a.—for l between 0.0 and 1.0 ms.

$A_{min.} = g_1(l)$

$A_{max.} = g_2(l)$

—for l greater than 1.0 ms.

$A_{min.}$ = constant

$A_{max.}$ = constant

b.—$F_l = h_1(A),\ A_{min.} \leq A \leq A_{max.}$ [1]

$F_A = h_2(1),\ A_{min.} \leq A \leq A_{max.}$ [2]

It is now interesting to explain as much as possible the form of the equations [1] and [2]. Figures 7 and 9 show the variations of the force for equal-width and equal-amplitude pulses. The general aspect is the same as Figure 6; as far as an important percentage of the maximum force allowable is concerned, it is possible to linearize this force. Such a linearization will make the solution of the control problem easier.

To bring the force developed by a muscle to a certain level, there are a great number of possible combinations of the control variables inside the zones previously defined. The evolution of the parameters is determined according to the following criteria: search for a minimization of energy provided to the nerve, search for the development of the maximum force, and search for a minimization of the time taken for the force to be established. Compromises may be needed to allow reasonable forces to be sustained for useful periods without fatigue.

PART II.—OPEN LOOP CONTROL OF MUSCULAR CONTRACTION

INTRODUCTION

The first part of this work defined the usable space for the stimulating parameters (frequency, width, and amplitude of rectangular voltage pulses) of the internal and external popliteal sciatic nerves which control the contraction of the anterior tibial and triceps surae muscles of the dog. Using amplitude as a parameter, however, does not give repeatable results because the stimulation depends on the particular electrodes used. The electric charge[b] sent to the nerve is a better parameter because

b The charge (q) is the integral of the current: $q = \int i \cdot dt$.

it is independent of the electrodes and characteristic of the degree of muscle contraction. In practice, it is easier to measure the applied voltages and durations than to measure charge, and the results in both cases are similar if the real part of the impedance Z between the two electrodes remains constant.

The results given here have been obtained with new improved electrodes, compared to those used in Part I. The silicon has been replaced by a very flexible material (Rhodergon). The nerve-electrode junction has been improved, so there is less risk of damage to the nerve, which could become somewhat swollen during the experiments. The real part of the nerve-electrode impedance decreases significantly causing the excitation thresholds to decrease with it.

With these new experimental conditions, it is possible to synthesize stimulation signals, so that the force developed (isometric mode) or joint rotation (isometric mode) can be controlled in the desired manner.

THE CONTROL FORCE IN ISOMETRIC MODE

As in the previous paper (Part I), the force developed as a function of amplitude and pulse width is once more shown here. In Figure 10, the

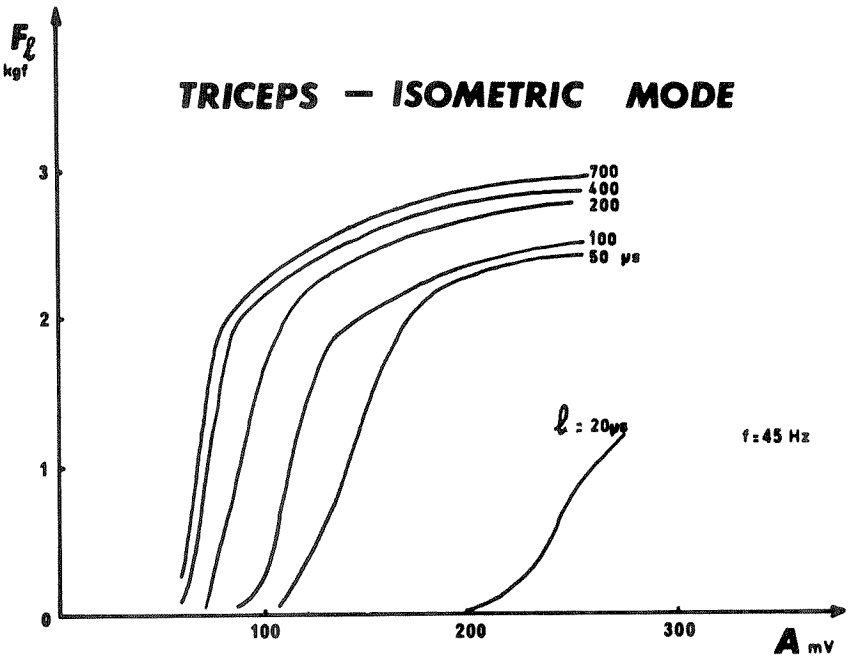

FIGURE 10.—Force developed by triceps in relation to stimulating amplitude for fixed widths.

independent variable is width; in Figure 11, it is amplitude. These curves were obtained using the new electrodes and the triceps of the dog. The frequency was held constant at 45 Hz.

