HYBRID CONTROL OF THE ANKLE JOINT

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Summary

In order to help the persons who suffer from upper motor neuron lesions, but because the partly atrophied muscles are not able to have the benefit of the functional effect of electrical stimulation, the feasibility of a hybrid system to control the weak extremity is tested. For that purpose the approach to the problem is stated, the property of the muscles is searched, the synthesis of the hybrid control system is made and hybrid system control in the laboratory condition is realised. At the end of the paper the survey of the experimental results and the comments on applicability are given.

Introduction

Patients who suffer from upper motor neuron lesion have trouble with dysfunction of the extremities. It depends on the character of the lesion and disfunction, whether we classify them as hemiplegic, paraplegic, tetraplegic or quadriplegic patients.

A way to help such patients to be able to control their affected extremity is the application of the Functional Electrical Stimulation (FES) /1, 2/. The fundamental idea is that the patient can contract his muscles by help of external electrical signals, instead of the original biological signals. These signals can be produced electronically or we can use modified biological signals from another siutable organ /3/.

It was proved that the FES possesses three different properties. The first one is functional work with the affected limb. The second one is that the FES keeps appropriate muscles moving and therefore prevent their atrophy. The third property, which has not been entirely explained is a certain regeneration of nervous paths /4, 5, 6/. One property alone would be sufficient reason to improve FES in order to help paralysed and paretic persons.

A case, which is for the moment out of the domain of the FES, but where it might be applied, is the case of partly atrophied muscles. Such muscles give response to electrical stimuli to a certain extent, but have no strength to perform normal functional work. The treatment to date for these muscles is the application of electrical stimulation without the functional loading. This treatment improves the state of the patient but during the time of treatment he is not able to perform

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the functional work. In addition, because there is no functional movement the patient cannot see it and consequently has no information about the stimulating program. Therefore he is not able to coordinate his willing efforts with the functional movement.

These two disadvantages may be avoided by giving the extremity means of full functional movement. Fortunately, it seems that the problem may be solved by accepting Vodovnik's and Tomović's idea of getting the additional force for performing functional movement from one or more other sources /7/. This is the way to preserve the three above-mentioned properties of the FES when applying it to the patients with atrophied muscles. An approach to this problem has been published /8/.

Basic Principles

In this paper the possibility of the hybrid control of a certain organ, i.e. combined action of FES and an external actuator, is proved with the control of an ankle. This joint was chosen for two reasons. First, the peroneal palsy or so called drop-foot condition, where a patient cannot control his ankle is a very common phenomenon; second, the Ljubljana peroneal brace has been developed in our laboratories and so we have certain experience on this field.

One of the most essential movements of the ankle is the dorsal-plantar flexion, i.e. the movement in the sagittal plane which enables walking in coordination with other movements of the body. The dorsal-plantar flexion is for walking on a flat surface. Since the hybrid control of the ankle is intended for patients with partly atrophied muscles, then relatively heavy patients, it can be assumed that the walking on a flat surface is most important for them. That is why in this paper the attention is paid only to the control of the dorsal-plantar flexion. In the further explanation the control of the ankle will be understood as the control of this movement.

Having in mind that the ankle control must be realized with optimally preserved good effects from the FES, the first solution of the ankle control is to apply the FES to the healthy muscle tissue. This problem has been worked on and a part of the results has been pulished /9, 10/. Further, a way of applying the energy which the partly atrophied muscle lacks to perform a complete functional work, must be found.

In the study described in this paper, additional energy is applied by means of a boot-like metal exoskeleton, so that the movement in the sagittal plane of the exoskeleton causes simultaneously the same movement of the ankle and vice versa. A comparison of the electric, hydraulic and pneumatic actuators has been carried out and the conclusion was made, that the hydraulic actuators are most suitable /8/. However, because of the technical reasons the experiments were performed with the electric actuator. That is, an armature controlled, d.c. servomotor, which is mounted onto the exoskeleton joint. This joint concides with the axis of the upper part of the ankle joint. The stator is fastened to the foot part of the exoskeleton while the rotor is geared to the calf part of it.

