AN ANALYSIS OF NEUROMUSCULAR CONTROL SYSTEM AND COORDINATION

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Summary

In this work the patient-orthosis relationship is treated from the system viewpoint. Also the results of the experiments of positioning the ankle joint by means of electrically stimulated antagonistic muscle groups are given and a method of electrical stimulation of muscles that minimizes the sum of the generated muscle force for a prescribed movement is proposed.

Introduction

It is known, that the human skeleton represents a multivariable nonlinear control system, with movements of the joints as outputs /1/. From the mechanical point of view the skeleton of a patient is equivalent to the skeleton of a normal subject. This means that the dimension of the control problem is the same for patients and normal subjects. Therefore the decision to imitate the hierarchical structure of natural control is quite obvious due to the performance of the evolution of the natural system.

In an engineering sense, the main differences between a patient and a normal subject lie in the fact that the patient's information flow is interrupted at some point and therefore the control over certain muscles is lost, Also, the muscle may physically detriorate to the point that it is hampered or prevented from contracting. Thus, in the structure of the orthosis, auxilliary information must be introduced and, if necessary, externally powered actuators. In any case the patient's brain, sensory feedback, and useable muscles remain as an integral part of the orthosis structure

It is difficult to specify details of the hierarchical control structure of orthoses except that a three-level hierarchical structure is sufficient and most convenient. But it is obvious that the synthesis of such a structure must always consider the degree of

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simplicity of controlling the orthosis by the patient as the most critical point. Namely, the auxiliary information flow that transacts the patient's decisions to the orthosis is very poor. Thus, the number of the patient's decisions and interventions has to be as small as possible.

Some data about the auxilliary information flow in a normal subject have been reported /2/. It is assumed that a patient would be able to generate no more than 5 to 10 bits of auxillary information flow. It would serve the patient in giving commands and instructions to the lower levels of his multi-level orthotic-prosthetic (O/P) device.

Furthermore,it is known that the dynamics of the multi-level O/P devices is strongly affected by dead times which occur during processing and transmission of the information. It can be expected that the optimal three level control structure of the orthosis will minimise the amount of processed and transmitted information for the required dynamics. It is desired that this be as close as possible to the natural dynamics. This demand greatly increases the difficulty of the whole problem as it is known from previously repoted theoretical, experimental and clinical work /3, 4, 5, 6, 7/.

A further step in the experimental field was made by the three channel open-loop electrical stimulation applied to paraplegic patients /8/. From the system point of view an analysis and a simulation of static and dynamic stability of a simplified model of locomotion has been reported /9/. The essential point in this work is that the solution of the stability problem imitates the natural behaviour.

Figure 1 shows the three level control structure of an orthosis and the relation of the environment to the orthosis. A patient recieves the data about environment by means of a sensory channel. Also, the processor can, in general, recieve the enviromental data.

The hierarchical structure can be separated to the level of decisions, the level of direct orders, and the level of executions. The levels are connected to each other by information flow paths.

The carrier of the level of decisions is the center of consciousness of a patient's brain. The patient mediates the flow of decisions of the lower levels by the previously mentioned auxillary information flow.

The decisions can be of different types such as choosing the kind of activity of the orthosis, defining the sequence of the subprograms, defining the desired final position of the orthosis, or giving direct orders to the actuators. The flow of decisions is thus separated into the flow to processor and the flow to the orthosis. In normal operating conditions the orthosis is controlled by the processor. Only in the case of a large disturbance can the patient control the orthosis directly. In this case his orders override the orders coming from the processor. The patient receives the information about the orthosis, the processor, and environmental data by means of sensory channels.