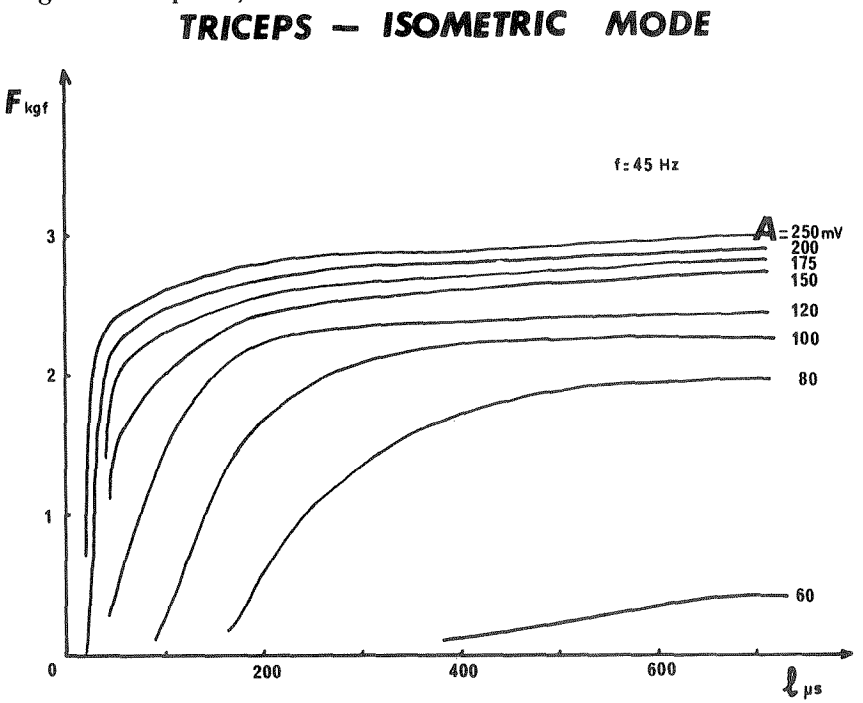

FIGURE 11.—Force developed by triceps as function of the width of stimulating impulses, for fixed amplitudes.

To bring the developed force by the muscle from zero to any desired value, it is possible to use variations of width, amplitude, or any other combination of these two parameters. It was seen that keeping frequency at a fixed point is more interesting (see Part I). But what value should be used in the 25-60 Hz range? To help in making this choice, three criteria that influence the design of muscle stimulators have been investigated.

A. Minimization of Energy

Minimizing the energy delivered to the nerve is important both for maintaining the nerve in good condition, and for miniaturizing and possibly implanting the stimulator. The equation for the power (or energy per unit time) is:

$$P = \frac{f \cdot l \cdot A^2}{Re\,(Z)}$$

where f is the pulse frequency, l the pulse length, A the amplitude and Re (Z) the real part of the nerve impedance. The force as a function of the power of the stimulating signal is shown in Figure 12.

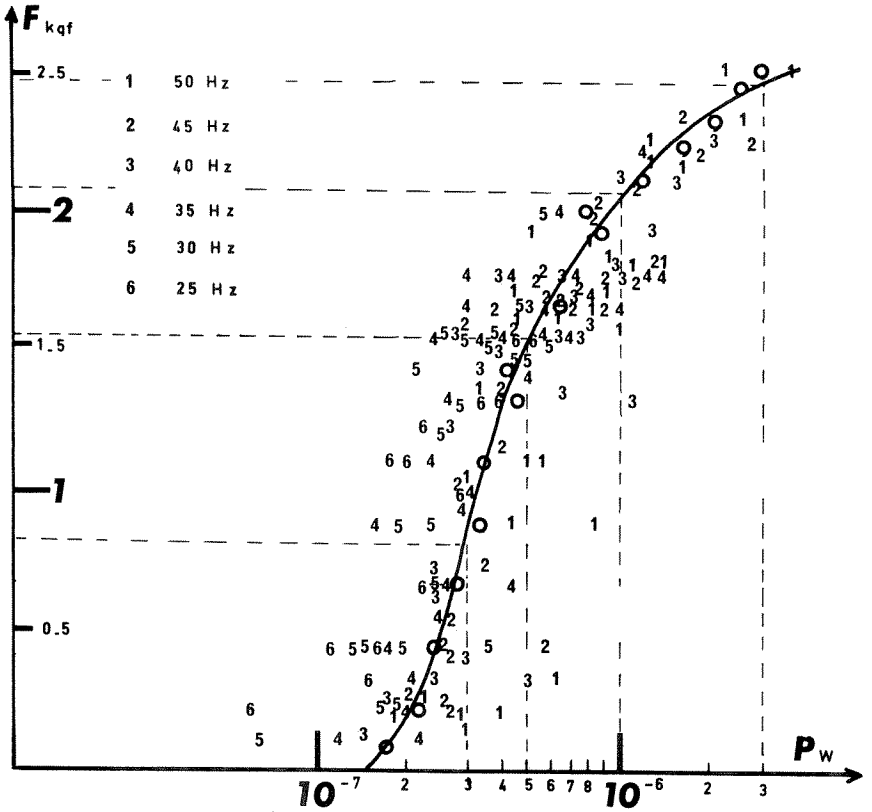

FIGURE 12.—Force developed for each frequency in relation to the power of stimulating signal.

Despite a fairly large dispersion of the representative points, a curve is obtained by averaging all the values at a given force but a varied frequency. Figure 13 shows the averaged results, which exhibit both a saturation and threshold. Although numerous mathematical expressions can be used to approximate this curve, we have chosen very simple ones. Segment AB of the curve is almost a straight line, which is expressed by the following linear equation:

$$F \text{ kgf} = 5.31 \,.\, 10^6 \, P_w - 0.85$$

$$(1.6 \,.\, 10^{-7} < P_w < 4 \,.\, 10^{-7})$$

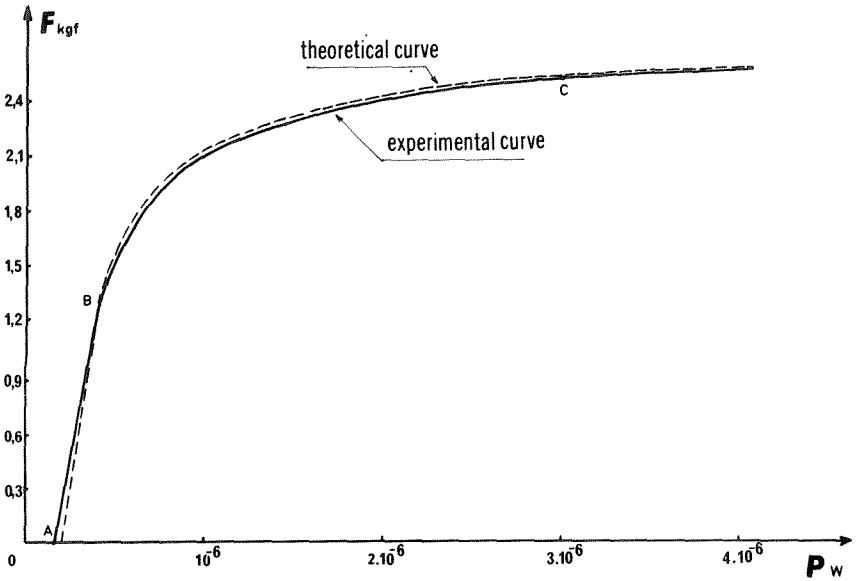

FIGURE 13.—Means of force versus power of stimulating signal dashed line: Fkgf = 2,7 – (0,57. 10^{-6}) /Pw.