Since the functional walking movement of the ankle in the sagittal plane is a certain periodic function, $f = \phi(t)$, a tracking servo, where tne output angle is following a certain reference periodic function, $V_0 = k \cdot \phi(t)$, is needed for the ankle control during walking.

To enable the muscles to participate as much as possible it was planned that with the hybrid control in the region of small power, where the partly atrophied muscles can still execute a certain work, the control is performed by the stimulation of the muscles. When the muscles cannot execute the work any more, the actuator takes over the control and keeps it until the system comes to the small power region where the stimulated muscles take over the control again.

Experimental Set-up

The block diagram of the experimental set-up for testing the possibilities of the hybrid control of the ankle is shown in Figure 1.

In the small power region the ankle control is performed by means of the closed loop muscles stimulation. In the forward path there is a P I controller and two channels with the modulator 1 and M. gastrocnemius and M. soleus for the plantar flexion control and modulator 2 and M. peroneus and M. tibialis anterior for the dorsal flexion control respectively. In the feedback path, there is the ankle potentiometer which is mounted on the exoskeleton opposit to the motor. Its purpose is the indication of the angle ϕ_3.

On the right side of Figure 1 the exoskeleton is schematically shown, driven by the motor and fastened to the ankle by the elastance K_3. To prevent the loading of the ankle with the motor, a signal is taken from the elastance K_3 by means of a strain gauge. This signal is proportional to the difference of angles ϕ_2 and ϕ_3, and is

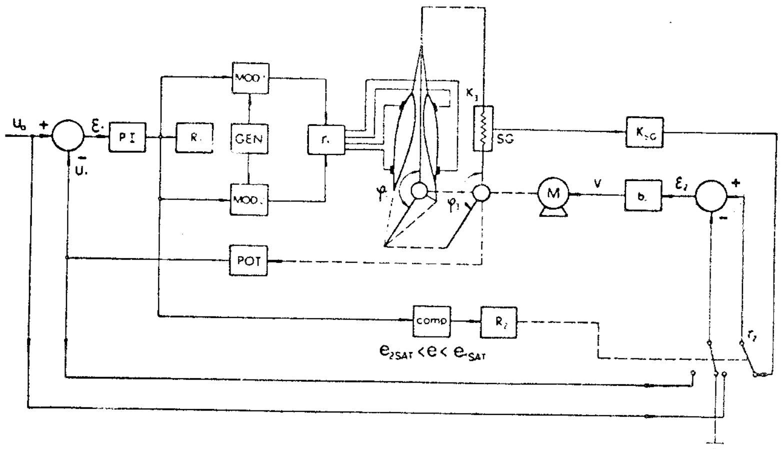

Fig. 1.

fed into the motor which now follows the ankle movements with the small resistance. A compromis was chosen between the small loading of the ankle and the stability of the system. It was found that the best response was obtained with the loop gain was set to the value where the motor produced the starting loading torque of the 0.5 Nm. The block diagram for this region of operating is shown in Figure 2.

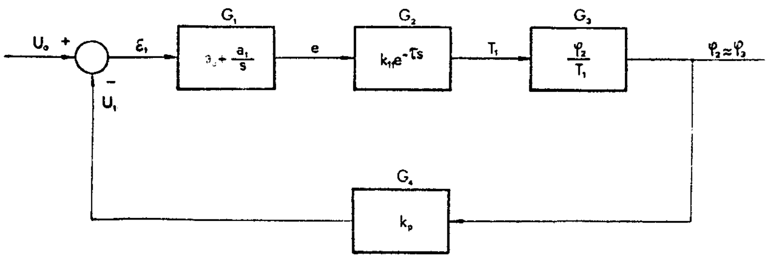

Fig. 2.

In the large power region, the manipulated error e is fed into the relay R_2, which switches the motor input to the reference input, and to the feedback path to the ankle potentiometer output. In this region, when the functional work is performed by the actuator, the muscles are kept all the time under a constant maximal stimulation.

The block diagram for this region of operation is given in Figure 3.