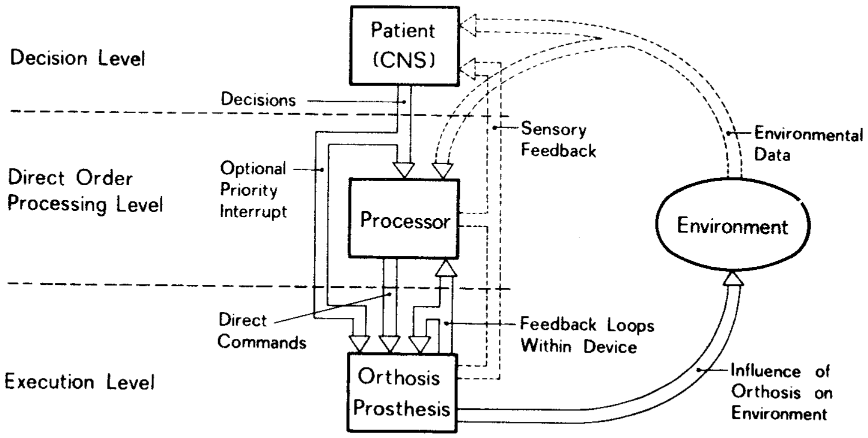

Fig. 1. The hierarchical control structure of the O/P

The carrier of the level of direct orders is in general a specialized processor. The configuration of the processor depends also on the type of the controller (performance index). The processor, based on the decisions received from the patient and the state of the orthosis, computes the sequence of the orders that are emitted by the flow of direct orders to the level of executions. It was mentioned before that the role of the level of direct orders can be in the case of a large disturbance transfered to the patient's consciousness centers.

The carrier of the level of execution is the orthosis with its actuators. This level has the property of generating force and realises the movemenents demanded by higher levels. On the the orthosis there are several different measurement points which give information about the state of the orthosis to the higher levels, The measured values are usually angles of joints, angle velocities, and accelerations, pressures, torques etc.

There are three feedback loops to be distinguished: the feedback inside the level of execution, the feedback to the level of direct orders, and the feedback to the level of decision.

In the prosecution of this work the position regulator with an electrically stimulated antagonistic pair of muscles will be introduced. In a hierarchical structure of an O/P device such a regulator realises the level of execution.

Synthesis of the Position Controller in Vivo

The idea of an artifically positioned joint or extremity by means of an electrically stimulated /10/ agonistic-antagonistic pair of muscles is not new /11, 12, 13, 14, 15/. In experiments workers used different types of controllers ranging from proportional to bang-bang. In general the experiments involved the positioning of the elbow joint.

Our experiments were performed mainly on healthy normal subjects. The ankle joint was positioned. Figure 2 shows the block diagram of the position controller. The principle of the controller is a position closed loop servomechanism which by selective stimulation of antagonistis and agonists tries to maintain the reference position of the joint.

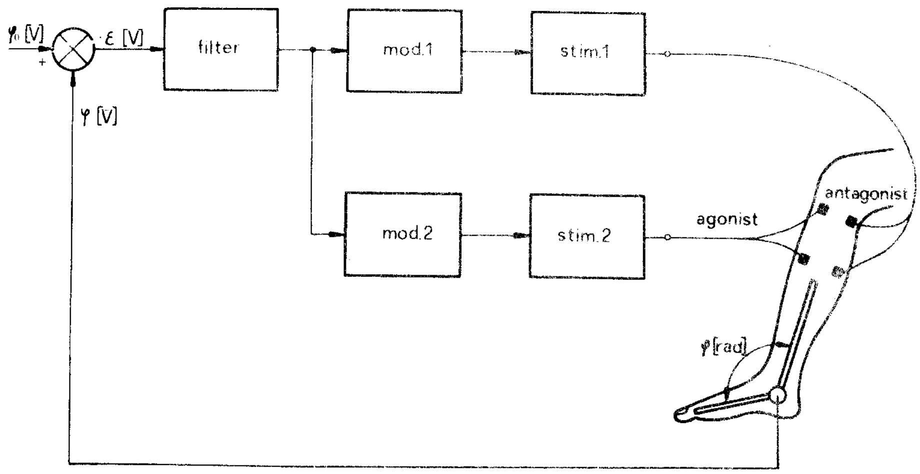

Fig. 2. Block diagram of the position controller of the ankle joint.

In the block diagram ϕ_o represents the reference or the desired position, while the feedback connection gives the ankle angle ϕ. In this diagram the influence of external load moments and gravitation is not shown.

