ELECTRICAL STIMULATION OF LIMBS

PART I.—BASIC STUDIES

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SUMMARY

This report is concerned with experimental work on the nervous stimulation of the agonist muscles governing the movement of a single joint of a dog's foot (ankle). Such stimulation will determine if it is possible to control smoothly the movement of the joint and the forces developed by these muscles.

Previous studies analyzed the muscle either at rest or during tetanic contraction. The data presented here are concerned with the mastery of muscular contraction. The ultimate goal is ability to produce artificially, after spinal cord injury, a rapid contraction to oppose exactly an external force or to attain a desired new position and to maintain that force or position without fatigue. The present paper reviews selected literature and describes electrodes and stimulators. It then reports certain studies on isometric contraction. Useful forces are found at frequencies of 30-50 Hz for application of rectangular pulses to the motor nerve leading to the anterior tibial muscle of an anesthetized dog. The influences of pulse amplitude (millivolts) and of pulse width (microseconds) on force are studied. For each there is a rather rapid but largely linear rise from threshold to saturation.

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INTRODUCTION

There has been considerable research on the contraction of striated muscle and its relations with the central nervous system. Hill (1) (1938) and Huxley (2) (1957) carried out fundamental studies to obtain better knowledge of the phenomena and ultimate structures of muscle in order to explain its behavior during a voluntary contraction—altogether three classes of muscle models have thus been proposed:

1. The visco-elastic model: Voigt (3), Levin and Wyman (4) (1927), Hill (1) (1938), Pringle (5) (1961), Caspi (6) (1969), Pertuzon (7) (1972), and many others, describe characteristics that are really valid only during tetanization. They do not explain the transient phenomena.
2. The calorimetric model of Hill (1) (1938) relates the coefficient of "muscle load-reducing velocity" to the measurement of heat exchange in the muscle. But these results are difficult to use.
3. The physico-chemical model of Polissar (8) (1952), Goodall (9) (1957), Huxley (2) (1957), with which it is difficult to give an explanation of the muscle mechanical properties.

The objective of all these different models is to explain muscle behavior during voluntary contraction. This report describes a different approach to the problem. We use a normal muscle-nerve unit (without any physiological, physical, or chemical lesion) in which the possibility of voluntary contraction has been suppressed by cutting the spinal cord. The nerve root exiting near the spine or the peripheral nerve near the muscle is stimulated electrically to make the muscle contract. The results are intended to determine if it is possible to develop a model of a muscle adapted to optimal control, that will lead to artificially controlled movement in a paraplegic patient.

METHOD OF APPROACH

Means of Muscle Excitation

A muscle will only contract when it is exposed to a certain stimulation. This may be of chemical, thermal, mechanical, or electrical origin. The signals naturally transmitted by the nerve and motor end plate are of the electrical kind. Variation of the distribution of ions (or charges) and the resulting field potentials propagate along the nerve. The easiest method to alter this distribution of ions is to use electrical stimulation.

Three solutions to the electrical stimulation problem are commonly used. With external stimulation, signals can be transmitted to the muscle by electrodes applied on the skin (3). This method, however, has three main disadvantages:

1. It is necessary to have a good external spatial location of the muscle to be excited.

2. There is a risk of exciting several muscles at the same time (parasitic excitations of nonselected muscles).
3. High potential drive signals are necessary.

Direct stimulation of the muscle, by placing electrodes in direct contact with the muscular mass itself, also has three main disadvantages:
1. The electrical signal strength required for a good contraction is relatively high.
2. The reproducibility of the phenomenon mainly depends on the position of the electrodes.
3. There is a risk of very quickly exceeding the fatigue threshold of some muscular fibers.

Direct stimulation of the nerve (10) is very tempting since the nerve/motor-end-plate/muscle unit works as a strong energy amplifier with a gain that may reach 10^6. The propagation of a signal along a nerve is analogous to the propagation of a signal by an electric line; a difference in potential (due to an unbalanced distribution of the charges in the nerve fiber) spreads along the nerve with a given velocity. The phenomenon is not sensitive to the location of the stimulating electrodes on the nerve, but only to their separation. Because of the low intensity stimulation required, the stimulator may be small, and therefore its chronic implantation is feasible.[b]

Nervous stimulation can be carried out at several levels. Stimulation on the muscle nerve directly poses a problem because there is an afferent proprioceptive sensitive nerve contingent. During stimulation, signals from these sensor fibers may create inopportune reflex contractions, as well as possible painful sensations in a patient, with an incomplete lesion of the spinal cord. In addition, stimulating on the common nerve trunk may excite several muscles at the same time.