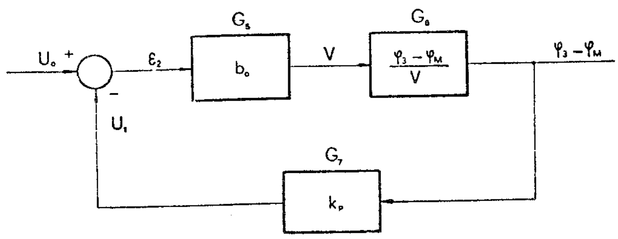

Fig. 3.

Figure 4 shows the responses to the step reference signal, taken at characteristic points.

The reponse in the region I is $T_1 = K_1 (U_{st} - U_{tr}) e^{-\tau s}$ where T_1 is the isometric muscle torque, U_{st} stimulation voltage, U_{tr} threshold stimulation voltage and τ the delay /9/. Because of this response, the error ε_1 keeps its maximum value $\varepsilon_1 = U_A$. The manipulated error e (after the controller) assumes the value $e = a_o\varepsilon_1$ at $t=t_o$, than it is integrated and reaches the value e_1 at $t=t_o+\tau$. During the same time, i.e. from $t=t_o$ to $t=t_o+\tau$, the stimulation voltage, consisting of rectangular pulses (frequency = 50 Hz, pulse width = 0.3 ms /9/), increases according to the equation $U_{st}=K_M e+U_{tr}$, where K_M is the modulator gain. During this time there is no change of angle ϕ_2 (in resting position it equals angle ϕ_3) and so there is no signal on the elastance K_3. Therefore the error on the actuator input ε_2 equals zero.

In the region II the stimulation moves the ankle and produces the change of angle ϕ_2, which through the feedback produces the error signal ε_1. This causes the reducation of $\Delta e/\Delta t$, and $\Delta U_{st}/\Delta t$. The error signal ε_2 first increases according to the dynamics of the both systems (the ankle and the actuator), then decreases when the motor starts moving.

Since, at the end of region II, the muscles cannot produce enough force to equalize ϕ_2 with ϕ_A, the error signal ε_1 assumes a certain constant value, while the manipulated variable e assumes the value e_{SAT}, which can switch the actuator from the strain gauge to the input signal by means of a comparator and the relay R_2. At

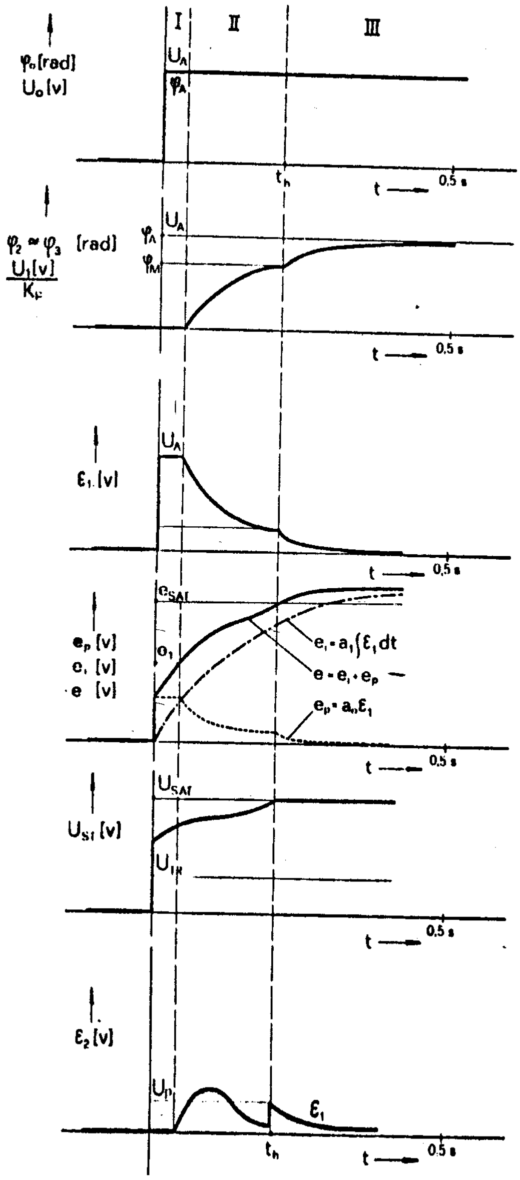
I
II
III
φ_0 [rad]
U_0 [v]
U_A
φ_A
t_h
t
0,5 s
$\varphi_2 \approx \varphi_3$ [rad]
$\frac{U_1[v]}{K_F}$
φ_A
φ_M
ε_1 [v]
e_p [v]
e_i [v]
e [v]
e_{SAT}
$e_i = a_1 \int \varepsilon_1 dt$
$e = e_i + e_p$
$e_p = a_0 \varepsilon_1$
U_{SAT}
U_{St} [v]
U_{1R}
ε_2 [v]
U_p
ε_1

