SIMULATION OF PURPOSEFUL MOVEMENTS BY ELECTRICAL STIMULATION OF MUSCLES

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Indroduction

In general, patients with central or peripheral lesions of their motor pathways are unable to perform useful movements with the corresponding extremity. In cases where the reflex loop is still intact (central lesions) and if spasticity is negligible, useful movements can be achieved by electrical stimulation of the muscle motor points. The present paper deals with some aspects of the problem of how to stimulate the muscles in order to move the extremity similarly as when normally innervated. The investigations were performed under the following limitations:

1. Only rotation of the lower arm about the elbow joint was considered. The experimental technique was essentially the same as used by Vodovnik and Crochetiere[1,2].

2. From many possible purposeful movements three frequent types were chosen: withdrawal, slow tracking, and minimum time control.

3. Physiological evidence shows that there is no essential difference in the excitability of muscle of normal subjects and of patients with central lesions but not suffering from spasticity. To avoid experiments on patients as long as possible, the results described below were obtained on normal subjects told to have their arm completely relaxed. In the next stage of the project comparative measurements will be performed on patients.

An analysis of problems encountered in achieving simulation of the chosen three types of movements is given in the following paragraphs.

Withdrawal Movement

To obtain data of withdrawal movements the subject was told to move his lower arm after a combined acoustical and optical stimu-

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lus as fast as possible from extension to full flexion. In Figure 1 curve (a) shows a typical result. It is hypothesized that for such a motion the command from the CNS is »maximum contraction of flexor muscles«. Automatically inhibition of the antagonists sets in, but no feedback is necessary and the movement could be performed by an open-loop system. This type of movement was simulated by a step function of tetanizing voltage applied over the motor point of the biceps, and a typical response is shown in Figure 1, curve (b). It can be seen that curve (b) simulates curve (a) rather well. The maximum velocities are about the same and, in general, agree with data obtained in Ref. 1. The dead time of curve (a) is much longer than the delay of curve (b), but this is not essential as the commands to perform movements were given in both types of experiments quite differently, and the delay in (a) is mainly due to reception, processing, and transmission of nerve pulses. In a reflex withdrawal movement (e. g., by painful skin stimulation) the delay would be much shorter.

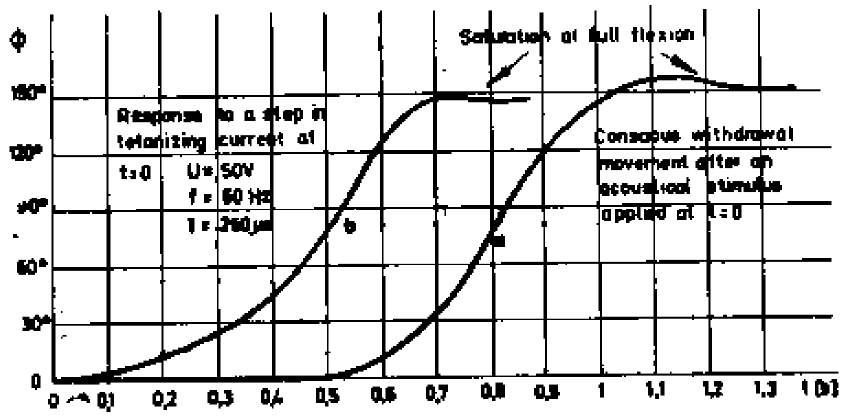

Figure 1.

The transfer function between angle Φ and amplitude of stimulation current I is of the type

$$\frac{\Phi(s)}{I(s)} = K \frac{e^{-Ts}(s+a)}{s^n (s+b)(s+c)} \quad \ldots\ldots\ldots \quad 1$$

with $n_{max} = 2$. A rather useful approximation is reached[1] with $n = 0$, $b = 0$, and $(s + a) = 1$. However, many times linearisations in biological systems are risky, and the behaviour of a musculoskeletal system under electrical stimulation can be represented realistically only by nonlinear methods. The major nonlinearities are: threshold and saturation in stimulation, position dependent reaction torques at end points (full flexion or extension), decrease of the delay T with increased stimulation amplitude, and position dependent influences of the reflex loop. Besides, hysteresis effects were observed between stimulation currents and muscle force[4], and torques due to stimulation were found to be position dependent[3] (moving of motor point).