Segment BC is expressed by the following hyperbola:

$$F\ kgf = 2.70 - \frac{0.57\ .\ 10^{-6}}{P_W}$$

$$(P_W \geqslant 3.10^{-7})$$

In order to find a pulse width which represents the minimum power consumption, the ratio (F/P) is drawn as a function of P for each pulse width (Fig. 14). The results show that to minimize the energy consumption, there is no preferential width. (Actually, the integral of the ratio F/P over all P values must be considered). The same calculation is required for each frequency. The results are relatively identical and in terms of energy spent inside the selected frequency range, no pulse frequency appears to be preferential.

The criterion of energy minimization does not give the necessary elements to fix either of the parameters. It is possible to control the force from zero up to its maximum value while continuously minimizing the power (e.g., $A^2 l$). In Figure 15, lines of equal force are represented in logarithmic coordinates (log A and log l). To move on this plane following the minimum energy rule, the displacement from one curve to

another should stay on the tangent point of the curve F = constant with the parallel to the straight line:

$$\log l + 2 \log A = 0$$

This line is shown dashed in the lower left area of Figure 15 and a series of parallel lines are shown tangent to the curves for constant force.

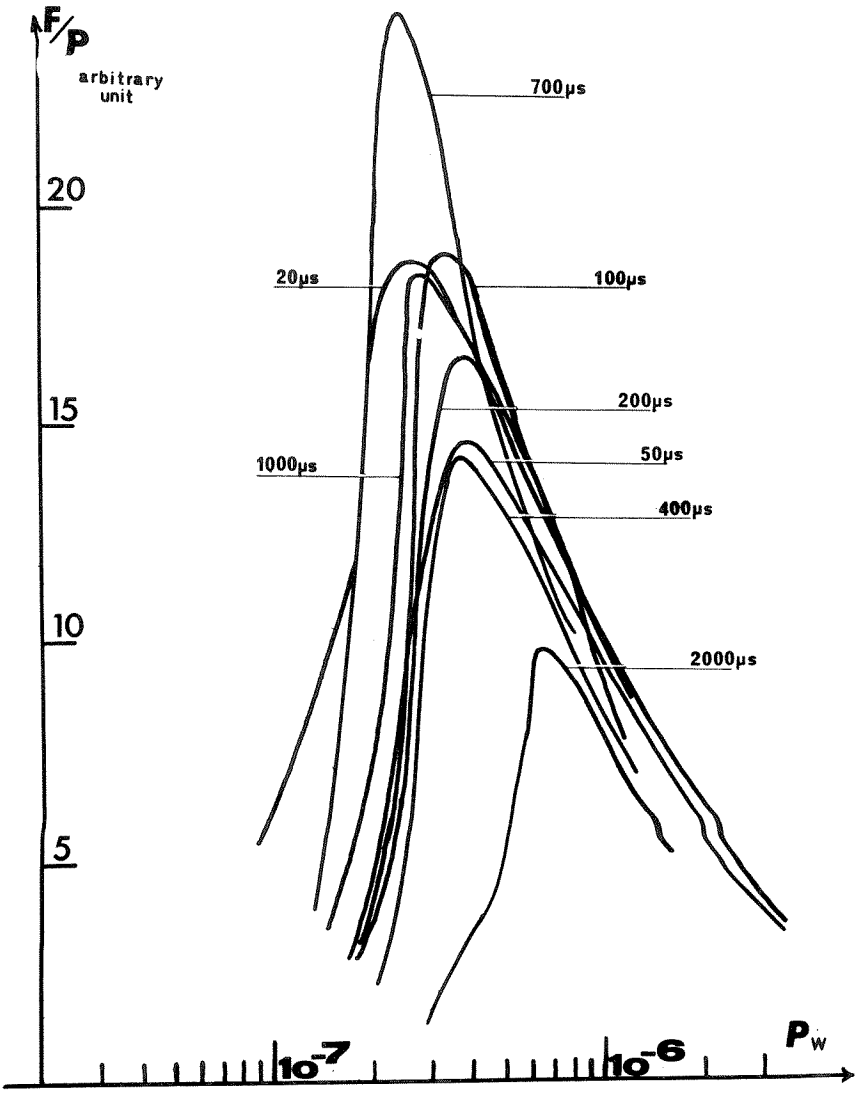

FIGURE 14.—F/P output as function of power of each impulse width.