Fig. 4.

the moment of switching a step function $U_p=U(\phi_A) - U(\phi_M)$, where ϕ_M is the angle at which the switching is done and ϕ_A is the angle which corresponds to the height of the step U_A, appears on the input of the actuator clodes loop. The motor by means of the exoskeleton moves the ankle to the steady state (region III).

The error ε_1 limits to zero, while the manipulated error e and U_{SAT} keep the constant values $e_{SAT}+\Delta e$ and $U_{SAT}=K_M e_{SAT}+U_{tr}$ respectively.

When the gains are properly adjusted in the plantar and dorsal flexion channels, the same response appears at the stimulation of the antagonistic muscles.

Dynamic Model of the Hybrid Ankle

In spite of the fact that the ankle is in general a nonlinear system in a limited region where the equation

$$T_1 = K_1 \, (U_{st} - U_{tr}) \, e^{-\tau s} \qquad (1)$$

is valid, it can be approximately described by linear differential equations.

This can be done because, in our case, the stimulation always remains within the limits of submaximal stimulation. After these limitations the dynamic system of the hybrid ankle, i.e. the combination of the ankle and the actuator can be approximated by the model shown in Figure 5.

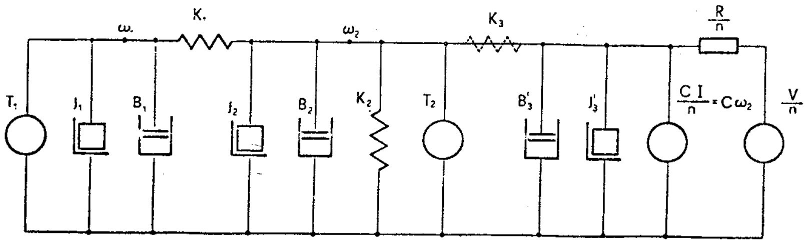

Fig. 5.

T_1 is the isometric muscle torque; T_2 is the outer loading torque, J_1, J_2, B_1, B_2, K_1, K_2 are the coefficients of inertia, viscous friction and elastance of the ankle /9/: $B_3=B_3/n^2$; $J_3'=J_3/n^2$ are the damping and inertis coefficients of the actuator on the output shaft (after the reductor). R and C are the resistance and the constant of the motor, V is the voltage at the input of the motor, K_3 is the coefficient of the elastic band between the ankle and actuator, from which the actuator controlling signal is taken during the tracking period. Since the value of this coefficient is, in reference to other coefficients, very high, it can be omitted.

The given model can be described by three equations. Two for the points ω_1 and ω_2 and one for the tension loop, ω_2, and then ϕ_2, we get the equation

$$\phi_2 = \frac{V(b_2's^2+b_1's+b_o') + T_1 b_o'' + T_2(b_2''{}'s^2+b_1''{}'s+b_o''{}')}{s^4+a_3s^3+a_2s^2+a_1s+Q_o} \tag{2}$$

where

$$b_2' = \frac{CJ_1}{nR(J_3'+J_2)J_1} ; \quad b_1' = \frac{CB_1}{nR(J_3'+J_2)J_1} ; \quad b_o' = \frac{CK_1}{nR(J_3'+J_2)J_1}$$

$$b_o'' = \frac{K_1}{(J_3'+J_2)J_1} ; \quad b_2''{}' = \frac{J_1}{(J_3'+J_2)J_1} ; \quad b_1''{}' = \frac{B_1}{(J_3'+J_2)J_1}$$