Slow Tracking

In normal tracking, both antagonistic muscles are reciprocally activated, and γ-innervation, proprioceptive feedback, visual feedback, prediction and memory effects in the CNS form an extremely complex control mechanism. Therefore any attempt to simulate such movements by crude electronic means seems to be too ambitious, by far. During the experiments these thoughts were often confirmed, our ignorance of biological systems being the main obstacle in the design of electronic controllers. But as the ability to control the movement of one joint of a paralyzed extremity is of vital importance for any further work in functional stimulation of extremities, investigations have to be continued at least until we have been able to prove definitely that joints controlled by electrical stimulation cannot be clinically useful. However, for the time being we are still optimists.

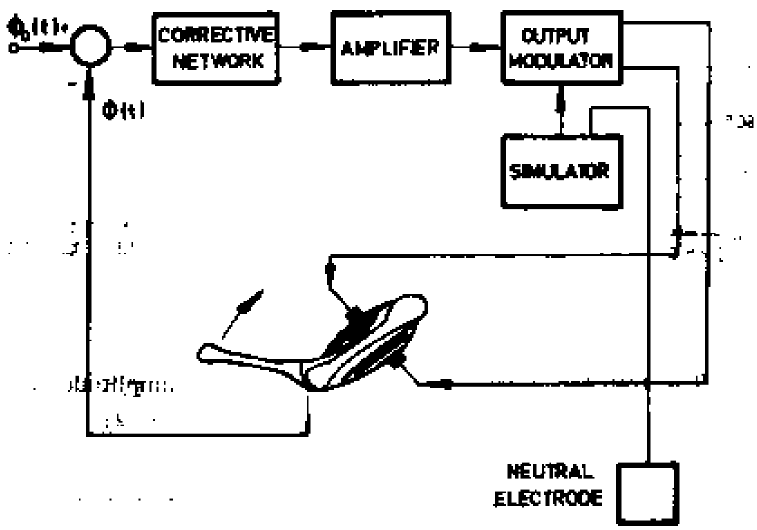

Figure 2.

The control system used is essentially the same as desribed in Ref. 2. Some modifications and improvements were made in the electronic circuits and in the experimental technique (e. g., ball-bearings at the elbow and hand). The block diagram of the system is seen in Figure 2. When the loop was closed without the corrective network and the error was made zero before switching on the stimulation, the arm remained stable due to the dead-zone. After a change in position the arm began to oscillate, a typical example thereof being shown in Figure 3. It is interesting to note that for the majority of our subjects the frequency of oscillation was above 1.5 Hz in contrast to the findings in Ref. 2 where the frequency was always about 1 Hz. Before

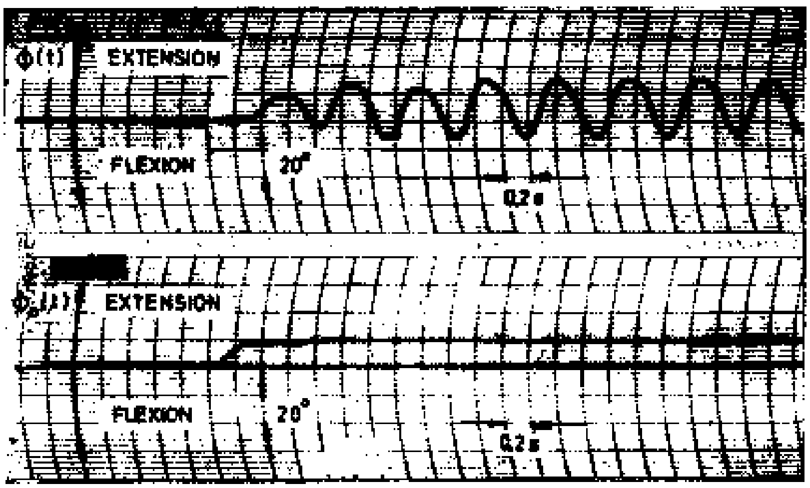

Figure 3.

