CLOSED-LOOP CONTROL OF THE HUMAN LEG USING ELECTRICAL STIMULATION

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ABSTRACT

This project is part of a major program that has the long-term goal of developing implantable electrical stimulation systems to help partially paralyzed individuals walk faster with less effort. The specific problem addressed in this project is the control problem, i.e. 1) development of a method for determining the stimulation sequences to be applied to particular muscle groups to produce a desired movement of the leg, 2) derivation of a feedback control law which will minimize deviations from the desired trajectory and 3) derivation of a method for modifying on-line the parameters of the controller to compensate for system variations over time.

The experimental portion of the study will be restricted to electrical stimulation of the agonist/antagonist muscle groups which produce flexion and extension of the human knee. The experimental protocols are designed to provide quantitative data required to solve the control problem. Because the system to be controlled is highly nonlinear, a perturbation method is proposed to estimate the impulse responses of the nonstationary linear system that approximates the dynamic behavior of the system in the neighborhood of a nominal pattern of motion. Based on these data, numerical algorithms will be derived to iteratively calculate parameters of the closed-loop controller.

INTRODUCTION

Walking on level ground, inclines or stairs is a complex but repetitive activity. The requirements of a controller for a functional electrical stimulation (FES) system to assist partially paralyzed individuals during walking are to produce a basic periodic pattern of activity in the stimulated muscles plus compensate for inconsistencies in response and disturbances that perturb these patterns. A logical form for the control system is shown in Figure 1. The target trajectory $\bar{y}(t)$ is the desired cyclical pattern of movement. It is compared with the measured trajectory $y(t)$ to obtain $\delta y(t)$, the instantaneous deviation from the target trajectory. (In this project, y is scalar and represents the anterior/posterior angle of the knee, but in the walking subject it could include kinematic measurements of other joints, center of pressure beneath the weighted foot, and acceleration of the center of the body mass.)

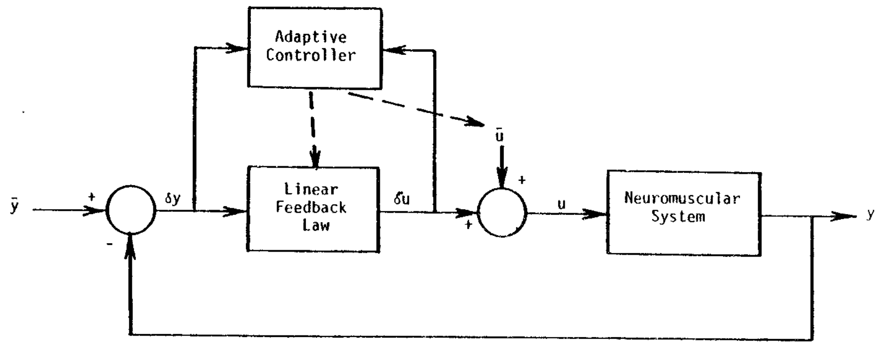

Fig. 1. Block Diagram of Feedback Controller

The purpose of the closed-loop controller is to select stimulus pulse trains u(t) to keep $\delta y(t)$ small. The controller contains a stored set of stimulation sequences $\bar{u}(t)$, called the nominal control, which on the average produce the desired pattern of movement. The nominal control may be cycled automatically or triggered by an event such as heel lift-off. Small changes in the amplitudes of the control pulses $\delta u(t)$ are added to $\bar{u}(t)$ to compensate for deviations from the target trajectory. $\delta u(t)$ is calculated from a feedback law which in general will be a linear function of the current value of $\delta y(t)$ and prior values of $\delta y(t)$ and $\delta u(t)$. (Although variables are shown as continuous functions of time, all controller computations will be performed digitally from sampled values of u and y.)

Since the neuromuscular system and the electrode/tissue interface vary with time, it will be necessary to periodically update the nominal control $\bar{u}$ and the feedback coefficients. This will be done by comparing the actual dynamic response of the trajectory deviations with the response predicted from a linear model stored in the controller.

