A NEW SYSTEM FOR PROVIDING INDIVIDUALIZED, MULTI-MODE, A/K PROSTHESES

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SUMMARY

During the past several years, two amputee-interactive simulator systems have been used in a variety of A/K prosthesis control studies. These systems allow an amputee to move about in a laboratory wearing a prosthesis which is configured like a "normal" prosthesis, but is controlled by a computer via an umbilical line. The laboratory systems have been used to demonstrate semi-automatic, real-time optimization of a knee torque profile for a particular individual and mode of locomotion. Also a finite state approach has been used to automatically switch active and passive prosthesis control characteristics. These studies have shown that the desired control characteristics for A/K prosthesis knee mechanisms vary not only from individual to individual, but also change as the amputee changes mode of locomotion. Recent advances in microcomputer technology have been combined with the use of a proportional electromechanical brake to provide a prototype of such a versatile prosthesis. This prototype prosthesis is controlled by a purse-sized, battery-powered microcomputer, and has been used by an amputee subject both in the laboratory and outdoors. The amputee's use of this prototype has re-emphasized the need for multi-mode control. This project aims toward demonstration of hardware and software to develop an optimum prescription for an A/K prosthesis as an amputee moves about in the laboratory. That prescription would then be transferred from the laboratory computer to a microcomputer in the prosthesis so that the patient could leave with a personalized, multi-mode prosthesis.

1.0 INTRODUCTION

The research described in this paper began with the development of an amputee-interactive simulator system which allows an amputee subject to use a simulator prosthesis for gait evaluation in a laboratory setting (2). This simulator prosthesis is physically very similar to a conventional prosthesis, but contains an electrohydraulic knee torque control mechanism capable of exhibiting a wide range of active and passive behavior. While being controlled by a stationary computer via an umbilical line, the prosthesis can simulate the dynamics of a prosthesis which is described by a computer model, but which is not necessarily existent as hardware. Through use of this system, control characteristics for improved knee mechanisms are being sought.

Experience with the electrohydraulic system and a similar, passive system pointed out the importance of being able to adjust the prosthesis characteristics for the individual amputee and to suit their current activity. The need for individualized knee mechanism control characteristics seems to be as fundamental as the need for prescription eyeglasses.

Thus, the project changed its focus from the development of an "ideal" knee mechanism to the development of "personalized", multi-mode knee mechanisms and methods for custom tailoring these devices to their user and his/her current activity.

Recent progress in microcomputer technology and the development of an electrically controllable knee mechanism have made such prostheses feasible and possibly practical. The following sections describe progress toward realization of such prostheses and prescription systems.

Section 2.0 describes research concerning multi-mode prosthesis control and the possible benefits of providing limited, active prosthesis behavior. Section 3.0 discusses computer-augmented development of swing phase control characteristics for an individual amputee and Section 4.0 describes a prototype, microcomputer-controlled prosthesis which can be "tuned" by its user.

2.0 MULTI-MODE A/K PROSTHESIS CONTROL

To be acceptable, a multi-mode prosthesis must make reliable, smooth transitions from one control mode to another. The multi-mode approach taken here uses finite state (7) techniques to recognize when a given function is required and then switches in the necessary control scheme for that function. The finite state intent recognizer shown in Figure 2.0-1 will use a number of inputs from the man/prosthesis system to recognize which locomotion mode the amputee intends to perform and which functional mode is required at a particular time. In addition to the finite state intent recognizer, the multi-mode approach uses a flexible proportional controller for each functional mode. The approach has been tested in a control scheme for level walking and work is underway to include additional locomotion modes.