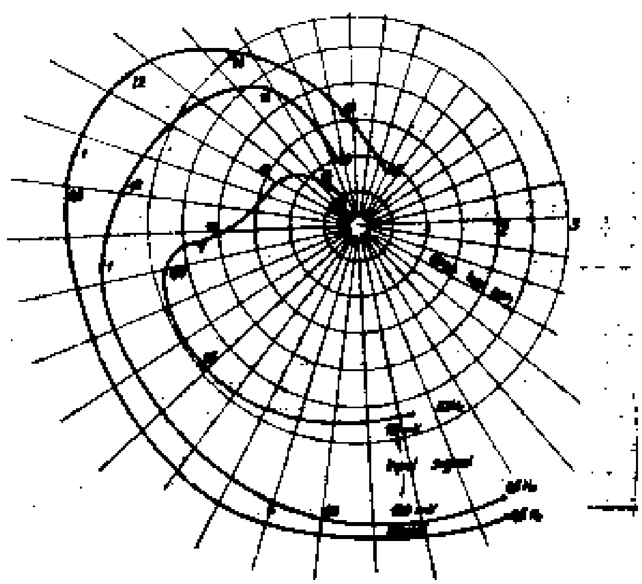

Figure 4.

attemps to stabilize the loop were made, input-amplitude dependent Nyquist diagrams were measured (Fig. 4). By means of these diagrams the nonlinearity and the transfer function of the system can be determined[9]. The diagrams measured on two subjects showed that the transfer function is of at least the third order; that there must be a dead-zone and saturation nonlinearity; and that memory effects (hysteresis) are present as the describing function has a complex character. In short, the method applied in our experiment was the following. The sinusoidal output was measured at n frequencies and m different input amplitudes. The results were of the form $M_i e^{-j\alpha_i}$ Therefore

$$\left.\begin{aligned}
M_{11}\cdot e^{-j\alpha_{11}} &= R_1 r_1 e^{-j(\Phi_1+\varphi_1)}\\
M_{12}\cdot e^{-j\alpha_{12}} &= R_1 r_2 e^{-j(\Phi_1+\varphi_2)}\\
&\vdots\\
M_{1m}\, e^{-j\alpha_{1m}} &= R_1 r_m e^{-j(\Phi_1+\varphi_m)}\\
M_{21}\, e^{-j\alpha_{21}} &= R_2 r_1 e^{-j(\Phi_2+\varphi_1)}\\
&\vdots\\
M_{nm}\, e^{-j\alpha_{nm}} &= R_n r_m e^{-j(\Phi_n+\varphi_m)}
\end{aligned}\right\} \quad \ldots\ldots\ldots \quad 2)$$

where. R_i, Φ_i are the absolute value and phase of the transfer function, and r_i, φ_i being the absolute value and phase of the describing function. Taking points with constant ω_i, the magnitude and angle of the describing function are obtained, and by means of the inverse describing function[7] or by mere inspection of existing tabulated describing functions, the nonlinearity can be deduced. If points with a constant input amplitude are taken at different frequencies,