The movement of plantar flexion is performed by the muscles soleus and gastrocnemius, while that of dorsal flexion by the muscles tibialis anterior and peroneus. The foot can be moved from 90^o to 160^o . The surface nerve and muscle electrical stimulation was applied. The details concerning the problem of the location of the electrodes for "clean movements", the pressure of the electrodes, the size of the electrodes and the method of preparing the electrodes are given in the literature /16/.

The muscles were stimulated with a voltage amplitude regulated rectangular pulsed stimulus using a repetition frequency of 50 Hz and a pulse width of 0.3 ms. The polarity of the upper and lower electrodes was selected to produce a good dorsal-plantar flexion movement.

An electrically stimulated muscle is a very nonlinear motor. Some of its nonlinear properties are described in the literature /17/. With the modulator, attempts were made to linearize the relation between the stimulation voltage U_{st} and the generated moment T_{iso}, as long as the muscle did not become saturated, The approximate nonlinear relation is given by Equation 1.

$$T_{iso}(s) = (U_{st} - U_{tr})\, k_{11} e^{-sT} \quad \text{for } U_{sat} > U_{st} > U_{tr}$$

$$T_{iso}(s) = 0 \quad \text{for } U_{st} < U_{tr} \qquad (1)$$

In this equation U_{st} is the stimulation voltage, U_{tr} is the threshold value of the stimulating voltage, k_{11} is the gain of the muscle, and T is the sum of the dead times, which are introduced by the final transmission velocity of nerves and synapses.

Figure 3 shows the functional relationship between the stimulating voltage U_{st} and the input voltage of the modulator, (output voltage of the filter), and, below, the functional relationship between the internally generated moment T_{iso} and the modulator's input voltage. U_{dec} designates the voltage of the dead zone in which no moment can be produced. All the modulator voltages are adjustable. Thus symmetrical operation of the antagonistic group of muscles can always be achieved. The gain of the modulator is adjustable and is defined by the slope of the modulator characteristic between the voltage U_{tr} and U_{st}.

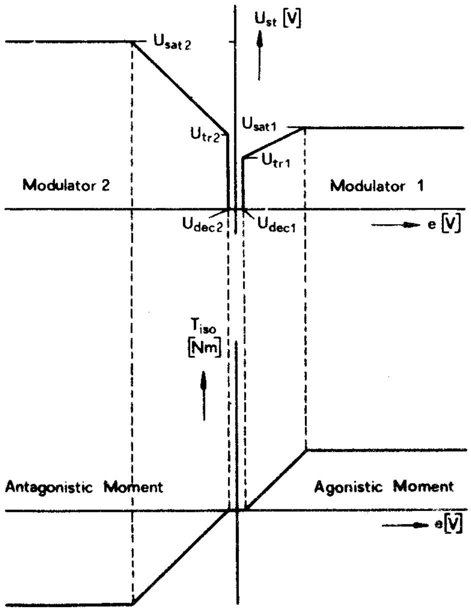

Fig. 3. Influence of the modulator structure upon the linearisation of internally generated moment.

The first step in adjusting the position controller requires a direct preliminary measurement of the nonlinear relation of the antagonistic group of muscles. This relation is given by Equation 1. This measurement is taken only after the optimal electrode site has been determined.

Results of the in Vivo Experiments

The position controller was synthesized with the aid of an EAI-580 hybrid analog computer. In order that the desired dynamic properties of the regulator could be attained, the error signal was processed by a filter. For this purpose a proportional integral differential filter (PID filter) was used. The feedback loop used a potentiometer which was attached to the leg with Velcro tape along the axis of the ankle joint.

Figure 4 shows the response of the foot to a step change of the reference signal in the direction of dorsal flexion, followed