An active control scheme for level walking was developed and tested (4) using the instrumented electro-hydraulic A/K prosthesis simulator. The control

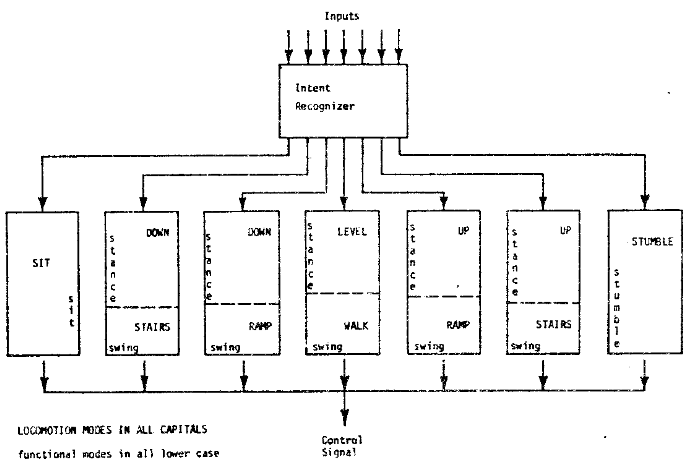

Figure 2.0-1 A Multi-Mode Control Scheme (3)

scheme used two functional modes and controlled the prosthesis to approximate the normal knee position pattern. During most of the stance phase, a position feedback loop controlled the prosthetic knee joint, while torque feedback was employed at the end of stance and during the swing phase. By controlling the position of the knee during the stance phase, stability during weight bearing and the desired position output were attained. Position control throughout most of stance was followed by torque feedback through the remaining portion of the gait cycle. This torque feedback loop controls the prosthesis to simulate a damped pendulum with variable damping factors which depend upon the state of the man/prosthesis system. A finite state approach was used to select the appropriate (swing or stance) controller. Results of this study are shown in Figure 2.0-2. This data is from an amputee subject wearing the electrohydraulic prosthesis simulator under active control. The prosthesis simulator with active control provides a smooth knee position profile which is very close to the position pattern followed by a normal knee. The amputees had favorable comments about the active control scheme and preferred the active scheme to their conventional prostheses. Examination of the power curve shows the energy feasibility of active control for level walking. By integrating the power curve with respect to time to obtain the energy required at the knee, one can see that the net energy required is considerably negative. Even with an inefficient storage and release system, it may be possible to store energy in some parts of the cycle and release energy where required in other parts of the cycle. In such a system, external power sources would not be necessary.

This initial success has led to additional research on active A/K prosthesis control. Research is currently underway to develop active control schemes for different locomotion modes (3). This work will utilize the multi-mode

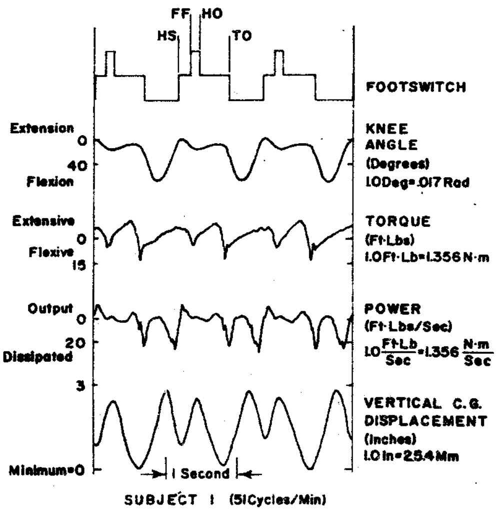

**Figure 2.0-2 Gait Parameters:
Level Walking Active Control (4)**

approach to study level walking, walking up ramps, and climbing stairs. Swing control will vary with locomotion mode and will control the prosthesis to simulate a pendulum with appropriate springs and dashpots. Stance control will use "echo" control to duplicate the angular position of the sound knee. See Figure 2.0-3. This "echo" control scheme controls the angle of the prosthetic knee to mimic the previous cycle of the sound knee. This method of control would be individualized, cadence responsive, and does not require the storage or calculation of a set of angle trajectories for weight bearing. Each locomotion mode and then functional mode would be recognized and the appropriate controller would be chosen automatically. The overall scheme will be tested with the assistance of amputee subjects who will use the scheme in locomotion trials. These trials will assist in the evaluation of multi-mode A/K prosthesis control as well as determine the practical feasibility of active knee control for A/K prostheses in different locomotion modes.