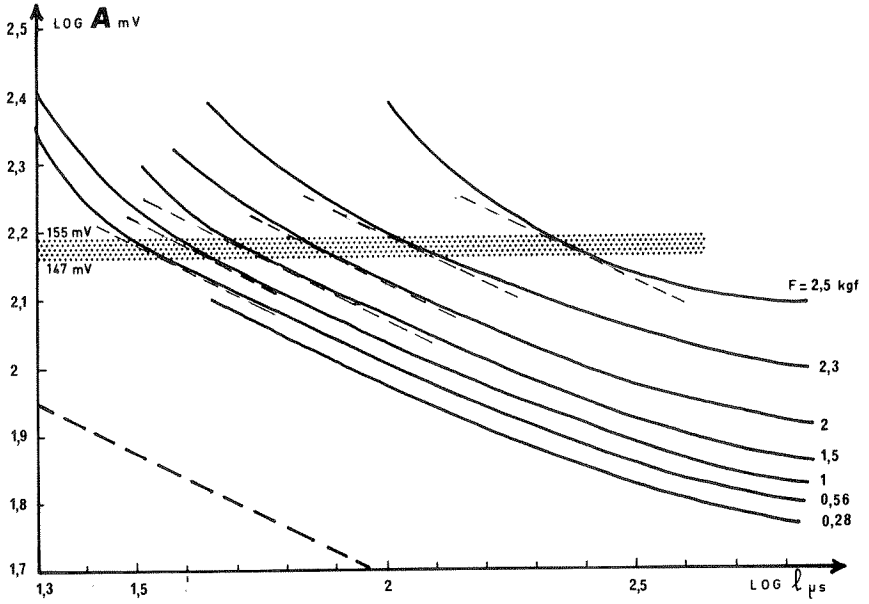

FIGURE 15.—Search for a control of triceps by minimization of stimulating power—A≈150 mV lies between 30 and 300 μs.

It is then found that by keeping the amplitude fixed (about 150 mV) and by increasing the pulse width by about 30 μs, it is possible to control the muscular force while simultaneously minimizing the consumption of energy. This procedure for optimal control of the muscular contraction is very easy to implement since it leaves two parameters fixed and requires only one parameter to change. It is shown as the shaded area corresponding to the zones where tangents to the curves of constant force are parallel (approximately) to the desired line for maximum power (this zone is at 147 – 155 mV).

B. Minimization of Response Time

The transient response of the muscle to a step application of the stimulating signal is characterized by the time separating the application of the stimulus and the appearance of the maximum force. This interval defines the "response time" of the muscle. If useful muscle control is wanted, the response time must not exceed a few tenths of a second. Obtaining short response time then is more important than minimizing the power.

In Figures 16, 17, and 18 the response time is given as a function of the developed force respectively for fixed frequency, fixed width, and fixed

amplitude. To obtain a certain value of force at a given frequency (Fig. 16) or given amplitude (Fig. 18) two response modes are possible: one exhibiting quick response (0.1 to 0.2 sec.), the other a slow response (up to 10 sec. or more). Figure 17 demonstrates this point by showing that slow responses are the result of short pulse widths and that fast responses are due to long pulse widths.

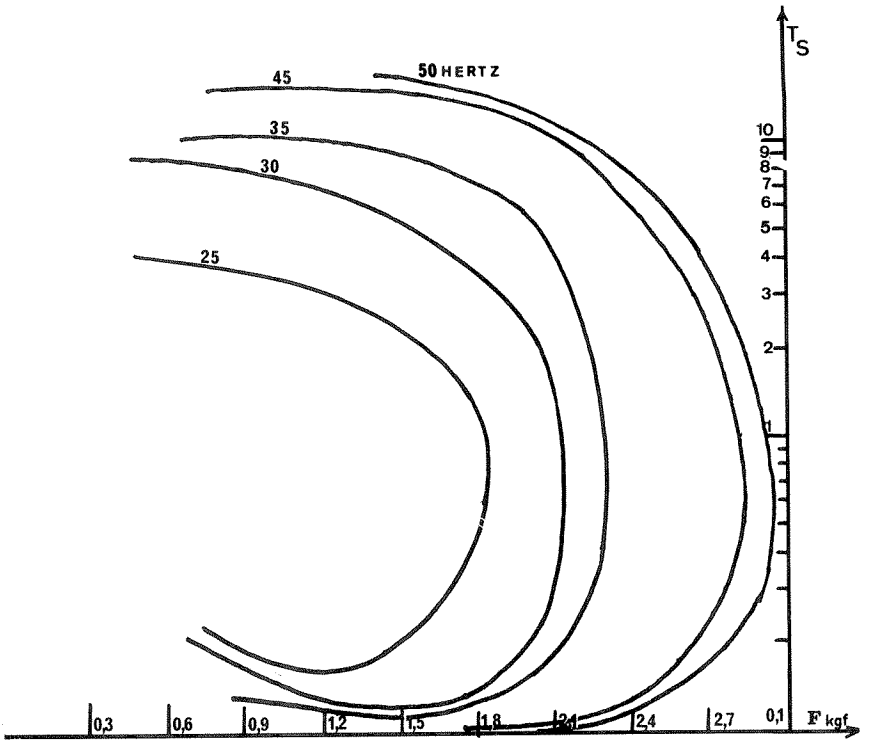

FIGURE 16.—Response time versus force with fixed frequency.

To reduce the response time it is necessary to work with large widths. This is the opposite of the previous criterion (energy minimization) which calls for a fixed amplitude and increasing width. To obtain reasonably quick response it is necessary to fix the width and to control the amplitude.