The problem of controlling the patterns of movement during walking with electrical stimulation is extremely complex, involving a host of experimental and analytical difficulties. This perhaps explains why the field of FES is still at a relatively elementary level of development. Feedback has not yet been incorporated into stimulation systems used with walking subjects, and selection of open-loop control has been based on trial and error and intuitive judgement.

We believe that procedures for selecting the nominal control and optimal feedback laws can best be developed by first considering a sufficiently simple, carefully controlled experiment in which results can be meaningfully evaluated. Methods developed from this research, however, will be applicable to the control of hip and knee in walking subjects, a problem which will be addressed in subsequent projects. A summary of our approach is as follows:

1. A restricted pattern of limb movement, specifically flexion and extension of the knee, will be controlled. The subject will be seated, the thigh will be restrained, and the lower leg will be free to swing at the knee. Target trajectories will include patterns of free movement and patterns in which the leg moves freely during part of the trajectory and against a force-loaded constraint during the remainder of the trajectory.

2. A series of experiments will be conducted to determine methods for stimulating the agonist/antagonist muscles to control flexion and extension of the knee and to generate quantitative data for use in the analysis and design of the closed-loop controller.

3. Computational techniques will be derived to compute open-loop control sequences, to calculate feedback coefficients and to modify these coefficients on-line to automatically adapt to time-varying elements of the system. These algorithms will be based on experimentally determined impulse responses of the linear system that approximates the behavior of the system in the neighborhood of the nominal (average) trajectory.

4. The efficacy of the derived computational techniques will be verified in additional experimental studies.

SPECIFIC OBJECTIVES

The objectives to be achieved are as follows:

1. Develop a fast, efficient procedure for automatically computing the nominal control.

2. Develop a method for calculating the coefficients in the feedback law to minimize a weighted mean-square cost function.

3. Develop procedures for modifying the nominal control and the feedback coefficients during prolonged runs to compensate for variations in the system.

The closed-loop controller will be subject and trajectory dependent; i.e., each of the above procedures must be repeated for each human subject and for each target trajectory to be considered. Therefore, it is essential that each of these procedures be performed as efficiently and quickly as possible.

The proposed research is a combination of experimental and analytical studies designed to meet the specific objectives stated above. Initial experimental studies will be conducted to 1) determine the target trajectories to be used in this study, 2) determine methods for precisely controlling the agonist/antagonist muscles of the knee and 3) obtain data that are required for synthesis of the feedback controller. Following derivation of the numerical techniques for calculating controller parameters, the techniques will be tested on subjects and modified if necessary. The initial experimental phase of this research is described and some preliminary results are presented in this paper.

METHODOLOGY

Subjects

Three groups of subjects will be tested: normals with no neuromuscular deficits, patients who are completely paraplegic with lesions at T11 or above, and incomplete spinal-cord-injured patients who walk with braces and/or crutches. Normal subjects will be used because they are readily available and volitional leg movement is required in preliminary studies. Subjects who are completely paraplegic will be tested because this is the only group in which the elimination of volitional effort during stimulation tasks can be guaranteed. Incomplete spinal-cord-injured patients will be tested because they are our ultimate target population for implanted FES systems. The total number of subjects to be tested will depend upon the consistency of response within each test group, but at least 5 subjects in each of the three categories will be tested. All subjects must meet the following criteria:

1. Moderate to no spasticity.
2. Normal or below average weight for their age and sex.
3. At least 60^{o} range of motion at the knee and lacking no more than 5^{o} of full extension.
4. No orthopedic problems of the knee.
5. No cardiovascular problems.
6. At least 19 years of age.