$$b_o''{}' = \frac{K_1}{(J_3'+J_2)J_1} ; \quad a_3 = \left[(J_3'+J_2)B_1+(B_3'+B_2)J_1+\frac{C^2}{R}J_1\right]\frac{1}{(J_3'+J_2)J_1}$$

$$a_2 = \left[(J_3'+J_2)K_1+(B_3'+B_2)B_1+(K_1+K_2)J_1+\frac{C^2}{R}B_1\right]\frac{1}{(J_3'+J_2)J_1}$$

$$a_1 = \left[(B_3'+B_2)K_1+(K_1+K_2)B_1+\frac{C^2}{R}k_1\right]\frac{1}{(J_3'+J_2)J_1}$$

$$a_o = \frac{K_1K_2}{(J_3'+J_2)J_1}$$

Small Power Region Operation

The dynamic model, simplified for the small power region, when the motor is controlled by the signal taken from the elastance K_3 and tracks the ankle movements, is shown in Figure 6.

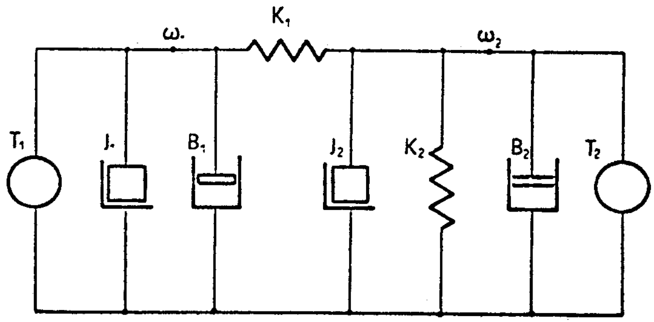

Fig. 6.

Thus the angular displacement ϕ_2 is

$$\phi_2 = \frac{T_1 c_o' + T_2 (c_2'' s^2 + c_1'' s + c_o'')}{s^4 + d_3 s^3 + d_2 s^2 + d_1 s + d_o}$$

where

$$c_1' = \frac{K_1}{J_1 J_2} \; ; \; c_2'' = \frac{1}{J_2} \; ; \; c_1'' = \frac{B_1}{J_1 J_2} \; ; \; c_o'' = \frac{K_1}{J_1 J_2}$$

$$d_3 = \frac{J_2 B_1 + B_2 J_1 + \frac{C^2}{R} J_1}{J_1 J_2}$$

$$d_2 = \frac{J_2 K_1 + B_1 B_2 + (K_1 + K_2) J_1 + \frac{C^2}{R} B_1}{J_1 J_2}$$

$$d_1 = \frac{B_2 K_1 + (K_1 + K_2) B_1 + \frac{C^2}{R} K_1}{J_1 J_2}$$

$$d_o = \frac{K_1 K_2}{J_1 J_2}$$

The elements are denoted by the same letters as before (Fig. 5) but T_2 which represents the external load when the motor is tracking.

The expression for the motor loading torque, when it tracks the ankle, is obtained from the block diagram shown in Figure 7.

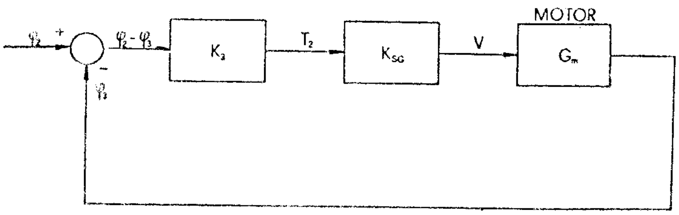

Fig. 7.

The following equation can be written,

$$\phi_3 = \phi_2 \frac{K_3 K_{SG} G_m}{1 + K_3 K_{SG} G_m} \tag{4}$$

where

K_3 – stiffness coeficient, dimension Nm/rad

K_{SG} – strain guage, Wheatstone bridge, and amplifier transfer function dimension V/K_m

G_M – motor transfer function.