C. Criterion of the Maxima

It is possible to limit the problem still further by imposing the following restriction: that with the system of stimulation, the maximum muscular force desired can be obtained. As we have shown, the maximum force developed by a muscle varies with the stimulating parameters (see Fig. 7,

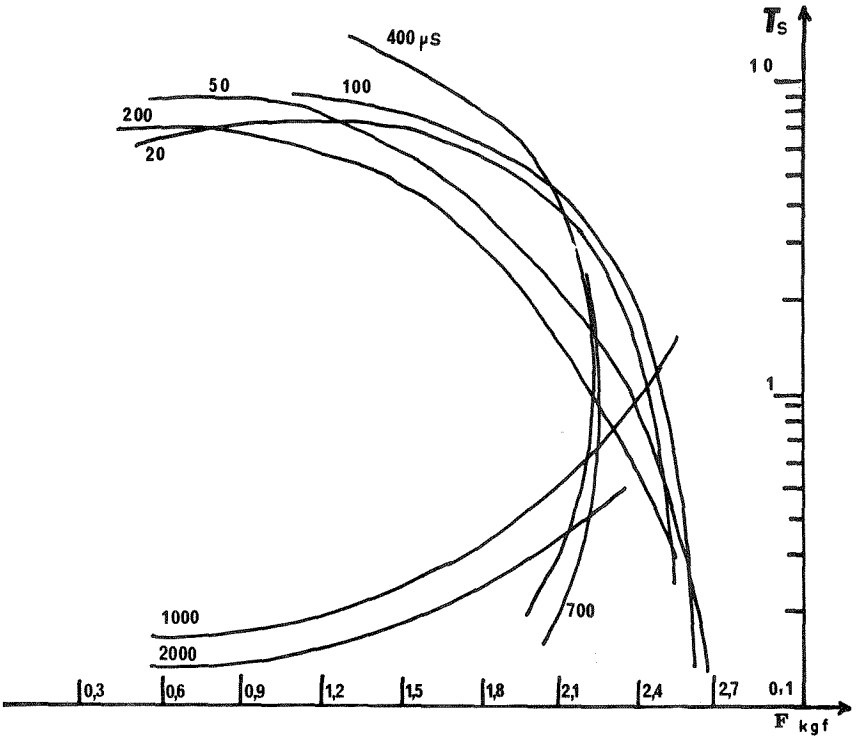

FIGURE 17.—Response time versus force with fixed impulse width.

8, and 9 of Part I). It was found that F max. increases slightly with the frequency but decreases as the pulse width increases.

The greatest muscle force obtained in all the experiments was with 20 μs pulses at a frequency of 50 Hz. This would fix the frequency and pulse width, and muscle control could be carried out by varying the amplitude. In practice it may not be necessary to develop muscle force to its utmost limits because muscle fatigue may develop more quickly than with another choice of excitation. Also, there is not a large difference between the maximum force developed with different types of stimulation, as the maxima in all cases are at least 75 percent of the largest value force obtained.

D. Summary

Let us summarize the results obtained during these experiments with isometric contractions:

a. The force developed by the muscle is almost independent of the excitation frequency provided that it is between 25 and 60 Hz. Operating in this field it is possible to ensure a reasonable muscular contraction.

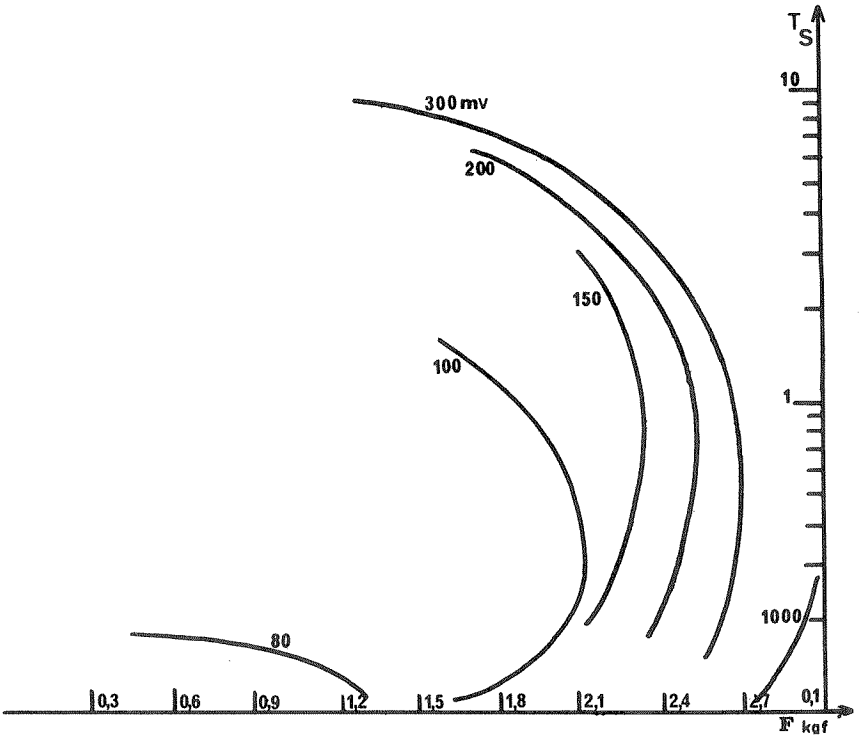

FIGURE 18.—Response time as function of force with fixed amplitude.

On the other hand, it depends strongly on the pulse amplitude and width.

b. Some intuitively reasonable criteria were applied in an attempt to limit the variables. It was found that criteria were not compatible with one another and for practical application the criterion of minimum response time was chosen to obtain a control model.

c. The experiments were carried on the tibialis anterior and triceps of 12 dogs. Except for physiological differences, the results were qualitatively identical in all the experiments. This uniformity will enable the control of the triceps or tibial contraction independently of the subject with the same stimulator. Individual differences may be compensated for by changing the amplitude or pulse width, both of which are easily adjustable.

CONTROL OF ISOTONIC CONTRACTION

In this case, the four parameters concerned are:

1. The angle of rotation of the joint.

2. The force developed by the paw in a given position.
3. The stiffness (or its reciprocal, the compliance).
4. The velocity of displacement or of rotation of the joint.

A. Summary of Experimental Results

As a function of the various stimulation parameters, output parameters were recorded and graphed. Figures 19 and 20 show the variation of the angle of rotation of the joint, caused by stimulating the triceps and tibialis anterior as a function of pulse width (for a frequency of 45 Hz) with several values of amplitude. Note particularly that the greater the amplitude, the more the variation of the angle with the pulse width. If the stimulating amplitude is inadequate, the maximum rotation may not be obtained, no matter what the width may be. The same type of reduction occurred in the isometric experiments.