Experimental Set-Up

In each experiment, the subject will be positioned as shown in Figure 2. The body will be supported as necessary to maintain the subject in a constant, comfortable position. The leg to be stimulated will be supported at the thigh and will be free to move at the knee. The hip will be flexed so that the knee is in $35\text{-}40^{o}$ of flexion when the subject is completely relaxed. Movement of the lower leg will be monitored by an electrogoniometer at the knee. The electrogoniometer data will be digitized prior to processing. In some tests, electromyograms will be recorded from fine wire electrodes placed in each of the muscles of the quadriceps and hamstrings during volitional control of the limb.

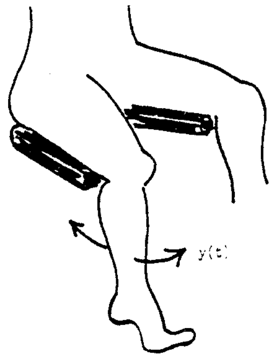

Fig. 2. Subject Position During Testing

Stimulation Technique

The quadriceps and hamstring muscle groups will be stimulated through surface electrodes. Carbon-impregnated silicon rubber electrodes will be used with a conductive gel interface between the electrode and the skin. It has been shown by Trnkoczy,(1) and we have found in our experience, that some subjects show excessively variable responses to surface stimulation. These subjects will be excluded from the study since our goal is to develop techniques of control synthesis that are applicable to implanted systems, and highly variable responses are not typical of such systems.(2)

The tendons of all four heads of the quadriceps fuse into a single tendon which inserts into the base of the patella. With proper positioning, a single active electrode can achieve good extension over the range of stimulus amplitudes and leg positions that are of interest. On the other hand, the muscles of the hamstrings attach separately on the medial and lateral surfaces of the leg and produce rotation as well as flexion. The biceps femoris laterally rotates the leg while the semitendinosis and semimembranosis rotate the leg medially. For this reason, it is sometimes difficult to achieve flexion with a single active electrode. In these cases, two channels will be used to stimulate the hamstrings, with one active electrode placed over the biceps femoris and another over the semimembranosis and semitendinosis.

A multichannel stimulator will be used to control the contractile state of the agonist/antagonist muscles. Each channel puts out a train of charge-balanced constant-current, biphasic pulses that are characterized by three parameters: pulse amplitude, pulse duration and interpulse interval. Pulse amplitude and pulse width can be independently controlled from one pulse to the next and will be used to control the active state of the muscle by motor unit recruitment. Interpulse interval can be varied, but will not be changed during a run.

In general, recruitment will be controlled by varying pulse amplitude with pulse width fixed at 0.3 ms. Crago et al have shown that the charge required to modulate muscle output is minimized when pulse duration is used.(3) Sensation is, however, a more important consideration when using surface electrodes, and Bowman et al have shown that 0.3 ms pulses are much better tolerated by subjects than shorter pulse durations.(4) Pulse amplitude will be limited to 120 ma or to the subject's tolerance. The interpulse interval will be set to a nominal value of 40 ms. Interpulse interval is a trade-off between dynamic response and muscle fatigue. Increasing interpulse interval prolongs time to fatigue (5) but diminishes dynamic response.(6) The nominal value of 40 ms will be adjusted if experimental results indicate that a change is desirable.

Target Trajectories

To insure that the computational procedures to be derived are broadly applicable, a variety of target trajectories will be employed in the experimental studies and in later tests on the efficacy of the derived procedures. Two basic trajectories to be used are a sine wave and a modified sine wave with a pause (Figure 3). The second trajectory was chosen to present a greater challenge for the controller because of the requirement for more abrupt accelerations and decelerations. Periods of 2 and 4 seconds will be investigated. These are chosen to approximate slow walking (1/2 of normal) and very slow walking (1/4 of normal). The amplitude of the swing will be $\pm 30^{\circ}$ to encompass the range of the angle of the knee during normal walking.