Stimulation at the level of the nerve root exiting from the spine seems to be the best general solution to the problem, since the ventral nerve root leaving the spine has only motor fibers and to each radicle corresponds a definite muscle (or a very small group of muscles). In spite of surgical and experimental difficulties concerned with the determination of radicular topography, such a stimulating technique may be considered in the future.

In the experiments reported here, we stimulated the muscle nerve directly since it is the easiest to use and since during experiments carried out under anesthesia, we are not impeded by the secondary firing of sensor nerve fibers.

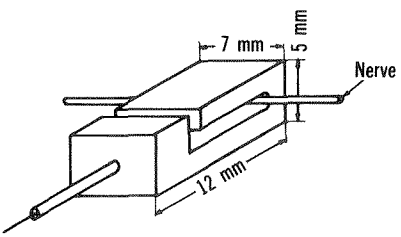
[b]In this article, we will not enter upon the implantation problems nor upon the chronic stimulation (nervous or muscular). Researchers such as J. B. Reswick, Rancho Los Amigos Hospital, Los Angeles, Calif., have already solved them in particular cases.

Choice of Stimulating Electrodes

When a nerve is stimulated with metallic electrodes, a cell effect, due to the mobility of the ions in an electrolytic medium, can be seen. This polarization appears very quickly, either preventing any nerve stimulation or making the muscle contraction phenomena unreproducible. Polarization is annoying since it alters the signals we want to send to the motor-end-plate. Corrosion soon appears at the electrodes (even if they are made of noble metal) (11).

In order to circumvent these problems, we use stimulating electrodes of pure carbon fiber. The advantage of this material is that it is a relatively good conductor and nonpolarizable. Moreover, it is well tolerated by the system, judging by the attempts carried out by P. Rabischong and his colleagues on dogs over periods of several months. This material, manufactured in the form of fabrics (woven threads) as well as casting, has given very satisfying results.

We have developed a stimulating technique using two woven carbon fiber loops which we very loosely ligate around the nerve. To avoid spreading of exciting signals in the noninsulating tissues near the nerve, a silicon insulator surrounds the ligatures (Fig. 1). This system was successful because with the stimulating amplitudes we use there is no spreading detectable in the neighboring tissues.



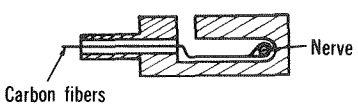


FIGURE 1.—Diagram of a stimulating electrode.

Electrical Stimulator

Let us approach now the problem of the form of the stimulating signal

which we want to send into the muscle nerve in order to initiate a muscle contraction. We know that to control a voluntary muscle contraction, the motor nerve transmits a signal composed of frequency modulated impulses. We therefore built a low output impedance transistor stimulator which delivers rectangular impulses[c] whose amplitude, width, and frequency can be regulated independently. Analog signals proportional to these three parameters are available for recording.

Two versions based on the block diagram shown in Figure 2 were developed.[d] One is a laboratory stimulator directly integrated with a hybrid computer. The other is a portable version (120 mm. × 80 mm. × 20 mm.) that runs with batteries. It is possible to modulate separately the amplitude, width, and frequency of the pulses, sinusoidally at a variable low frequency.

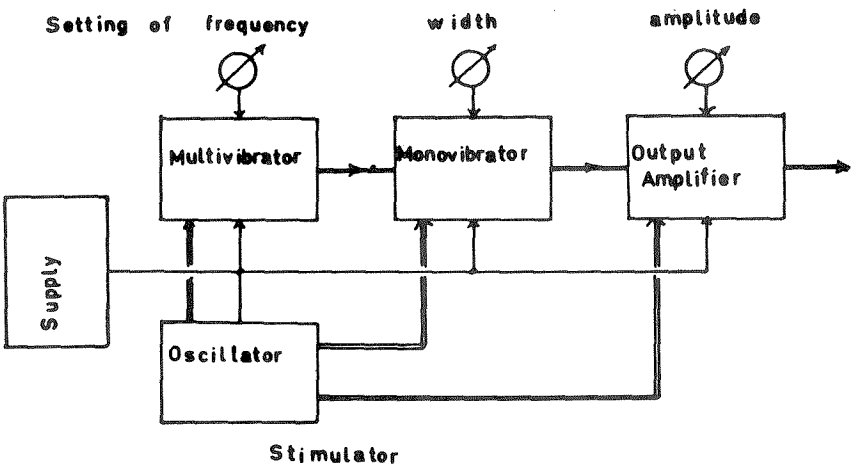


FIGURE 2.—Diagram of a stimulator.