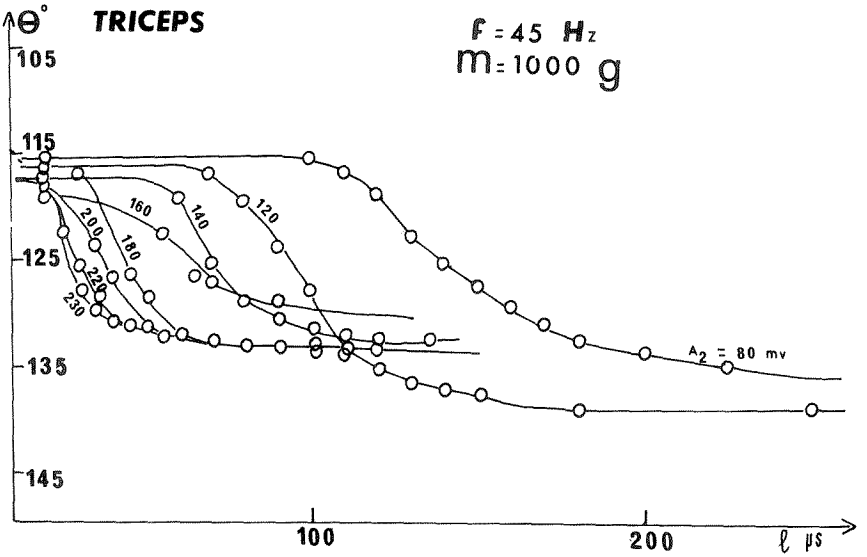

FIGURE 19.—Rotation angle of paw as function of triceps stimulating impulse width with fixed amplitude. The force to be pulled is 1 kgf.

As a general rule, the displacement of the foot needs the excitation of only one muscle; tibial is for flexion, triceps for extension. The unexcited muscle behaves passively but can oppose the rotation of the joint if its zone of elasticity (passive stretch) is situated too far from the zone of its length at rest. To obtain a given stiffness at the foot when it is placed at a given angle of rotation, both muscles must be stimulated. An increase of contraction of one must be compensated for by an increase of the contraction of the other; the foot does not move but the limb stiffens.

By acting manually on the output of the stimulator, the foot was brought to a predetermined position, in a time determined in advance and resisting a predetermined force (within certain limits), thus demonstrating that it is possible to completely control the paw in the open loop situation.

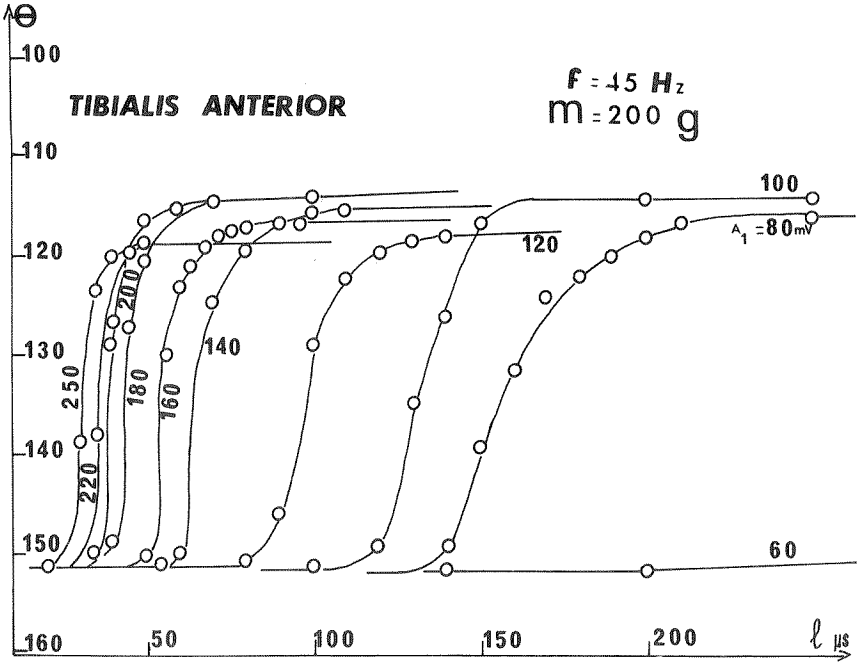

FIGURE 20.—Same experiment with tibialis, the force to be pulled being 200 g.

B. Modeling of Articular System

The external control of muscular contraction, and consequently, the joint rotation has been experimentally studied. It is possible to make a model of the foot which is operated by nervous stimulation of the tibial and triceps muscles. Figure 21 shows this model with a position control. Excited by the stimulator S_1, the muscle M_1 (for instance tibial) develops a force F_1 which is a function of its length D_1 and of the level of stimulation (l_1, A_1) of its nerve:

$$F_1 = F_1 \quad (D_1 \quad [l_1 \ , \ A_1]) \qquad [1]$$

Likewise

$$F_2 = F_2 \qquad (D_2 \ [l_2 \ , \ A_2]) \qquad [2]$$

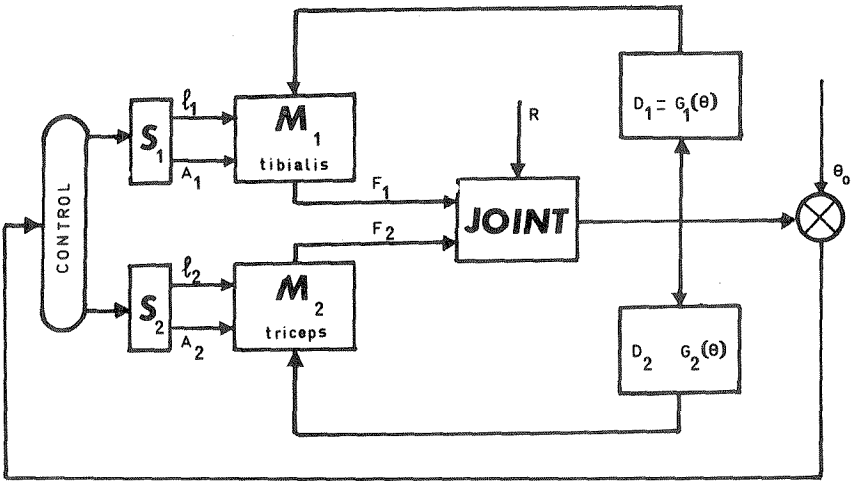

FIGURE 21.—Mechanical pattern of paw for position control: S_1—stimulation (A_1—impulses amplitude); l_1—width; M_1—flexor (tibialis); S_2—stimulator; M_2—extensor (triceps); F_1, F_2, forces developed by muscles on joint; R—exterior force; and θ—angular position of paw.

Both forces are exerted on the joint, as well as the exterior force R, so that the angle θ is a function of all three forces (and their respective lever arms).

$$\theta = \theta\,(F_1\,,\,F_2\,,\,R)$$

A mechanical relation binds D_1 , D_2 , and θ. To reach an angle θ_0 the difference $(\theta_0 - \theta)$ drives the stimulation.

So the mechanical function of the system (Fig. 22) is described by the following equations:

α. the force-length stimulation relations [1] and [2]

β. the geometric relations between the lengths of the two muscles:

$$g\,(D_1\,,\,D_2\,) = 0 \qquad [3]$$

γ) the equation of the movement:

$$I\,d^2\theta / dt^2 = M \qquad [4]$$

I: Moment of inertia of paw around joint axis

M: Moment of forces F_1 and F_2 and R in relation to joint axis.

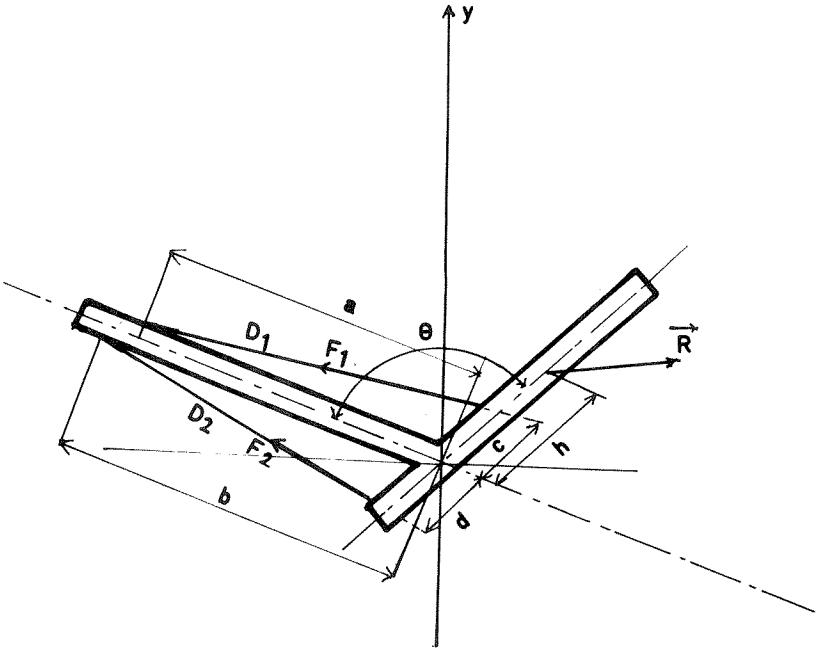

FIGURE 22.—Diagram representing the joint as it can be geometrically and mechanically described: D_1—tibial length; D_2—triceps length; F_1—force developed by the tibial; F_2—force developed by the triceps; R—exterior force; θ—angle of rotation of the foot; n—distance between the application point of R and the center of rotation of the ankle; a—distance between the insertion point of the tibial and the axis of rotation; b—distance between the insertion point of the triceps on the tibia and the axis of rotation; c—distance between the application point of F_1 and the axis of rotation; and d—distance between the application point of F_2 and the axis of rotation. The joint is supposed to be perfectly rotatable and free from friction.

With a knowledge of the lever arms, the paw mass, and the external force R, and measuring at each time F_1, F_2, and θ, it is possible to predict the operation of the system.

THE PROBLEM OF FATIGUE

Fatigue is the most important factor which can change the control

characteristic of the muscle. However, if the muscle is called on to generate a long sustained force, its power decreases progressively.

The time between the beginning of the stimulation and the appearance of the fatigue is a function of the force developed by the muscle (no fatigue under F max./5.0 (14)). It has been found that there is no fatigue during periods of up to 20 minutes, providing that the force developed by the triceps is about 0.5 Kgf ($\simeq$F max./5.5). Therefore, the maximum force behavior versus time has been studied.

Figure 23a shows a diagram of an experimental setup to test the triceps. The mass m is adjusted so that whatever the level of triceps stimulation the mobile part of the paw (OB) is not blocked. The levels of stimulation are adjusted to the minimum value consistent with the greatest value of angle AOB = θ.

θ is recorded versus time which is qualitatively the same thing as measuring the triceps maximum force. (Slow variations of static equilibrium are measured.)

Experimental results are shown in Figure 23a. Figure 23b shows a typical response.

a. Force F max. (equivalent to $\theta = \theta$ max.) can be seemingly held constant during about $t_1 = 20$ seconds.

b. For $t > t_1$ this force decreases regularly. In each case, it can be estimated up to 50 percent of the maximum force after 5 to 8 minutes.

c. The recuperation time, that is the amount of rest necessary for the muscle to obtain maximum force in two successive experiments, is about 15 minutes.