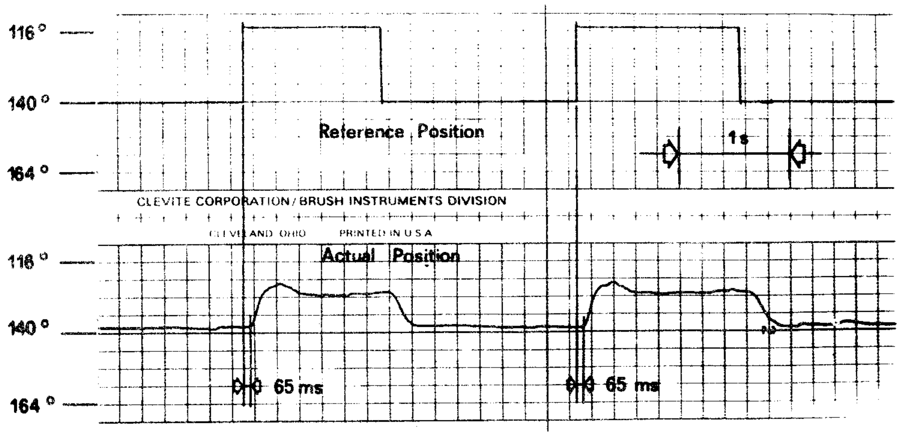

Fig. 4. Response of the ankle due to P filtering

by a step change in the direction of plantar flexion under the condition of the neve stimulation. The range of movement was between 140^{o} and 110^{o}. The filter was in this case only proportional as given by Equation 2

$$G_F(s) = 9 \tag{2}$$

From Figure 4 it is clear, that the position controller has a static error. The static error ranges from 2^{o} at the relaxed position at 140^{o} to 12^{o} at the dorsal flexion's required angle of 117^{o}. The ankle joint offers a passive elastic resistance. The static error can be reduced by increasing the proportional gain, in this case, however the response becomes too oscillatory. The dead time was approximately at the beginning of the movement 65 ms, while the rise time was not longer then 150 ms. The movement of plantar flexion has also similar dynamic characteristic except for the overshoot.

In order to avoid the static error an integral filter was added, so that the complete filter is now represented by Equation 3

$$G_F(s) = 9 + \frac{11}{s} \tag{3}$$

The values of the proportional and integral gains were determined experimentally. The response of the ankle due to the proportional (PI) filtering is shown in Figure 5. The positioning of the ankle is practically without static error, differing from the reference position by no more than 1° to 2°. The dead time during dorsal flexion is unchanged, while the dorsal flexion rise time ranges from 200 to 250 ms for a swing of the ankle from 140° to 116°.

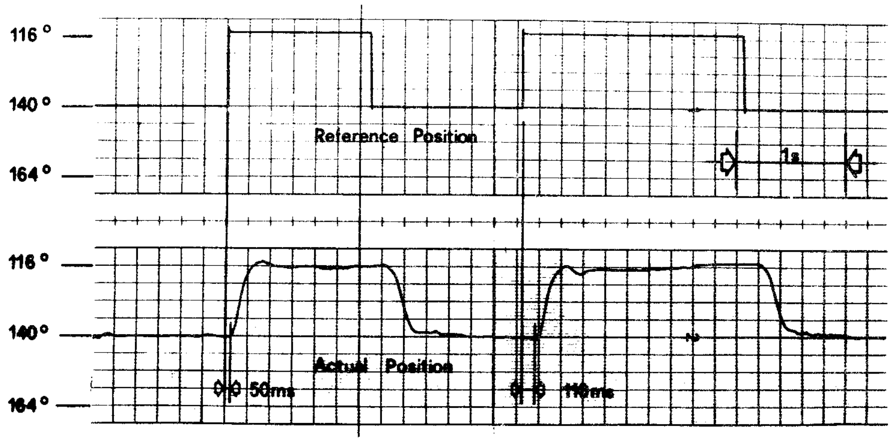

Fig. 5. Response of the ankle due to PI filtering

Dorsal flexion has an overshoot of 1° to 2°. The dead time during plantar flexion is increased to 100 ms, while the rise time toward the angle of 140° is practically the same as that of dorsal flexion. Another experiment was made in which the ankle attempted to follow a randomly varying reference voltage. The results are shown in Figure 6.

The figure shows that the ankle never lags behind the reference function more than 100 ms. The static error never exceeds 3°, while the overshoot is within the range from 1° to 2°.

Until now the disadvantages of the position controller have not been mentioned. The greatest problem concerns the repeatability of the results. All the results shown until now were recorded during only one experiment in the period of half an hour. One week later the experiment was repeated in exactly the same experimental conditions but the results were much worse.