Muscle Response Measurements

We studied the muscle response by measuring displacements and strains to which a limb is submitted by a single pair of muscles (agonist and antagonist). In the present state of our study, we were mainly interested in three parameters:

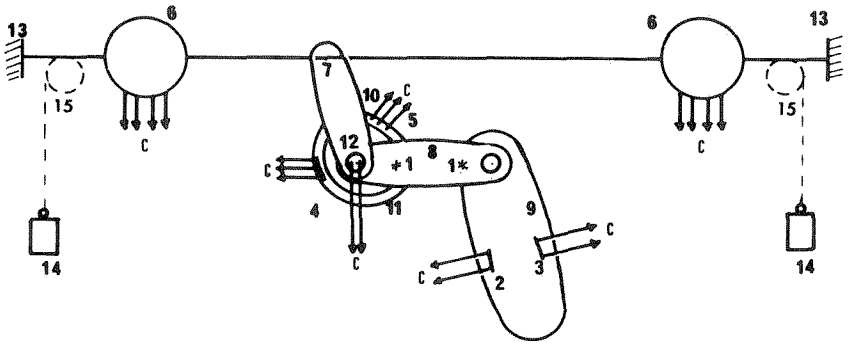
1. The force (or torque) developed by the limb.
2. The angle of rotation of the limb.

[c]We use rectangular-shaped impulses, but considering the present state of our study, we cannot say whether this is the optimal shape.

[d]The stimulators generate rectangular impulses with a rise time of less than 1 microsecond (μs), at a rate between 0.1 and 60 Hz, and pulse width between 10 and 2000 μs. The amplitude is between 0 and 4V or up to 80V with an external stimulator; usual values in these experiments were 60 to 500 millivolts (mV).

3. The force exerted by the tendon of each muscle.

The experiments were carried out on the paws of dogs. The agonist muscle is the anterior tibialis and the antagonist is the triceps surae. These muscles control the rotation of the ankle. The plan of the experiment is given in Figure 3.



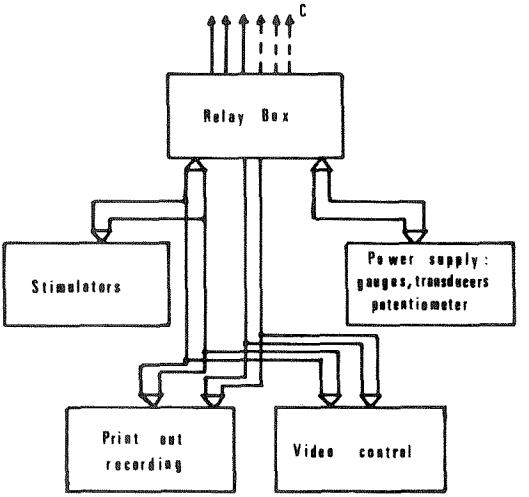


FIGURE 3.—Synoptic diagram of the experimental setting: 1. orthopedic pins through the tibia, 2. stimulating electrodes on the external popliteal sciatic nerve (tibial control), 3. stimulating electrodes on the internal popliteal sciatic nerve (triceps nerve), 4. JFET transducer included in the triceps tendon, 5. JFET transducer included in the anterior tibial tendon, 6. metallic gauges bridges, 7. foot, 8. calf, 9. thigh, 10. flexor M1—anterior tibial, 11. extensor M2—triceps, 12. potentiometer enabling measurement of θ, 13. attachment for isometric tests, and 14. setting with weight for isotonic tests.

The force developed by the limb is directly measured with a ring on which are attached four bridge-connected metallic strain gauges. The working principle is given in Figure 4. A voltage varying linearly with force F, reported in kilograms force (Kgf), is obtained.