From the same diagram, it is seen, that

$$T_2 = (\phi_2 - \phi_3)\ K_3 \tag{5}$$

Applying Equation 4 to Equation 5 gives

$$T_2 = \phi_2 \frac{K_3}{1 + K_3 K_{SG} G_m} \tag{6}$$

G_m is the transfer function of an armature controlled d.c. servomotor with a reductor at its output; it can be written

$$G_m = \frac{\phi_3}{V} = \frac{K_m}{s(T_m s+1)} \tag{7}$$

where

$$K_m = \frac{C}{n(RD_3' + C^2)} \tag{8}$$

$$T_m = \frac{J_3' R}{RD_3' + C^2} \tag{9}$$

And after introducing Equation 6 into Equation 3,

$$G_3 = \frac{\phi_2}{T_1} = \frac{c_o'\left[g_2 s^2+g_1 s+g_o\right]}{h_6 s^6+h_5 s^5+h_4 s^4+h_3 s^3+h_2 s^2+h_1 s+h_o} \tag{10}$$

where

$g_2 = T_m$; $g_1 = 1$; $g_o = K_4 = K_3 K_{SG} K_m$

$h_6 = T_m$; $h_5 = T_m d_3 + 1$

$h_4 = T_m d_2 + d_3 + K_4 - K_3 T_m c_2''$

$h_3 = T_m d_1 + d_2 + K_4 d_3 - K_3 c_2'' - K_3 T_m c_1''$

$h_2 = T_m d_o + d_1 + K_4 d_2 - K_3 c_1'' - K_3 T_m c_o''$

$h_1 = d_o + K_4 D_1 - K_3 c_o''$; $h_o = K_4 d_o$

After the transfer function G_3 has been known, the closed loop transfer function for the low power region can be written according to the block diagram in Figure 2

$$G_{LP} = \frac{\phi_2}{U_o} = \frac{(a_o+\frac{a_1}{s})\ k_{11}\ e^{-\tau s}\ G_3}{1+k_p k_{11} e^{-\tau s} G_3 (a_o+\frac{a_1}{s})} \tag{11}$$

<u>High Power Region Operation</u>

The block diagram for the high power region operation is shown in Figure 3 and the corresponding transfer function is:

$$G_{HP} = \frac{\phi_3-\phi_M}{U_o} = \frac{b_o G_6}{1+b_o k_p G_6} \tag{12}$$

G_6 is a transfer function, which can be determined from the block diagram in Figure 5 with the presumption, that T_1 is constant and equals 0 and that the operation starts at ϕ_M i.e. at the extreme angle to which T_1 can move the ankle.
Then

$$G_6 = \frac{\phi_3-\phi_M}{V} = \frac{b_2' s^2+b_1' s+b_o}{s^4+a_3 s^3+a_2 s^2+a_1 s+a_o} \tag{13}$$

The values of the coefficients are given by Equation 2.

That completes the Equation 12 and gives the transfer function for the high power region operation, when the motor executes the functional work.

Simultaneous Acting of Muscles and Actuator

The simultaneous work of the muscles and the actuator happens in two cases. First, in the short period of time just after the switching from the small to the high power region (because the switching takes place before the muscle control brings the ankle to the steady state). Second, when the reference step U_A is so high that the proportional part of the manipulated error reaches the value $e > e_{SAT}$ immediately at $t=t_o$.

In our experiments on a healthy patient, this problem was solved by adjusting the actuator dynamics to the muscle dynamics. For the partly atrophied muscles, where the actuator dynamics are dominant, the dynamics of the simultaneous actuation almost corresponds to the dynamics of the system in the high power region.

Experimental Results

The experiments were carried out on an experimental setup connected according to the scheme in Figure 1 with surface electrodes (5x5cm) sited on the subject as shown in Figure 8. During the measurements the subject was assuming the position shown in Figure 9. The strain gauges were stuck to the right bar of the exoskeleton's upper part (which is directly connected to the motor shaft) just above the axis of rotation, for it has proved to be the best place as far as the amplitude of K_{SQ} and the signal-noise ratio are concerned.