There are many possible causes for such a poor operation of the controller. It will be recalled, that attempts were made with the modulator to linearize the relation between the internally generated moment and the simulation voltage in the region where the internal generator is not saturated. Since the conditions of

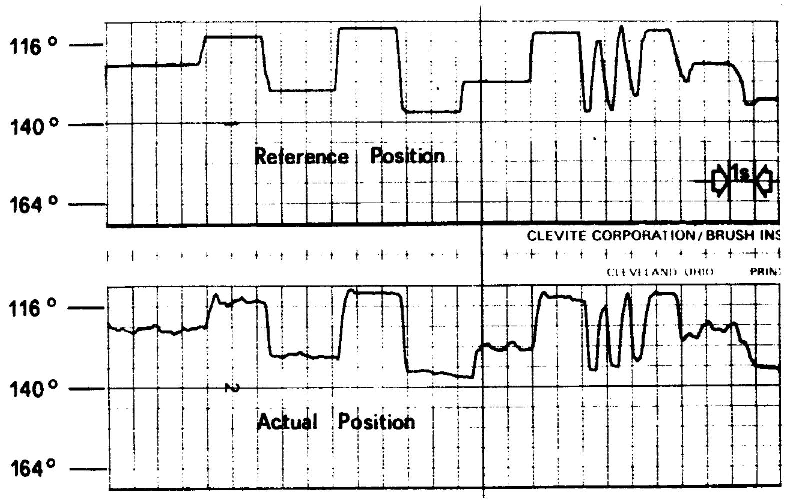

Fig. 6. Response of the ankle due to varying reference signal.

nerve stimulation are changed quickly the linearization of the characteristic is soon destroyed. Sometimes it is sufficient that the experimental subject rises from a half sitting position to a sitting position and the results are completely different.

The main difficultes stem from the great variability of nerve stimulation from the surface. Therefore the same experiment was repeated using surface muscle stimulation and some more repeatable results were obtained. Also the dynamics of movement was not behind those obtained using surface nerve stimulation.

It was desired to estimate the shortest possible rise time of dorsal and plantar flexion at voluntary control. In Figure 7 the response obtained on the same person after a training period is shown. Before the beginning of the experiment the subject was trained to reach the reference position as quickly as possible and whenever he desired. He used visual feedback.

The Figure 7 shows that the rise times of dorsal and plantar flexion are never shorter than 200 ms. Such quick movements can also be obtained using the position controller in vivo.

It can be concluded from the measured results that the position controller in vivo has dynamic properties as good as those observed by positioning with visual feedback. If long term reli-

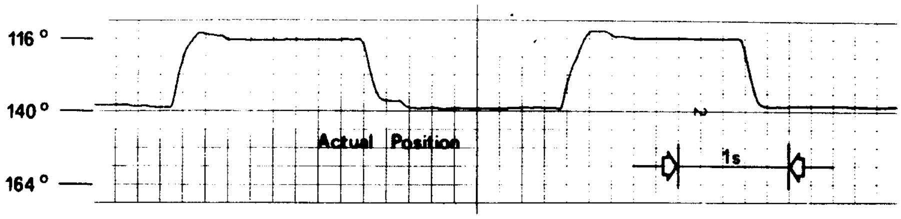

Fig. 7. Fastest voluntary movement of the foot.

ability and stability of operation of the regulator in vivo can be achieved, the controller would be sufficient as the output stage in a multilevel controller for coordinated movement.

In normal conditions it is desired to improve the efficiency of the FESE muscles and the system. These problems are the subject of the next chapter.

Optimisation of the Exerted Muscle Force

Because of the simplicity of the problem until now an agonistic and an antagonistic group of muscles have been treated as one bidirectional single muscle. Of course, it is physiologically wrong. Theoretically, such a single muscle applied to a bone is not optimal from the energy consumption viewpoint and from the controlling viewpoint. The natural criterions of muscle coordination are very complicated,neurophysiologically conditioned and to a great amount unknown.

At the construction of orthoses and prostheses we found a new mechanical criterion of muscle coordination that minimizes the necessary muscle force of different angles of the joint for the desired torque /19/ and is based on the anatomical properties of a joint.