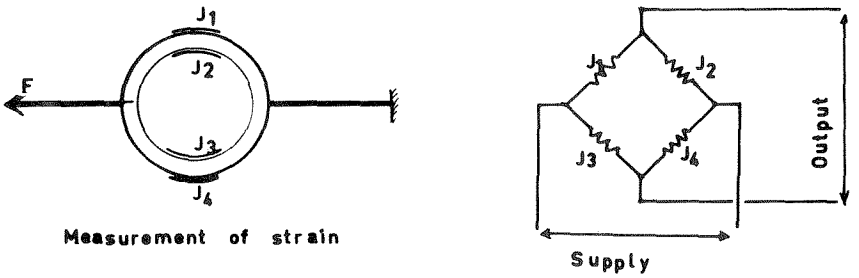


FIGURE 4.—Diagram of a metallic gauges transducer.

The force developed by a muscle may be measured by tendinous transducers: there are semiconductor gauges (JFET) with a volume near 0.2 mm^3, surrounded by an insulating material and bound up in the tendon by insertion inside it. The magnitude of the voltage is approximately proportional to the muscle force. (The first attempts were carried out by P. Rabischong et al., 1965 (12), on tendons, with transducers of another nature.) The difficulties now lie in the testing and mainly the method of attaching these transducers which must not irreversibly damage the tendon.

The angle of rotation of the limb is measured by a potentiometric system with negligible friction which gives the joint angle independent of the position of its apex and the plane of rotation.

Experimental Procedure

After deep general anesthesia of a selected dog, the surgeon isolates the internal and external popliteal sciatic nerve or a hind leg and fixes the stimulating electrodes and the insulators. He inserts the JFET transducers in the anterior tibialis and the triceps surae tendons. After ascertaining that the muscle can be stimulated and that the tendinous transducers are functioning, the wounds are resewn. In order to maintain the calf in a firm position, two orthopedic pins are fixed through the tibia and secured in a metallic frame fastened to the operating table. If isometric measurements are made, the foot is maintained at a fixed angle with regard to the shank and two rings with bridges of metallic strain gauges measure the net force to which the paw is submitted. If the isotonic pattern is to be studied, weights (Fig. 3, no. 14) are attached to cords over pulleys (Fig. 3, no. 15), the foot can freely turn around the

ankle, and its movements are measured with the potentiometric sensor. All the input and output signals can be checked by viewing on an oscilloscope and can be recorded.

RESULTS WITH ISOMETRIC CONTRACTIONS

In order to determine whether a model of muscle associated with an optimal nervous control is possible, we had to determine the influence of the different stimulating parameters on the contraction of the muscle *in situ* with both isometric and isotonic conditions. The present paper reports only isometric experiments.

Experiments with Isometric Contraction

The location of the foot with regard to the shank is fixed so as to keep the length of the muscles constant. One of the muscles is stimulated using impulses of different frequency, amplitude, and width.

1. Influence of The Pulse Frequency

With a pulse amplitude strong enough to affect the muscle (and a fixed pulse width) a force is developed which depends on the frequency. For the very low frequencies, the muscle responds in a one-to-one relationship, but the average value of the force developed rises with frequency. At a pulse repetition rate of about 20-25 Hz, an "integrated" response appears. That is to say, the muscle has no time to slacken between impulses, the inertia of the limb segment exerts a smoothing effect, and a continuous force is obtained (Fig. 5, force is expressed in Kgf, 0.1 Kgf = 1 Newton). Beyond 25 Hz, the force seems to reach saturation and it does not increase significantly even at 50 and 60 Hz.

These results with varying frequency indicate that the static force developed by the muscle cannot be controlled in a useful way by variation of the stimulating frequency. Indeed, in the zone where force rises with frequency, we obtain a one-to-one response from a single pair of electrodes, and in the zone where the force seems to be continuous, it depends only slightly on frequency.

2. Influence of Pulse Amplitude

With a stimulating frequency between 30 and 60 Hz and a fixed pulse width, the force generated by the muscle is a function of pulse amplitude, as shown in Figure 6. As the amplitude rises, the force developed quickly increases from threshold and reaches saturation. Increasing the amplitude still further causes a decrease of the net force due to the parasitic excitation by diffusion of the antagonistic muscle. In the case shown, the useful range is 200-300 mV.