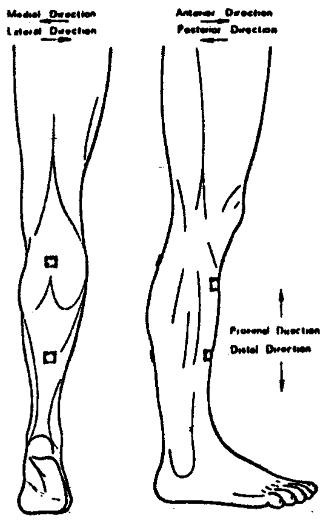

Fig. 8.

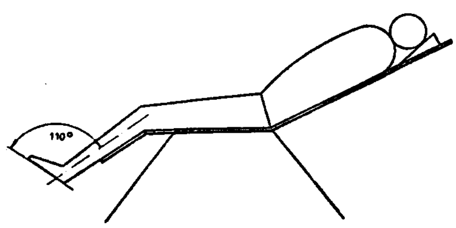

Fig. 9.

Figure 10 shows the record of the random varying reference input and the hybrid response. The upper track is the reference and the lower track is the reponse.

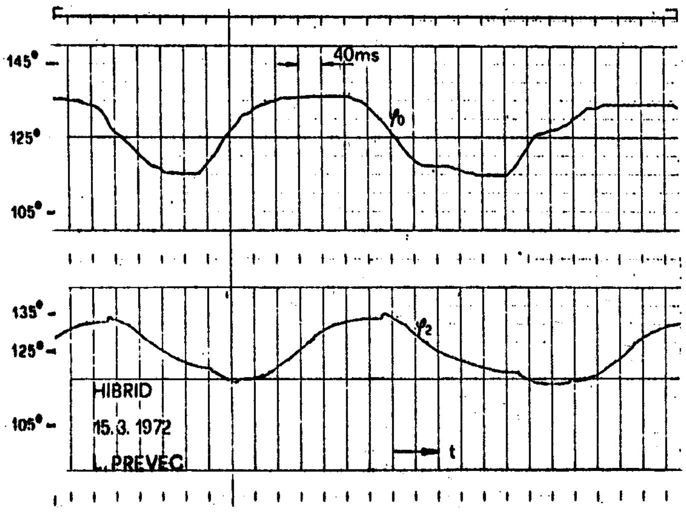

Fig. 10.

Figure 11 shows the step function response of the improved system with the indication of switching from the low to the high power region. This indication is shown by a marker on the upper part of the record. The positive jump marks the small to the high power switching and negative jump the opposite. The discontinuities are due to the bag connection between the exoskeleton and the ankle.

Conclusion

To help the patient with partly atrophied muscles the possibilities of realising the hybrid ankle control had to be tried. Based on the hypothesis of the FES therapy effect, a synthesis of an ankle controlling system was made. It consists of two subsystems, the first one with the FES control, and the second one with the outer actuator control and the elements for their coordination. The hybrid control is obtained in such a way that the muscle control acts in the low power region, while the outer actuator in the high power region. Thus

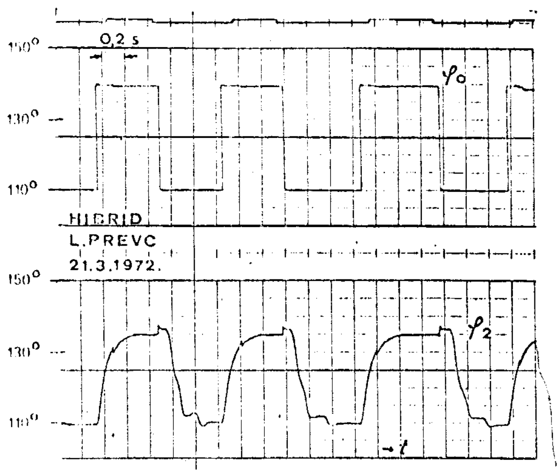

Fig. 11.

the possibility of the hybrid control was proved.

It must be mentioned, that there are still some problems which must be solved before this system could be advised and applied. One of these problems is the varying transient resistance between the electrodes and the muscle tissue, the second one is the poor mechanical connection between the exoskeleton and the ankle, the third one is the weight and the aesthetic appearance of the actuator. When these problems are solved, the hybrid ankle control could be successfully used in hospitals for rehabilitation purposes, as can be concluded from the results.

Acknowledgement

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