It is known that levers which transform muscle forces to a torque are functions of joint angles. Also the maximal muscle force is a function of the muscle length which is changing with the joint angle. Thus the following equation can be writen for the exerted torque.

$$T_{iso} = \sum_{i=1}^{N} r_i(\phi)\ F_i[U_{st},\ l(\phi),\ \mu(t)];\quad i = 1,2...N \tag{4}$$

N - the whole number of muscles in a group of muscles working in one direction at a certain joint

$r_i(\phi)$ - the moment radius of each muscle as the function of the angle

U_{st} - the amplitude of the stimulation voltage applied to the muscles

$l(\phi)$ - the length of each muscle as the function of the angle

$u(t)$ - changes introduced by time (fatigue, health, physiological condition, etc.)

T_{iso} - the torque at a certain joint

From the Equation 4 it is clear that the exerted muscle force is in general a nonlinear multivariable function.

If the extremity is loaded, for a prescribed change of the joint angle from ϕ_1 to ϕ_2, a certain torque T is necessary. The question is how to stimulate the muscle to get the minimum muscle work and hence the minimum fatigue. To simplify the problem we shall reduce the problem to a four muscles model (two agonists and two antagonists) of a hypothetical human arm joint orthosis. The following expression is desired to be minimized.

$$F_1 + F_2 \rightarrow \min \tag{5}$$

F_1 and F_2 are muscle forces working in the same sense but on the different moment radius. The Relation 5 is correct when considering the following limitations,

$$0 \leq F_1 \leq F_{1max} \quad \text{and} \quad 0 \leq F_2 \leq F_{2max} \tag{6}$$

Where F_{1max} and F_{2max} represent the maximum values of force of certain muscles at different angles. The needed torque T is expressed by the Equation 7

$$T = F_1 r_1 + F_2 r_2 \tag{7}$$

From equations 5 and 7 it is clear that the muscle working at the greater moment radius must put out more energy. The problem at a fixed joint angle can be solved using a linear programming technique. For the demanded movement from ϕ_1 to ϕ_2 the muscle contributions which continually fit Equation 5 can be found.

If the Equation 5 is to be considered the structure of the modulator must be completed as shown on the Figure 8.

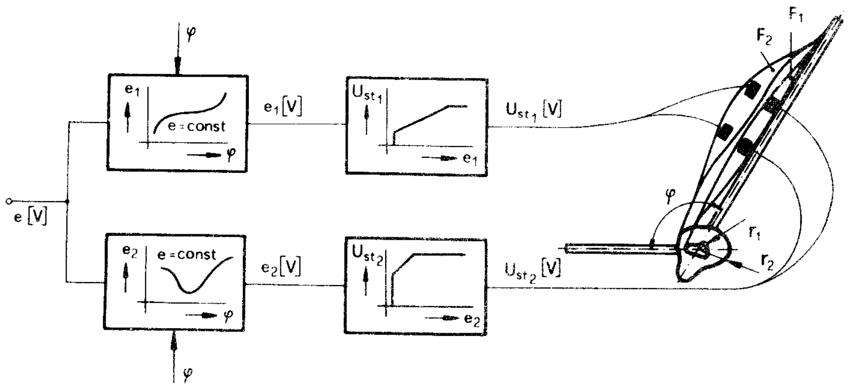

Fig. 8. Modification of the modulator considering minimization of exerted muscle forces

From the Figure 8 it is clear that from each antagonistic group of muscles two modulators and two stimulators have to be used. The modulation functions between e and e_1 and e and e_2 are defined by the anatomical forms of the moment radius r_1 and r_2, fulfilling the conditions represented by Equation 5.

Until now the optimisation of the exerted force based only on the nonlinear forms of moment radius has been discussed. For a more sophisticated optimisation the fact that the maximum muscle force depends upon the length of the muscle and the velocity of shortening must be taken into account /20/. In this case we have to make another change in the structure of the modulator which means also more instrumentation. At this point feasibility is being studied. We do not know a method for stimulating only a single muscle without influencing other muscles.

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