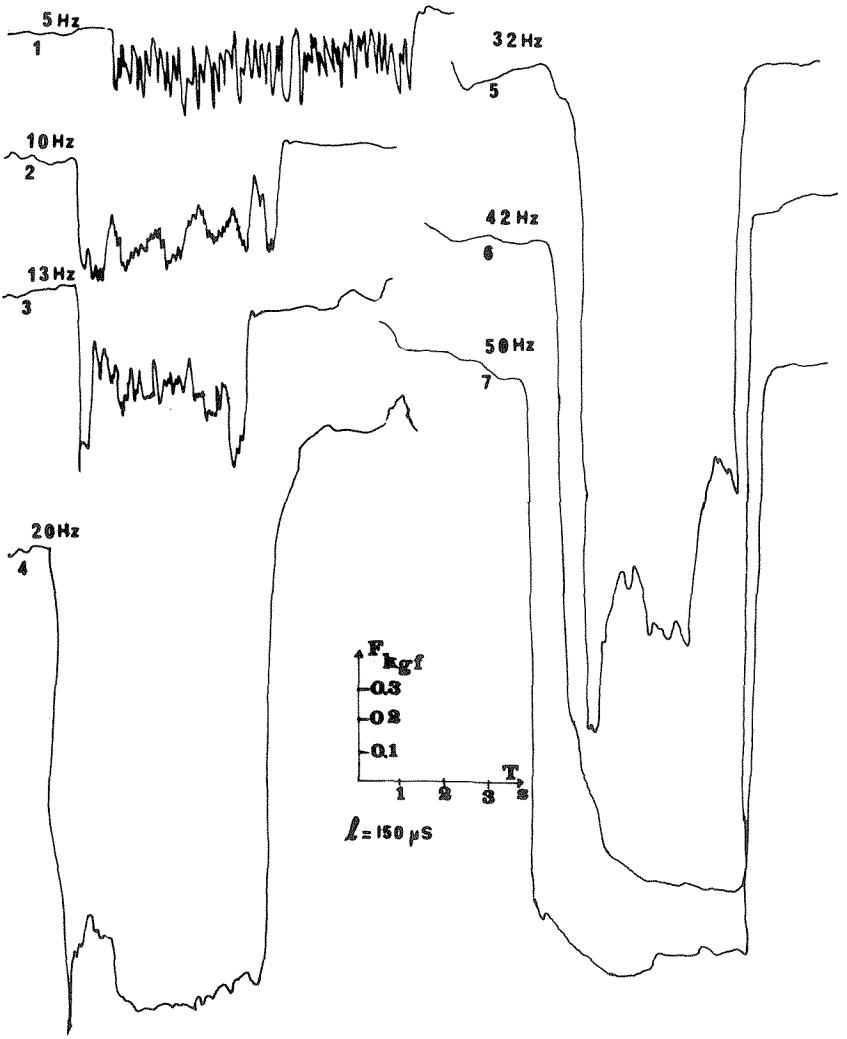


FIGURE 5.—Recordings of the tensile force developed by the triceps surae. Tibial nerve excitation: fixed amplitude; fixed pulse width (150 μs); frequency: (1) 5 Hz, (2) 10 Hz, (3) 13 Hz, (4) 20 Hz, (5) 32 Hz, (6) 42 Hz, (7) 50 Hz. Tensile force measured by pure crystal silicon transducer or MOS (metal oxide semiconductor) transducer.

These results show the existence of an upper limit for the stimulating amplitude, a threshold which must not be exceeded so as not to work in the saturation zone. There is typically a long linear portion.

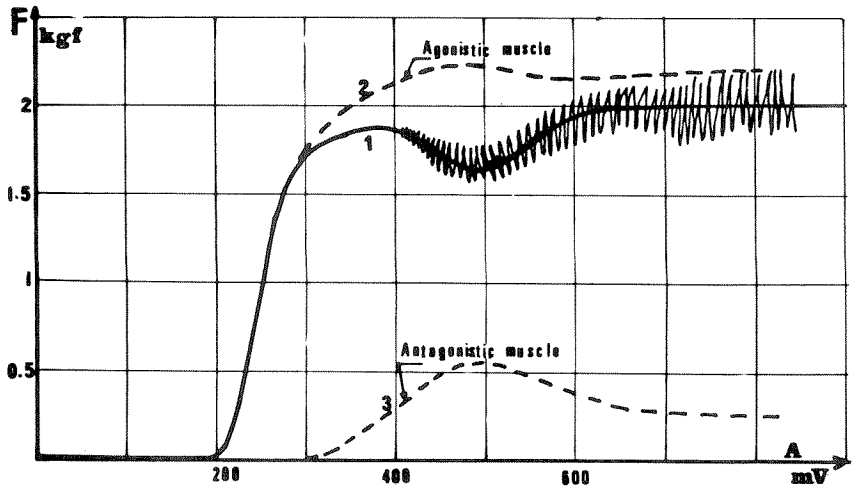


FIGURE 6.—Variation of the force developed with the stimulating amplitude: 1. Measured force—excitation of the tibial nerve, 2. actual force developed by the tibial, and 3. force developed by the triceps (consequence of the diffusion of the stimulating signal when amplitude becomes too high). On the Y axis is the signal given by the metallic gauges bridge. The signal is directly proportional to the force developed.