Trajectories that approximate the phasic activity of the quadriceps and hamstrings during slow walking will also be used. These muscle groups are active simultaneously prior to and after heel strike. To duplicate this with a seated subject, it is anticipated that a force-loaded constraint will have to be added in the forward-swing path to simulate the conditions that exist at heel strike. Tests will be conducted on normal volunteers to determine the exact time history of these trajectories.

Tests will also be conducted to determine the "normal" phasic activity of the agonist/antagonist muscles during volitional performance of each of the target trajectories to be used in this project. Normal subjects will be asked to voluntarily move their limbs and follow one of the target trajectories as closely as possible. To assist the subject, knee angle will be displayed in real time and superimposed with the target trajectory on a CRT terminal. The subject will be allowed to practice the maneuver, and then a run consisting of at least 15 cycles will be recorded. In addition to knee angle, EMG data will be recorded from each muscle of the quadriceps and hamstrings. These data will be used to select a priori estimates of the open-loop control sequences in the studies described below.

Experimental Studies

A series of studies will be performed to provide the quantitative data that are required to derive procedures for calculation of the parameters of the controller. The specific objectives of the experimental studies are:

1. To determine how close an intelligent operator can come to the desired trajectory by manually varying the control sequences.

2. To determine the number of cycles required to reach steady-state at the beginning of the run.

3. To determine the minimum number of cycles required to obtain an accurate estimate of the average trajectory.

4. To determine the effect of fatigue and other system variations during an extended run.

5. To determine the trajectory deviation caused by perturbing a single control pulse at different phases of the cycle.

6. To determine the range over which the dynamic respone of these deviations can be described by a nonstationary linear system.

7. To determine how consistent these trajectory deviations are over time.

The methodology to achieve these objectives is as follows. Based upon EMG data and intuitive judgement, a priori estimates of control signals to the quadriceps and hamstrings will be selected. Two methods will be used to alter the control sequences to better approximate the target trajectory. In the first, a series of 15-cycle runs will be made with a one-minute rest period between runs. During a run, each cycle of the actual trajectory will be superimposed over the target trajectory. At the end of the run, the average trajectory will be displayed over the target trajectory. Based on this information, the operator will modify the control sequences for the next run. In the second method, changes to the control will be made during a run of several minutes while observing the actual trajectories displayed over the target trajectory in real time.

Using the nominal control determined above, a run of at least 100 cycles will be conducted. The first few cycles will be compared to determine the number of cycles required to reach steady-state. Once steady-state is attained, average trajectories and confidence intervals (95% probability) will be calculated along the trajectory for M successive cycles where M=3, 5, 10 and 20. These data will be used to determine the minimum number of cycles that must be averaged to yield an acceptable estimate of the trajectory. Preliminary data suggests that this number will be 10 cycles or less (see Preliminary Results section). Using this number, average trajectories and confidence intervals will be calculated throughout the 100-cycle run to determine the effect of fatigue and other factors that may fluctuate with time.

A series of short runs will be performed (using the minimum number of cycles required to compute the average trajectory as determined above), with a 30-second rest between runs. On the first run, the nominal control will be used. On successive runs, one of the control pulses will be perturbed by $\pm\Delta, \ldots, \pm k\Delta$ where k is an integer and Δ is a small increment in pulse amplitude. The average trajectory for each run will be calculated, and the deviation from the nominal trajectory (the average trajectory measured with no perturbations) will be calculated. This process will be repeated for various pulses throughout the cycle. Perturbations of specific pulses will be repeated during this test to determine the consistency of the responses over time. Additional runs will be made in which two or more pulses are perturbed.

Even though the system is nonlinear, there will be a range (perhaps small) in which the deviations caused by perturbing the control pulses is linear; i.e., 1) the deviation caused by $k\Delta$ is approximately k times the deviation caused by Δ and 2) the deviation caused by perturbing two or more pulses is the linear sum of the deviations caused by individually perturbing each of the pulses. Data from this study will determine the range over which the linear approximation is valid and will provide information required for the computational techniques to be derived.