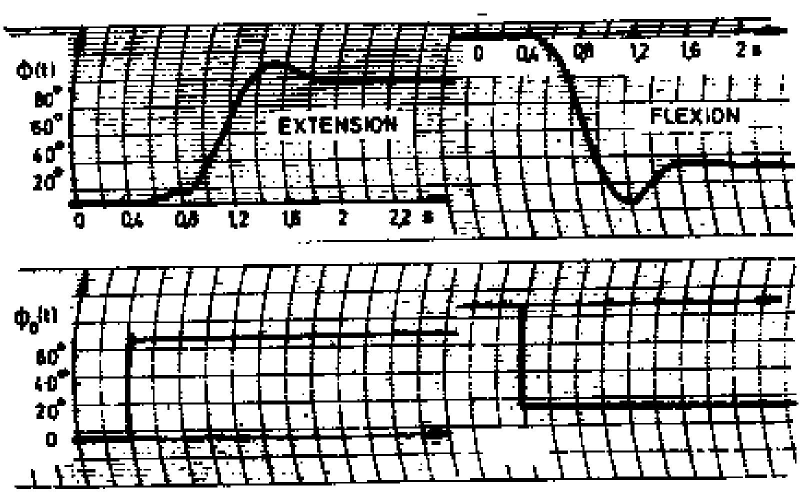

Figure 3.

the conventional Nyquist diagram is obtained. Measurements of n frequencies at m amplitudes are theoretically redundant if frequency-independent nonlinearities may be expected, and under this condition instead of $n \cdot m$ only $n + m$ measurements are required. However, for checking purposes and to get a better insight, all $n \cdot m$ points were measured in our experiments. Many inconsistencies were found upon evaluation of the curves, and some experiments gave quite nonconclusive results. For one subject a few output frequencies were lower than the input frequencies. We presume that at these frequencies a »spinal oscillator« took over the command to move the arm, or that nonlinearities caused subharmonic oscillations, but no further investigations were made in this direction.

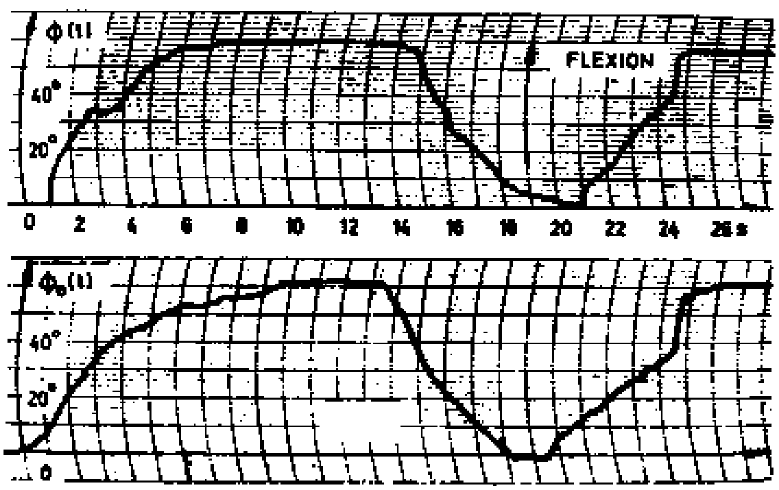

Figure 6.

As our knowledge regarding nonlinearities in musculoskeletal systems is still incomplete, as a first approximation amplitude-independent correcting networks were inserted to stabilize the closed loop. Lag-lead and twin T networks were tried, and the results are satisfactory, though by far not as good as when the extremity is under normal innervation. Figure 5 shows responses to step inputs in position, and Figure 6 gives an example of slow arbitrary positioning. In both figures the loop was stabilized by means of a twin T filter with a resonant frequency of about 1.5 Hz.

Minimum-Time Movements

In this type of movement the initial and final positions of the limb are known. The nervous system or stimulating mechanism must operate the antagonistic muscles in such a way that the movement is performed in the shortest time possible.