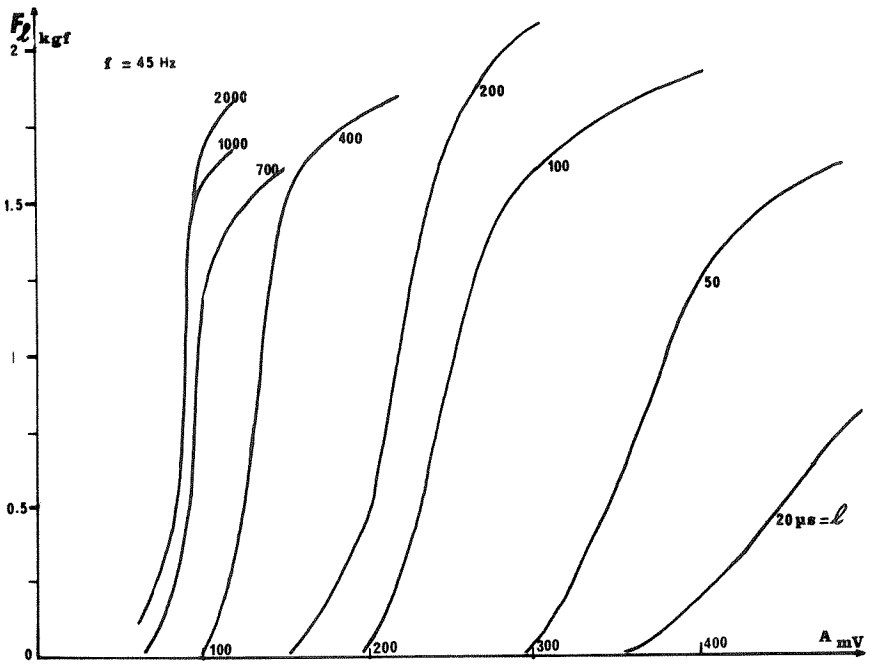


FIGURE 7.—Force developed by the tibial as a function of stimulating amplitude for fixed widths.

3. Influence of Pulse Width

Now we must determine whether both previous limits of useful force depend on the pulse width. The force-amplitude curve is repeated for a great variety of conditions of frequency and width (e.g., Fig. 7). We can then draw (Fig. 8 and 9), the zones where stimulation may be regarded as adequate for the dog's anterior tibialis.