PRELIMINARY RESULTS

Preliminary data has been collected to demonstrate the feasibility of the methods described above. In both of the tests described below, the subject is positioned as shown in Figure 2 so that the knee is flexed approximately 40° when the subject is relaxed (neutral position). Knee angle is referenced to the neutral position with extension from neutral measured positively.

The subject was asked to follow a periodic trajectory (2-second period) in which the limb starts at rest in the neutral position, extends to $+30^{\circ}$, swings back to -30° and then returns to neutral (Figure 3). Electromyograms were recorded using fine wire electrodes placed in two muscles of the quadriceps (vastus lateralis and rectus femoris) and two muscles of the hamstrings (semitendinosis and short head of the biceps). Data from three cycles are shown in Figure 3. The rectus femoris was found to be inactive during this maneuver and is not shown. It is seen that the vastus lateralis acts to provide the initial forward-swing (extension). The semitendinosis acts primarily to accelerate the leg during backward-swing. The short head of the biceps assists

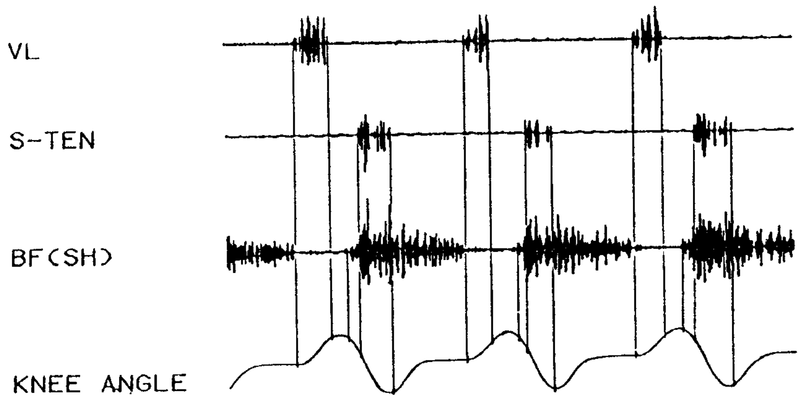

Fig. 3. EMG Data during Voluntary Performance of a Target Trajectory (Modified Sine Wave with a Pause). The pause is at the neutral position (40° of knee flexion). Extension of the knee is indicated by an upward deflection. The magnitude of swing is ±30°. Period of one cycle is 2 sec. Abbreviations: VL, Vastus Lateralis; S-TEN, Semitendinosis; BF(SH), Short Head of the Biceps Femoris.

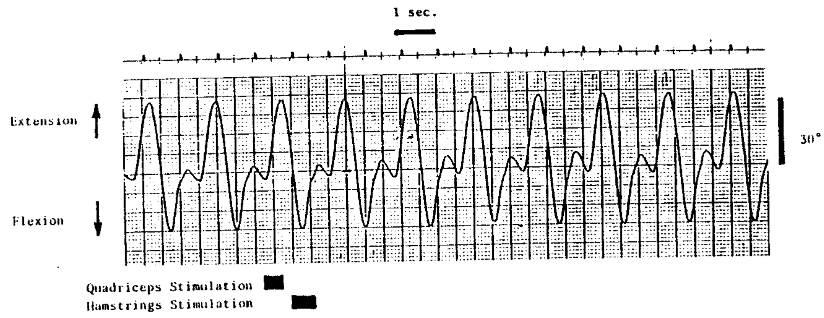

Fig. 4. Knee Angle versus Time with Stimulation

the semitendinosis during the backward-swing phase and then continues to be active while the leg comes to rest and while the neutral position is held until the next cycle is initiated. The inactivity of the rectus femoris is curious, but may be due to the fact that the rectus femoris is the only muscle of the quadriceps that crosses the hip joint and therefore acts as a hip flexor as well as a knee extensor. Since the subject was only attempting to move the lower leg, the subject's neural control mechanism may selectively activate only those muscles that are pure knee extensors. It is also interesting to note that the antagonistic muscles never co-contract during this maneuver.