Under normal innervation it is generally believed that quick, directed movements are achieved by an »... alternation of contractions of agonist and antagonist muscles ...« in the end phase of the movement[5]. Such a viewpoint seems to be in agreement with theoretical considerations regarding time-optimality. Bellman[6] and others namely showed that for systems which can be described with the vector-matrix differential equation

$$\dot{x}(t) = A(t)\,x(t) + D(t)\,m(t) \qquad 3$$

the time optimum controller is of the bang-bang type, i.e. the control vector $m(t)$ components always have their maximum possible values $\pm M_i$. In eq. 3, x is the state vector, A is the coefficient matrix of the process, and D the driving matrix. In the case of control when muscles are used as motors, the components of $m(t)$ can switch only between 0 and $\pm M_i$ as muscles can produce active force only in one direction. In a simple musculoskeletal system only two muscles are activated. The agonist switches from 0 to $+M_1$, and the antagonist changes from 0 to $-M_2$.

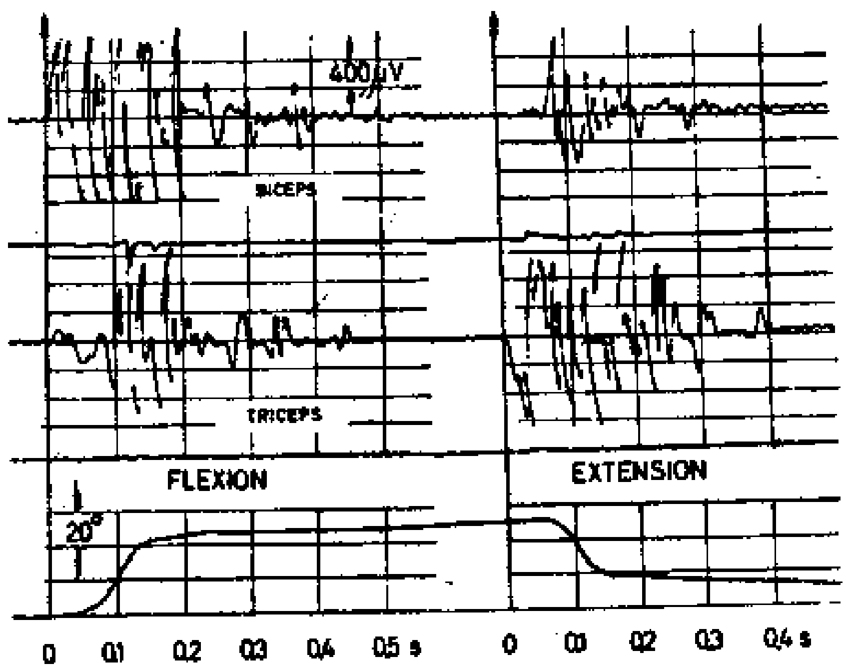

Figure 7.

Experiments with forearm movements did not confirm the findings of Partridge[5]. In quick, horizontal movements from one position to another, we regularly found that in the vicinity of zero error there is a time interval where both antagonistic muscles are active (Fig. 7). This simultaneous activity of antagonistic muscles (SAAM) is uneconomical from the viewpoint of energy consumption, Under normal innervation it is generally believed that quick, directed movements are achieved by an »... alternation of contractions of agonist and antagonist muscles ...« in the end phase of the movement[5]. Such a viewpoint seems to be in agreement with theoretical considerations regarding time-optimality. Bellman[6] and others namely showed that for systems which can be described with the vector-matrix differential equation