3.0 SWING PHASE DAMPING PROFILE INDIVIDUALIZATION

Several damping profile optimization schemes have been developed for implementation with the passive simulator system (1)(6). Currently, two complementary techniques for profile development are being used in determining the viability of various optimization criteria. A PDP 11/40 mini computer functions as the prosthesis controller and provides off line graphics capabilities for the display of data sampled from the amputee/prosthesis system.

The first profile development technique employs the use of light-pen

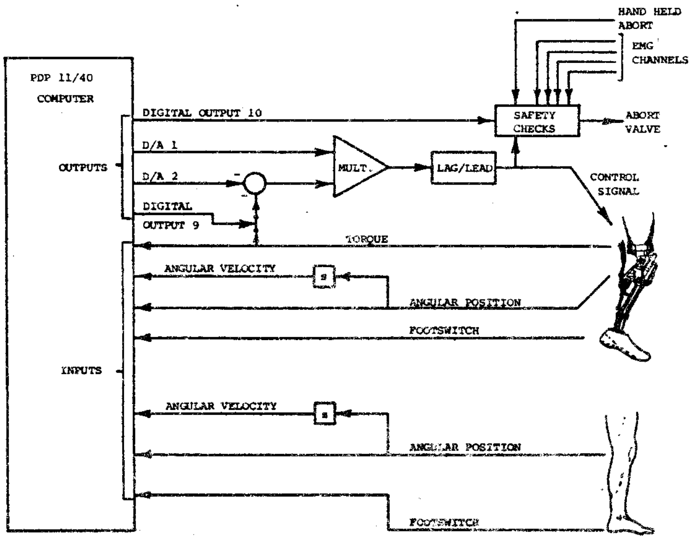

Figure 2.0-3 Hybrid Control System Schematic

interaction with the CRT display terminal. The computer operator is able to draw a damping function $\bar{B}(\theta, \text{Sgn } \dot{\theta})$ on the CRT to be used in a torque control algorithm such as:

$$\text{Torque} = \bar{B}(\theta, \text{Sgn } \dot{\theta})\dot{\theta}^2, \text{ where } \theta = \text{knee angle.}$$

A control program uses this damping function to command the torque output of a magnetic particle brake which controls the prosthesis knee mechanism. Velocity squared damping was chosen to simulate the turbulent fluid damping employed on many conventional prostheses. During execution of the control program, data is recorded from angular position and velocity for both legs, foot/floor timing information for both legs, and prosthetic knee torque. After voluntarily exiting this program, data for either leg can be displayed as a function of time or in the form of a phase plane trajectories. Such physical measurements and subjective feedback from the amputee can be used to optimize the performance of the prosthesis.

Another individualization technique was suggested by experiments with the light pen programs. Automatic damping profile optimization is possible if control criteria can be quantified and stored in the computer. An experiment performed using this technique used modified phase plots from the amputee's sound leg as control criteria for the prosthesis. An error generated by differencing the phase plots from the sound and prosthetic leg was used to automatically update a damping factor array, $\bar{B}$, in an attempt to cause convergence of the phase-plane trajectories. A convergence constant scaled the error array before summing to the existing damping factor array to temper the effect of a single update. Figure 3.0-1 shows several of the damping factor arrays and associated phase plots (to be read CCW with the origin as heel contact). Convergence progressed quite well except in the region governed by knee buckling near the end of stance. In that region, the amputee would not allow the computer controlled prosthesis to impose the kinematics of his sound leg on the prosthesis. The reason for this lack of convergence is not yet clear.

Data sampled from the amputee's own prosthesis has also been used as a control criterion. In that case, convergence was straightforward, but the "optimized" prosthesis merely mimiced the conventional prosthesis. These studies have shown, however, that automatic or semi-automatic damping profile development is feasible. Current research aims toward determination of relevant optimization criteria.

4.0 PROTOTYPE MICROCOMPUTER-CONTROLLED PROSTHESIS

Results from experiments conducted with the passive simulator system indicated that the electrical power requirements for the prosthesis during level walking were quite low and suggested the feasibility of a completely portable prosthesis. With this idea in mind, a prototype of a portable prosthesis-controlled system was developed (6).

A new passive prosthesis knee mechanism was constructed which weighs 3 lbs 3 oz and has an overall length of 16 inches. See Figure 4.0-1. A magnetic particle brake was again used as a source of resistive torque, but the transmission mechanism was changed.