In a second test, stimulation was used to produce a similar periodic trajectory. An available 2-channel stimulator with constant output was used so only on/off times and the amplitude of each channel could be varied. The result is shown in Figure 4. With the restricted control available, it was impossible to achieve the desired trajectory exactly (e.g., it was not possible to obtain 30° of backward-swing and still bring the limb to an approximately neutral position), but the resulting trajectory is reasonably close to the desired trajectory. Comparing the EMG records of Figure 3 with these results, it is seen that the phasing of the EMG activity and stimuli is similar.

Ten cycles of the stimulated run were averaged near the beginning, in the middle, and near the end of a continuous ten minute run. The average trajectory and plus/minus one standard deviation are plotted in Figure 5. Although there is some change in the average trajectory during the run due to muscle fatigue (peak-to-peak amplitudes dropped by 10% and 19% at the midpoint and end of the run), the standard deviations are all less than 5% of the peak-to-peak amplitudes. Confidence intervals were computed for points along the trajectory using data from the ten cycles near the beginning of the run. The estimate of each point, averaged over the entire trajectory, was accurate to +0.5° with probability 0.95. The worst case (which occurred during the backward-swing) was still accurate to +0.9°.

DISCUSSION

The research described above is based upon previous results in FES. In 1977, Stanic et al (7) showed that improved performance could be achieved by modulating stimulation to produce a graded response in the stimulated muscle. Ankle dorsiflexors of two hemiplegic patients with footdrop were stimulated with on-off control and with graded control, and it was shown that the ankle motion resulting from graded control was closer to that which is seen in normal gait. Following this study, graded control has been incorporated into most clinical programs,(8-11) but selection of the proper open-loop sequences becomes a major problem as the number of stimulation channels increases.(12)

An analytical method for calculating the control to produce a desired movement has been proposed by Bajd and Trnkoczy.(12) This method was applied successfully to single channel control of ankle dorsiflexion, but is limited for more general applications