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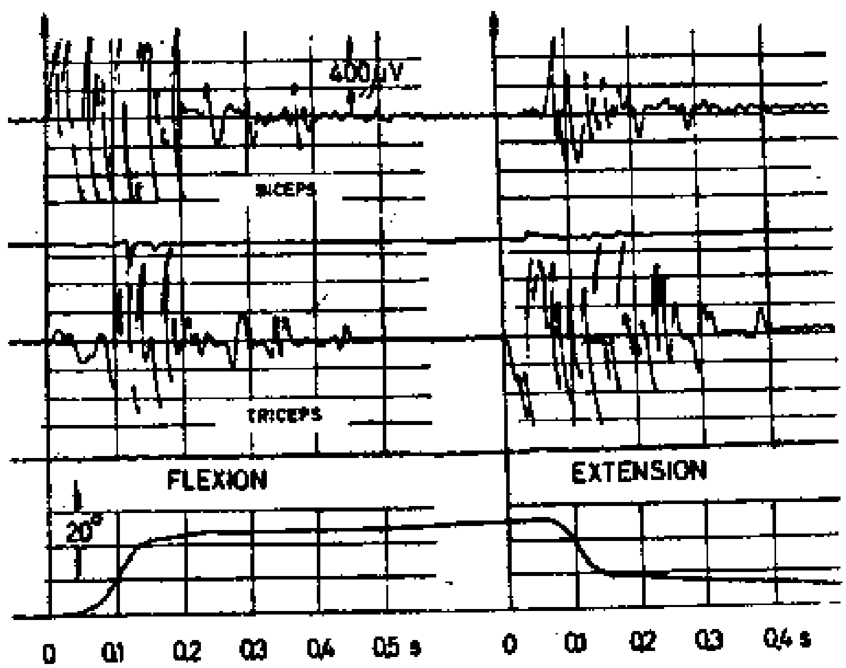

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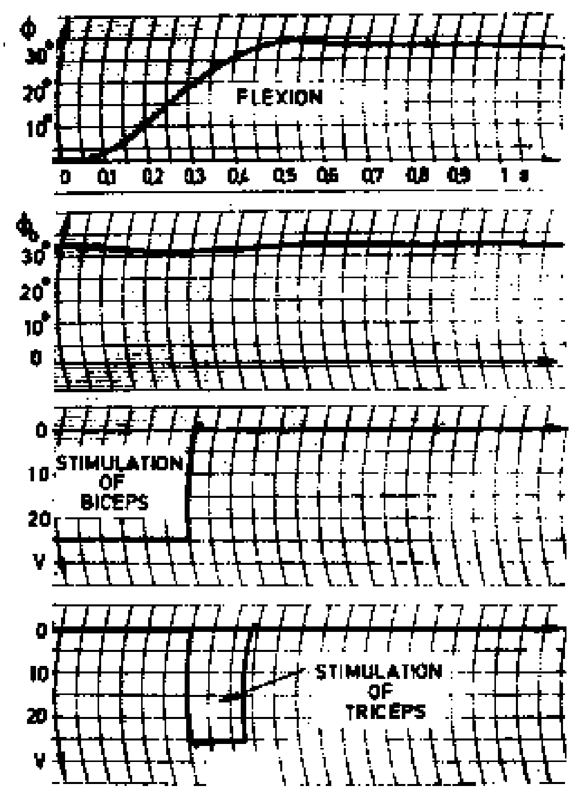

Figure 9.

With the above thoughts in mind, we tried to design a switched system for minimum-time movements by means of electrical stimulation. The block diagram is seen in Figure 8. If S_1 is open, the circuit operates as follows. At $t = 0$, the reference signal $\Phi_0(t)$ is suddenly changed, and an error $e(0)$ apprears at the input of the memory device. The output of the memory is a constant voltage $\alpha \cdot e(0)$ with $\alpha < 1$. Using a memory system, complicated feedback signals are avoided (e. g., the squared velocity used in optimum switching), and the switching moment can be determined in advance an arbitrarily adjusted as α can be varied between 0 and 1. The difference $e(t) - \alpha\, e(0)$

is through a nonlinear device fed to relay B. At t = 0, the error activates A, a_1 closes and, as c_3 and b_3 are closed too, the agonist obtains full stimulation and e(t) starts decreasing. When e(t) $\approx a$ e(0), switching occurs. B becomes unactivated, b_3 opens, b_1 and b_2 close, thereafter c_1 and c_2 close, c_3 opens and, as a result, the antagonistic muscle starts braking. The error continues decreasing and when e(t) $\approx$ 0, A switches off and a_1 disconnects the stimulator from the muscles. If S_1 is closed, there results the SAAM type of operation.