Figure 23c shows the reduction in muscle after the arterial and venous muscle circulation was stopped by clamping the blood vessels (p = clamping point).

When the nerve is clamped however, between the muscle and stimulation zone (Figure 23d, n = clamping point), the force (or the angle) decreases more rapidly than in Figure 23a, but follows the same type of curve.

The muscle developed force control, described in the previous paragraphs, can only be carried out if the maximum force is maintained for less than 20 seconds. This means that in practice, the research of the minimum developed forces suited to the desired result must be continually effected.

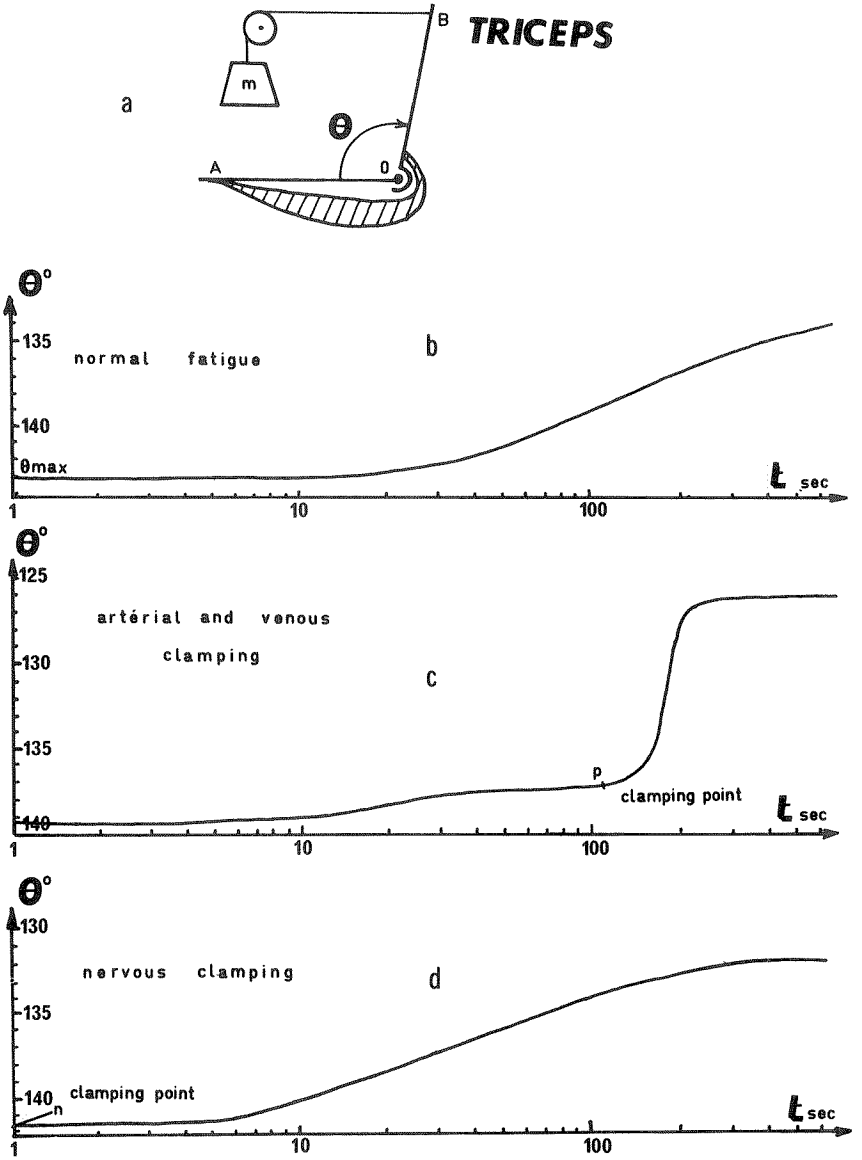

FIGURE 23.—a. Experimental setting for fatigue studies; b. θ versus time for maximal contractions; c. θ versus time for maximal contractions with arterial and venous clamping (clamping instant is shown); and d. θ versus time for maximal contractions with nervous clamping.

CONCLUSION

The experimental results summarized in this paper allow a better understanding of the joint motion control problem.

A given force can be developed (under isometric conditions) or the limb placed in a desired position (under isotonic conditions) by means of a simple electronic device.

The experiments described in this study lead to the following conclusions:

1. From a qualitative point of view, the problem of muscular control by external stimulation without fatigue seems to be solvable. Fatigue, resulting from the maintenance of a given force for an extended period of time, changes the relation between the stimulus and the developed force, and reduces the maximum amplitude of the developed force.
2. From a quantitative point of view, the parameters of the external nervous stimulation and their limits were determined for several experimental dogs.
3. With these results, adequate experimental stimulators and sensors were built which will allow more precise studies to be carried out.

The next logical phase of this work is to bring under control, by means of external feedback, either the amplitude of the force developed by skeletal lever, or its position and velocity. Thus, it is necessary to establish a mathematical model of the system formed by two stimulators controlling the agonist and antagonist muscles at a particular joint. This model, simulated on a hybrid calculator, will then permit the determination of a closed loop control system which will deliver a stimulating signal to the muscle to achieve the forces and positions commanded by a reference signal (2).

A study is being carried out on the form and materials required to chronically implant a stimulator. These developments will be reported in a later publication. Therefore, a simple control system without an external orthosis can be foreseen for the movement of a single joint even though the nervous system is not able to utilize the sensations of displacement or force. In later stages the sequential control of joints would offer the beginning of a solution to certain rehabilitation problems in paraplegic cases.

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