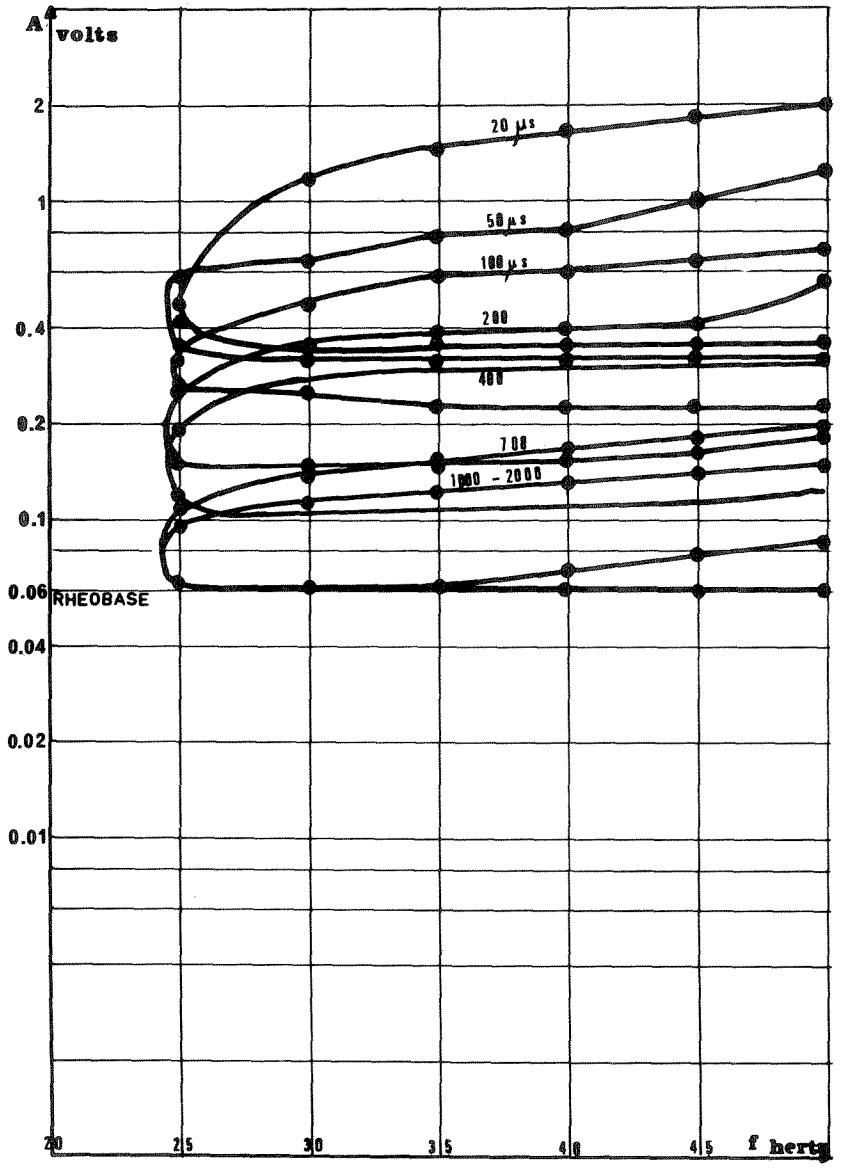


FIGURE 8.—Zones of acceptable stimulation of the muscular nerve in the amplitude-frequency impulses width space.

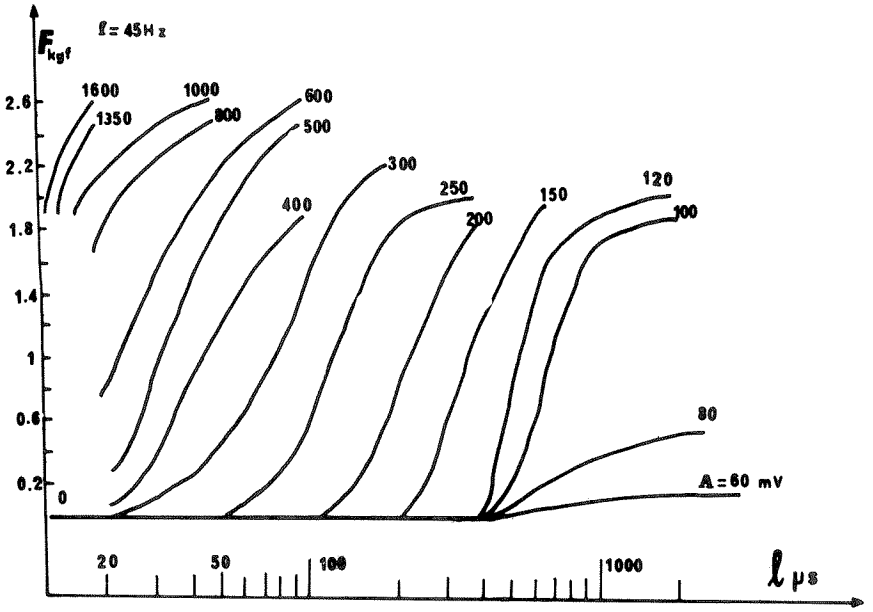


FIGURE 9.—Force developed by the tibialis as a function of the width of the stimulating impulses, for fixed amplitudes.

The study of these curves indicates that:

1. The zones are all larger with small pulse width.
2. The maximum force is obtained for an amplitude which varies inversely with width.
3. With pulses less than a millisecond in width, force seems to be independent of the width.
4. The rheobase is approximately 60 mV.

The chronaxy is about 400 μs. These values for striated muscle are in agreement with those provided by various authors (13).

Search for Isometric Muscle Control

Let us sum up in a mathematical way the constraints and laws to which the control of the muscular contraction is imposed. Let f, A, l, be the frequency, the amplitude, and the width of the stimulating impulses. F represents the force developed by the stimulated limb. Then with a constant frequency, f, lying between 30 Hz and 60 Hz:

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: $q = \int i \cdot dt$.