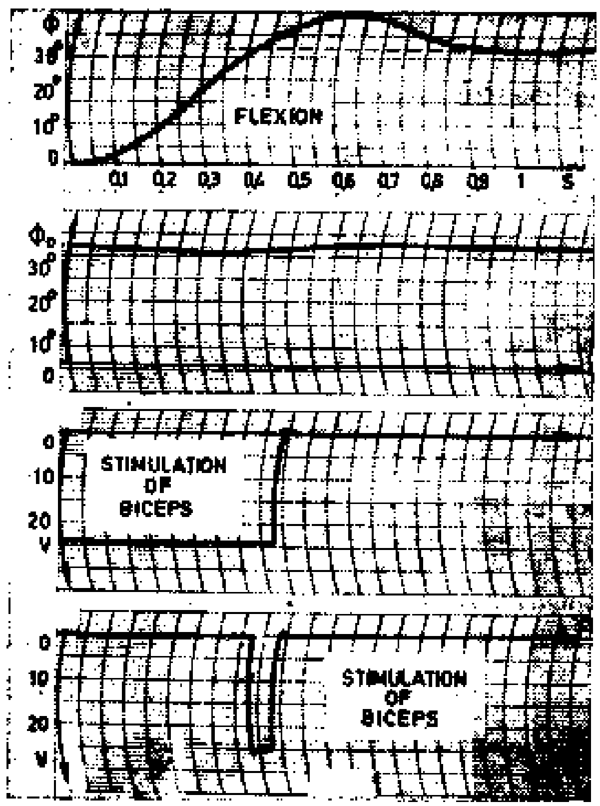

Figure 10.

Until the switching moment, the process is the same as before. After switching both muscles are contracted, and become relaxed when e(t) $\approx$ 0. Figures 9 and 10 show examples of minimum-time flexion using the circuit described above. In Figure 9 the switching moment of the bang-bang system is chosen optimally, and a response with practically no overshoot results, whereas in Figure 10, SAAM with a too late switching moment is presented, and therefore a large overshoot results. So far we have not been able to give definite opinions about which of both switching types should be performed in electrical stimulation. The main reason of our uncertainty are totally different sensations of our subjects, and the uncontrollable influence of spinal reflexes which sometimes even on normal subjects cause unexpected and unpredictable movements.

However, experiments performed up to now suggest that it would be worthwhile to design a combined control system for electrical stimulation which automatically changed its mode of operation from linear tracking for slow changing errors to switching for quick changes. Thus an adaptive system would be obtained which — in a crude way — simulated neural control.

Modelling of the Musculoskeletal System

Due to a definite lack of quantitive data necessary that optimum systems for electrical stimulation may be designed, some of our research is going to be devoted to analog computer simulation of neuromusculoskeletal control loops. This work will become even more exigent when combined electronic and neural mechanisms have to be taken in account. The ideas of Dimitrijević[8] certainly point in that direction.

In the first phase of this work we shall try to simulate the dynamic performance of volitional movements, considering only the most important elements. Rather simplified, Figure 11 shows (rotation transformed to translation) the active forces of antagonistic muscles Q_1, Q_2, their elastic and viscous components B_2, B_3, k_3 k_4, the series elastic components of muscle and tendon k_1, k_2, viscous and static or kinetic friction resp. B_1, F of the joint (eventual friction between the extermity and supporting surface should be added), and the bone with mass M. In this diagram no load forces (e. g., gravity) are included.

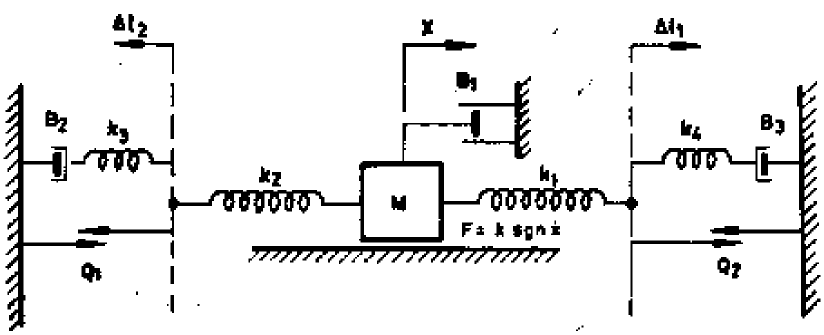

Figure 11.

