulatability of the muscle. For this reason, a more easily determined measure is defined, the chronaxie. Chronaxie is the length of time a stimulus of twice the rheobase amplitude must flow in order to produce contraction of the muscle. The chronaxie as measured in Fig. 16 is seen to have a range of 0·15–0·30 msec. The range for normal muscle is 0·04–0·35 msec. The experiment was repeated on the flexor digitorum superficialis of a different quadriplegic patient and a similar result was obtained.

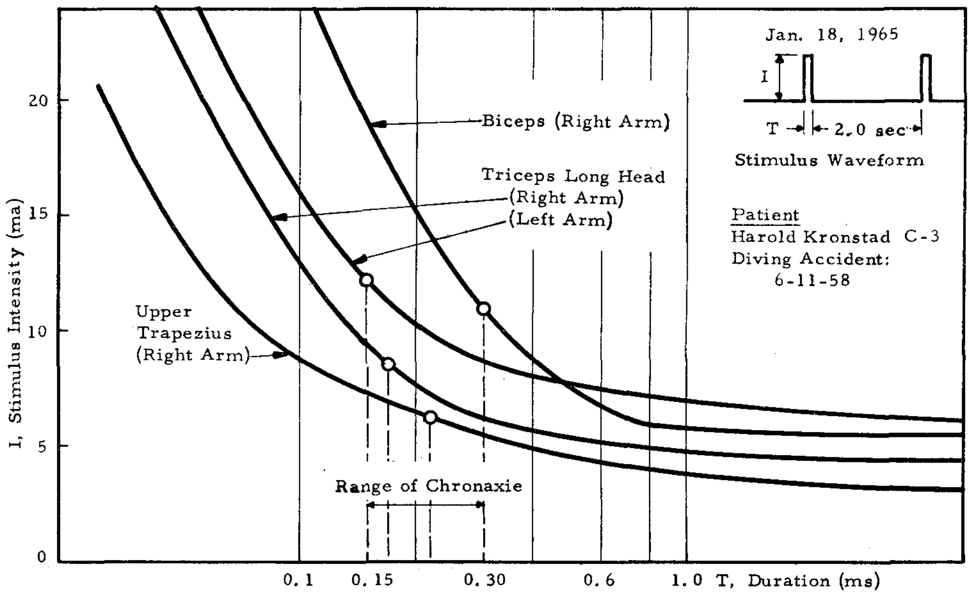

FIG. 16. Strength–duration curves of paralysed muscles.

COMMANDE D'UNE ARTICULATION PAR STIMULATION ELECTRIQUE DU MUSCLE ANTAGONISTE

Sommaire—On étudie la possibilité de redonner des mouvements contrôlés à un bras paralysé grâce à une stimulation électrique appliquée au muscle. On applique ce principe à l'articulation du coude d'un bras par stimulation superficielle du biceps et du triceps grâce à un système de contrôle de la position par une boucle de réaction. Les résultats indiquent que la méthode est possible mais que davantage de renseignements de base sont nécessaires avant qu'une commande optimale soit obtenue.

KONTROLLE EINES GELENKS DURCH ELEKTRISCHE STIMULIERUNG DER ANTAGONISTEN

Zusammenfassung—Die Möglichkeit der Wiederherstellung kontrollierter Bewegungen eines gelähmten Armes durch elektrische Stimulierung der Skelettmuskeln wurde untersucht. Diese Überlegungen werden auf das Ellbogengelenk angewandt. Bizeps und Trizeps werden oberflächlich in einem geschlossenen Positionskontrollsystem stimuliert. Die Ergebnisse zeigen, daβ die Methode anwendbar ist. Es müssen jedoch mehr Grundtatsachen bekannt sein, bevor eine optimale Kontrolle erreicht werden kann.