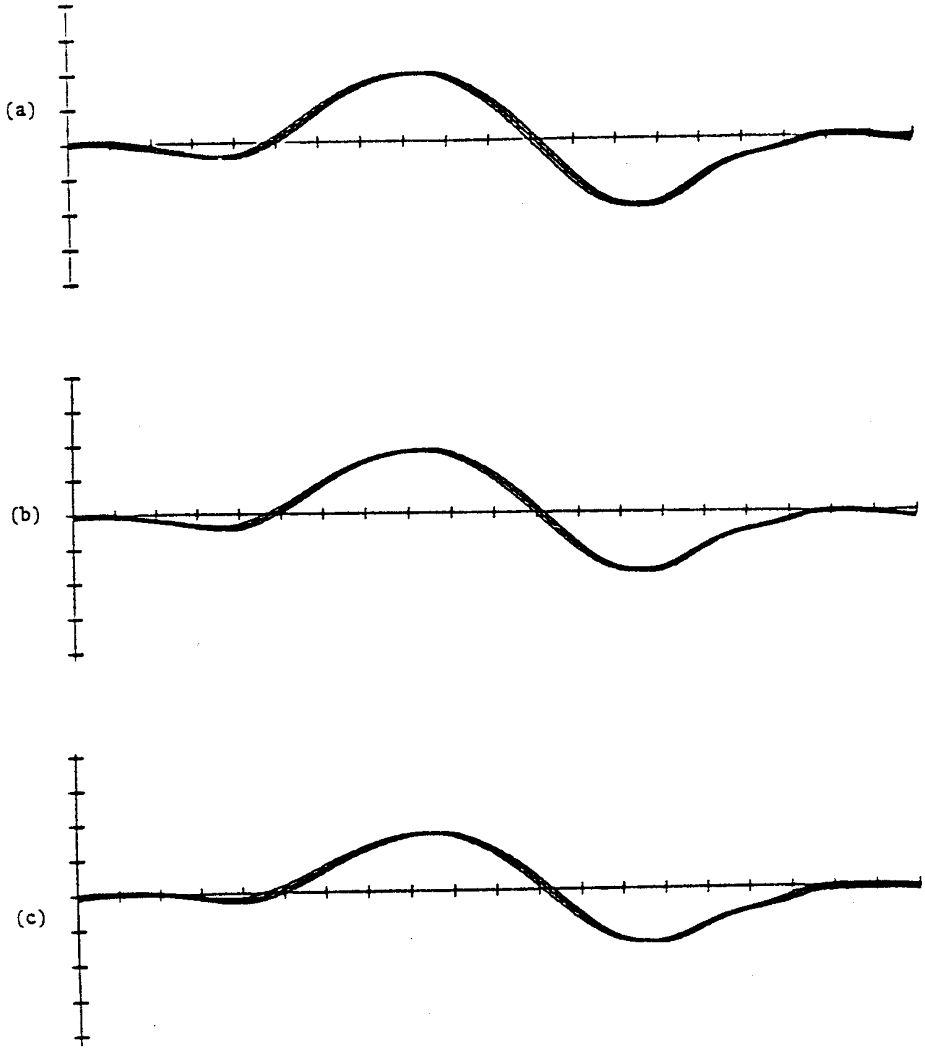

Fig. 5. Average Trajectories Plus/Minus One Standard Deviation. Average of 10 Cycles. (a) Beginning of a continuous 10-minute run. (b) Middle of run. (c) End of run.

by their assumption that the system to be controlled is linear.

Feedback control has not been used in any clinical application of FES assisted walking. In all cases, preprogrammed stimulation sequences are initiated and, in some cases terminated, by heel lift-off or heel strike. Closed-loop control has been studied, however, in laboratory experiments by a number of investigators.(1,6,13-15)

Trnkoczy recorded the isometric response of the ankle dorsiflexors to step inputs in eight hemiplegic patients and concluded that the variability in response was so great using the equipment available to him that integral feedback control would be totally unsatisfactory.(1) The variable responses were attributed to the sensitivity of muscle response to electrode position. Results were similar using either cutaneous electrodes or RF-coupled receiver/stimulators that were implanted near the peroneal nerve.(16)

Feedback control of the gastrocnemius muscle in cats under isometric (14) and isotonic (15) conditions has also been investigated. In the isometric studies, the objective was to maximize the length of time during which constant tension could be maintained. At tensions below 20% of maximum, constant tension was maintained for more than 5 minutes, but endurance times were less than 30 seconds for tensions greater than 50% of maximum.(14) In these tests, sequential stimulation was applied to the ventral roots of L6, L7 and S1.

Feedback was used in the isotonic studies to maintain a constant velocity of shortening.(15) During these studies, oscillations were observed under no-load conditions, indicating system instability, but simultaneous stimulation of the antagonist muscle was shown to reduce the observed oscillations. A cuff electrode placed around the sciatic nerve was employed for stimulation.

Intramuscular electrodes (17) placed into cat soleus muscle have been used to study isometric torque responses to step and ramp inputs.(6) Proportional control was found to be inadequate because the high values of gain required to compensate for system variability resulted in instability, however, stability was achieved by adding integral control. Although response times decreased with increasing integral gain, they were found to be adequate at gains that provided good compensation for system nonlinearities and variability.(6) Repeatable performance could not be ensured over a long time period, however, because of daily variations in muscle input-output properties,(18) and a digital identification scheme to automatically adjust the controller parameters is now under investigation.(19)

It is expected that the research to be performed in this project will extend the results of these previous studies and produce numerical techniques for controlling the lower extremities to significantly improve the ambulation level of paralyzed individuals.

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