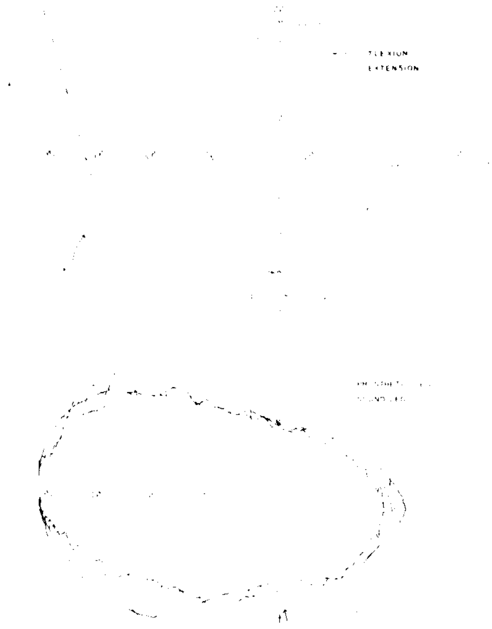

Figure 3.0-1 Initial and Final Damping
Profiles and Associated Phase Plots

The later model utilizes a ball-screw and ball-nut combination connected to a lever arm to act as a torque transformer. Motion about the knee axis is translated into a change in length of the ball screw mechanism, which in turn rotates the shaft of the magnetic particle brake. This arrangement results in an overall "gear ratio" of 40:1 (compared to 7.5:1 for the earlier version) between rotation about the knee axis and rotation of the brake shaft. This larger gear ratio allows use of a smaller, lower-power brake.

The prototype is capable of producing resistive torques from 0 to 400 in-lbs but requires only 2.5 watts of electrical power for its maximum torque. Experiments have shown that three 9 volt alkaline transistor radio batteries can supply the energy required for swing phase control for approximately 120 hours of level walking for an average amputee.

The controller for the prosthesis is a purse-sized microcomputer system complete with a lithium battery power supply (also shown in Fig. 4.0-1). It weighs 5 lbs 6 oz and is 13" x 5" x 3". The function of the controller can be readily modified by changing its control porgram. Also, the microprocessor can serve as a real-time data acquisition system in addition to acting as a controller.

The controller is currently programmed to generate a control signal of the same form presented in Section 3.0. However, in this case the damping function,

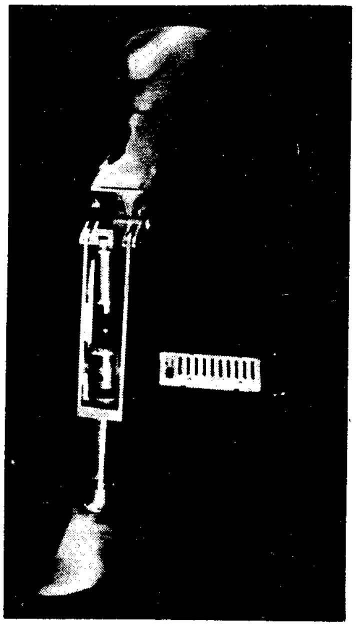

Figure 4.0-1 Microcomputer-Controlled A/K Prosthesis

$\bar{B}$, can be quickly input to the controller by means of a series of slide potentiometers or by a serial data interface from a remote computer.

The prosthesis is connected to the controller with a small cable which provides angular position information to the processor and control current to brake.

The system has been evaluated by an experienced amputee in the laboratory and outdoors and has received a positive reaction. A study is currently underway to evaluate the applicability of the prototype as a training device for recent amputees during their early ambulation training.

As mentioned in Section 2.0, improved gait can result if a small amount of energy were available to make a prosthesis behave in an active mode. Work is currently underway to design such an active/passive prosthesis. The ball screw transmission utilized in present prototype enables the easy incorporation of an active torque source. A motor/generator device could also be connected to the ball screw to make the prosthesis capable of active motion. When a passive characteristic is desired, the motor could serve as a generator to absorb energy from the knee as well as aid in recharging batteries.

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