The simulation in preliminary experiments is still further simplified. Instead of Q_1 and Q_2, the elongations Δl_1 and Δl_2 are considered as active. The springs k_1 and k_2 represent mainly the elasticity of the tendon. The elastic force is therefore dependent on the magnitude and direction of elongation. The parabolic relation between force and elongation was ignored, and linear relations applied. For negative elongations the spring was taken as a cord with no reacting force. With these comments the system can be desribed in terms of the following equations:

$$M\ddot{x} + B\dot{x} + K\operatorname{sgn}\dot{x} = k_1(\Delta l_1 - x) - k_2(\Delta l_2 + x) \quad \ldots \quad 5$$
$$\text{whereby } x \leqslant \Delta l_1,\ |-x| \leqslant \Delta l_2$$
$$M\ddot{x} + B\dot{x} + K\operatorname{sgn}\dot{x} = -k_2(\Delta l_2 + x) \quad \ldots \quad 6$$
$$\text{whereby } x > \Delta l_1$$
$$M\ddot{x} + B\dot{x} + K\operatorname{sgn}\dot{x} = k_1(\Delta l_1 - x) \quad \ldots \quad 7$$
$$\text{whereby } |-x| > \Delta l_2$$

For these equations (assuming $k_1 = k_2 = k$) an analog computer program was developed (Fig. 12), simulation of the biceps-triceps system being under way. It is hoped that this project will help us to understand dynamic and neural mechanisms at the control of human

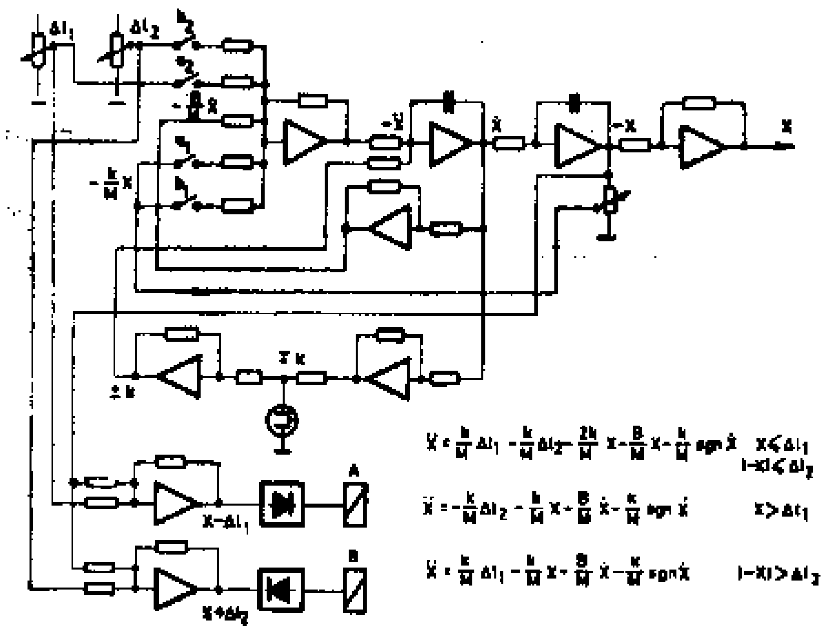

Figure 12.

extremities. As more experimental data from subjects and patients are available, it is planned that the simulation program with »neurophysiological« loops be expanded. We believe this to be the most efficient way to study hybrid (engineering and biological) control systems in order to obtain a better insight into the extremely complex mechanisms that must be mastered if functional electrical stimulation of extremities is to be extended beyond the very limited case of one-muscle stimulation, which actually is the only functional stimulation clinically applied at present.

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