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SIMULATION OF QUIET STANDING BY A HYPOTHETICAL CLOSED-LOOP ELECTRICAL STIMULATION ORTHOSIS

R. J. Jaeger

Pritzker Institute of Medical Engineering, Illinois Institute of Technology, IIT Center, Chicago IL 60616 USA

ABSTRACT

A simulation model of quiet standing has been developed to study the potential use of a hypothetical closed-loop electrical stimulation orthosis. The model consists of a single-link inverted pendulum, with electrically stimulated muscles providing balancing torques at the ankle joint. It also includes a proportional-integral-derivative controller and feedback compensation. The initial simulations have shown that quiet standing can be regulated within certain limits subject to the available torque produced by the stimulated muscles. The simulation results are similar to data observed in normals.

INTRODUCTION

The maintenance of upright posture in normal human subjects is not a trivial task. In the multi-link inverted pendulum, the exact means by which static stability is maintained during quiet standing is not yet fully understood (1). In recent years, several laboratories have demonstrated the feasibility of using electrical stimulation as an orthotic device for standing and forward progression in upper motor neuron paralysis (2,3). These feasibility demonstrations have been open-loop in the sense that the stimulation is applied to produce the movements without direct feedback information regarding the consequence of the stimulation. Two questions which remain unanswered are first, whether present open-loop systems can become clinically accepted or whether some type of closed-loop system is required. Given that the answer to the first question is that closed-loop control is required, the second question is: can such systems be realized?

Previous studies of closed-loop control of muscle have included isometric regulation of force (4), control of elbow position (5, 6), and control of ankle position (7). In all of these studies, reasonable control could be obtained over well-defined periods of time under laboratory conditions.

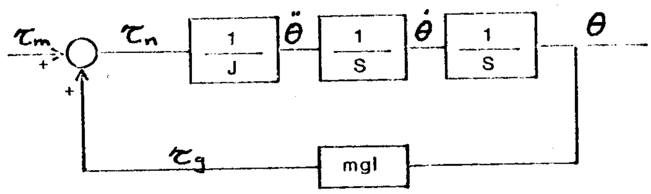
The purpose of the present study was to investigate the feasibility of providing closed-loop control of postural sway in the antero-posterior direction during quiet standing. A reduced-order single-link model was chosen and attention was restricted to $\pm$ 0.05 radians of the vertical standing position. A model of electrically stimulated muscle was chosen to produce balancing torques about the ankle. Finally, a proportional-integral-derivative controller and feedback compensation were selected. The complete model was simulated on a computer (8).

METHODS

Description of System Components

The block diagram for the inverted pendulum is given below. Control of the pendulum is obtained by the torques about the ankle produced by the plantar- and dorsiflexors of the ankle. The torque about the ankle due to gravitational force is a function of ankle angle for small angles. The net torque about the ankles is multiplied by the reciprocal of the moment of inertia to give net angular acceleration about the ankle. This is integrated to give net velocity about the ankle. A second integration gives the angular position of the body about the ankle. A positive feedback loop exists, and for the initial condition of absolute vertical, any uncontrolled torque signal applied will cause the angular position to diverge and the pendulum will fall.

τ_m = torque of muscles about ankle
τ_g = torque due to gravity
τ_n = net torque about ankle
J = moment of inertia
$\ddot{\theta}$ = angular acceleration
$\dot{\theta}$ = angular velocity
θ = angle
m = mass of body
l = distance from the ankle to the center of gravity
g = gravitational acceleration



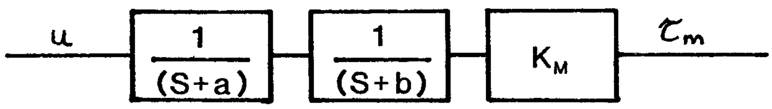
The transfer function of the inverted pendulum is:

$$\frac{\theta}{\tau_m} = \frac{1/J}{\left(S^2 - \frac{mgl}{J}\right)}$$

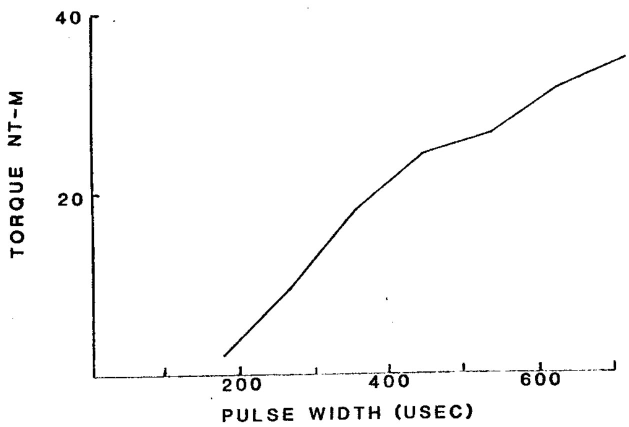
The block diagram for electrically stimulated muscle is given below. The input is pulse width in microseconds and the output is torque about the ankle in Newton-meters. This model is based on our observations in stimulated upper motor neuron paralyzed muscle and modeling performed in normal muscle (9). This is essentially a two-pole low-pass filter.

The pulse width is limited to 700 microseconds.

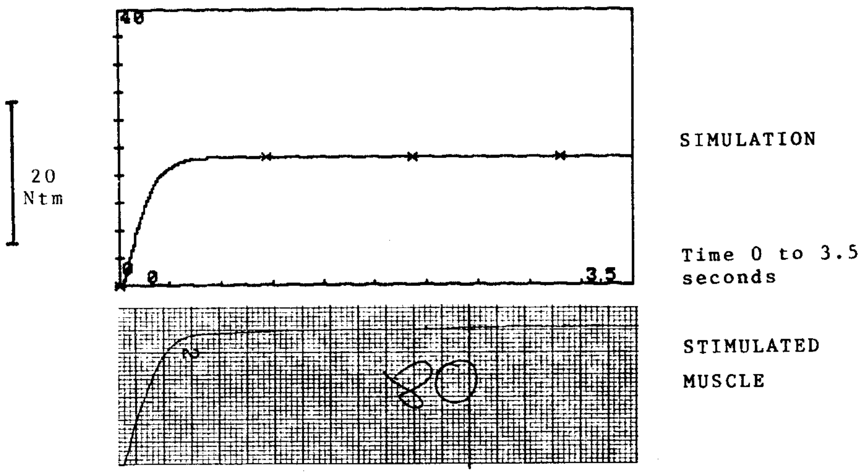
u = muscle control signal (pulse width)
τ_m = torque produced by muscles at the ankle (Newton-meters).



Using surface electrodes, we stimulated with pulse trains and measured the peak force. Using different pulse widths with each train, a recruitment curve could be obtained. The results from one experiment is shown below.



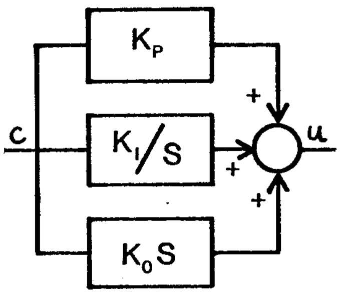
The response of a soleus muscle to a "step" input (a pulse train of constant frequency, pulse width and voltage) is shown below along with simulation results. In this particular case, the agreement is good. At lower voltages, the actual muscle response becomes more sluggish.



The transfer function for the muscle is:

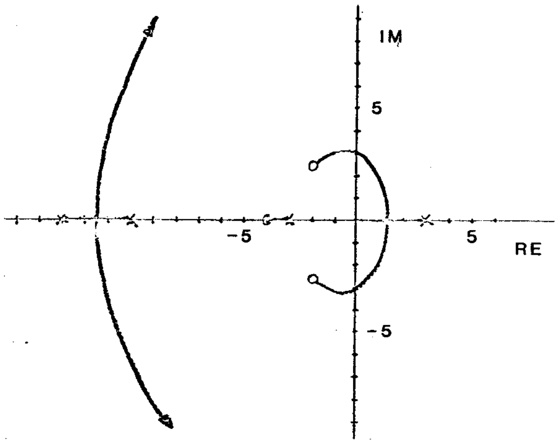
$$\frac{\tau_m}{u} = \frac{K_m}{(s+a)(s+b)}$$

The P-I-D controller is a standard configuration; the block diagram is given below:



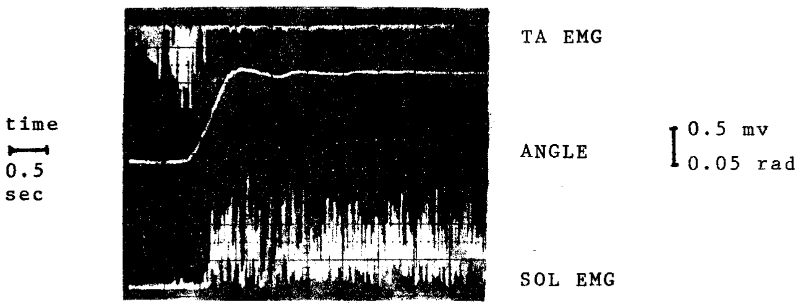
A proportional-plus-velocity block was chosen for appropriate feedback compensation.

The root-locus plot for the closed-loop system is given below:



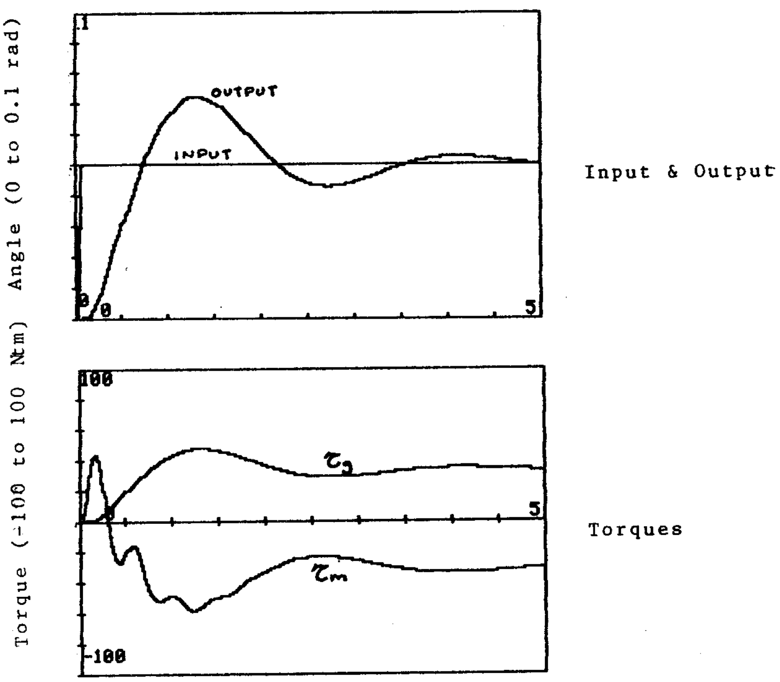
<u>Measurement of Normal Postural Adjustment</u>

A normal individual was used in an experiment to measure how the simple postural change of leaning forward was accomplished. The EMG activity of <u>soleus</u> and <u>tibialis anterior</u> were monitored with surface electrodes and displayed to the subject on an oscilloscope. Ankle angle as measured by a goniometer was also displayed. The subject was instructed to lock the knees, slightly hyperextend the hips, and attempt to stand with minimal or zero EMG in the two muscles. When a brief period of EMG silence in both muscles was observed the subject was instructed to lean forward 0.05 radians as quickly as possible. the <u>tibialis</u> <u>anterior</u> EMG, ankle angle and <u>soleus</u> EMG were recorded on a storage oscilloscope. A single record is shown below:



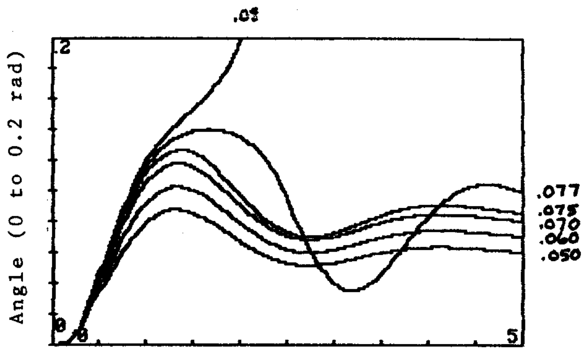
RESULTS

The step response of the system was simulated for a +0.05 radian input. For the sign conventions adopted, this was a command to lean slightly more forward. The command signal and output signal are plotted below, along with the torque produced by the muscles and the torque exerted by gravity at the ankle:



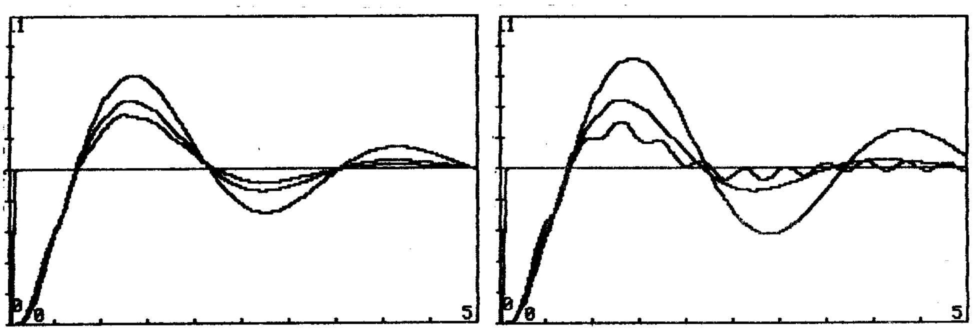
Time (0 to 5 sec)

If the amplitude of the input command is beyond the limits of the model, the muscles can no longer produce sufficient torque to control the system, and the pendulum diverges as shown below, for inputs of 0.05 to 0.08 radians:



Time (0 to 5 sec)

The robustness of the controller was tested by changing the muscle gain by ± 20%, and shifting the location of the muscle poles by ± 20%. Simulation results for these two conditions are shown below:



VARY GAIN VARY POLES

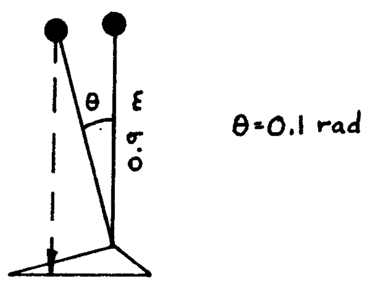
(Scales: vert 0 to 0.1 rad, horiz 0 to 5 sec)

DISCUSSION

The simulation results indicate that closed-loop control of standing over a limited range in the antero-posterior direction is feasible by electrical stimulation of muscles. Simulation results approximate observed data from a normal human subject.

Model Restrictions:

During quiet standing in normals, body sway does not exceed about 0.1 radians, and the center of force falls slightly anterior to the ankle. For excursions of ± 0.05 radians about vertical, this corresponds to movements of the center of force of approximately ± 4.8 cm for a 170 cm tall individual (center of gravity approximately 90 cm above ankles). Excursions beyond this leads to falls, unless the individual departs from the single-link ankle strategy (see (10) for a discussion of hip versus ankle strategies). This is illustrated in the figure below:



The single-link model with the above angular restrictions is valid for paraplegic individuals standing by a hypothetical electrical stimulation orthosis, since their knees will be actively locked and their hips will be passively locked in hyperextension. This is presently the only available standing strategy for paraplegic individuals using knee-ankle-foot orthoses. Thus, sway in the A-P direction will occur about the ankle joint and be influenced by electrically stimulated activity in the <u>soleus</u> and <u>tibialis</u> <u>anterior</u> and by gravity.

If the model restrictions are exceeded, the single-link inverted pendulum model ceases to be valid. This would be the case under the following conditions:

1. The hips are no longer in hyperextension in which case the individual would either be engaged in ambulation or falling (the so-called "jackknife"), or
2. the knees are no longer locked in which case the individual would be collapsing straight down.

<u>What is the Regulated Variable?</u>

In this simulation study, the angle of the ankle has been taken as the regulated variable. It might be more realistic to assume that the regulated variable is actually the position of the center-of-force within the base of support, and that in the reduced-order model, the two quantities are related. As soon as the center of force departs from its base of support, the model is no longer valid and the neuromuscular system must choose some other variable(s) to regulate.

<u>Controller Robustness</u>

Biological systems have typically been regarded as having plasticity, or being adaptive, that is, they change their control laws slowly over time in response to changes in the plant (11). It has typically been assumed that artifical controllers must also be adaptive. It has recently been suggested that design of adaptive control systems to interface with physiological systems may be unwise (12). Interest has also been generated in the design of robust controllers, that is, fixed controllers which give acceptable performance in spite of variations in the plant. One of the problems with this approach is that "acceptable" performance must be defined. This requires additional data from normal subjects to establish what the performance criteria are for the normal system.

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