Feasibility of an Active Control Scheme for Above Knee Prostheses

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The level walking process for an above knee (A/K) amputee with a conventional prosthesis greatly compromises the amputee's mobility. The fact that conventional prostheses lock in hyperextension during the stance phase, in contrast to the extend-flex-extend pattern during stance for the natural limb, has been suggested as a source of the amputee's uncosmetic gait and high energy expenditure. While the lock in hyperextension during stance provides stability to prevent buckling, it requires the person to vault over the prosthetic limb. This vaulting during level walking may cause higher vertical displacements of the body center of gravity (c.g.) and accompanying higher energy requirements for the amputee. This investigation employs an amputee-interactive prosthesis simulator system to evaluate the viability of controlling the prosthetic knee joint to follow a normal knee position pattern. In order to insure that the amputee-interactive prosthesis simulator system does not introduce gait anomalies, the system was controlled to simulate a conventional prosthesis. This showed that the simulator system has no undesired side effects since data from walking trials with the simulator system in "conventional prosthesis mode" are very similar to data from conventional prostheses in the literature. Then, an active position control scheme which controls the prosthetic knee joint to follow a normal knee position pattern was tested by two young, active amputees in level walking trials. The subjects experienced very little difficulty in walking with the active control scheme and preferred the simulator with the active control scheme to their conventional prostheses. Measured knee power requirements for the scheme indicate that this type of control is feasible without external power sources. However, measurements of the vertical displacement of the body c.g. show little difference between gait with the active control scheme and gait with a conventional prosthesis. It appears that the increased energy requirements for A/K amputees are not due in total to the lack of the extend-flex-extend position profile at the prosthetic knee joint.

Introduction

Some of the most frequently encountered difficulties that an A/K amputee must face occur during the level walking process. The level walking process for an A/K amputee requires more energy, is typically asymmetric and uncosmetic, frequently uncomfortable, and sometimes hazardous. The origin of these difficulties is the inability of the prosthetic replacement to duplicate normal leg functions. As a result, other body segments must adjust to compensate for the prosthetic segment in order to attain dynamic stability and similar function. These adjustments away from normal lead to level walking patterns which significantly affect the amputee's mobility. For A/K amputees adjusting body segments cause larger displacements of the body c.g. Larger vertical displacements of the body c.g. indicate higher energy consumption during locomotion [1, 2, 3, 4].[1] Metabolic studies of energy consumption have shown that level walking for an A/K amputee requires 20 percent to 100 percent more energy than normal gait [3, 5, 6]. This study investigates the feasibility of returning one function of the prosthetic segment to normal. More specifically, the prosthetic knee joint is controlled to follow the normal knee position pattern during level walking.

Extensive research has been directed toward A/K prosthesis design in order to alleviate some of the endurance, safety, comfort, and cosmesis problems of amputees. This research has led to some generally accepted design requirements for conventional knee mechanisms. In order to be effective such a mechanism

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[1]Numbers in brackets designate References at end of paper.

must provide stability during the stance phase and some damping during the swing phase. Most conventional A/K prostheses achieve stability during the stance phase by means of alignment stability. In this case the amputee causes the prosthesis to lock in full extension by exerting an extensive torque on the knee joint. Other conventional "stance control" prostheses either increase damping at the knee during weight bearing or are polycentric linkage devices which dynamically contribute to alignment stability. All conventional prostheses are essentially locked in extension during the stance phase of level walking to insure stability and prevent buckling. During the swing phase of level walking, damping is provided by dashpots or by dry friction to prevent excessive heel rise and abrupt stops at full extension. Swing phase damping may be assisted by (*a*) springs to aid the initiation of flexion at the beginning of the swing phase, and/or (*b*) springs to initiate extension at maximum heel rise.

The gait pattern of an A/K amputee wearing a prosthesis which locks in the stance phase and has damping in the swing phase is significantly altered from normal. Fig. 1 compares knee position, torque, power, and vertical body c.g. motion as a function of percent of gait cycle for normal subjects and typical A/K amputees. Fig. 1(*a*) clearly shows the prosthetic knee locked in extension through most of the stance phase. In contrast, the knee of a normal subject progresses through an extend-flex-extend process during stance. The position profiles through the swing phase can be very similar depending upon the swing control provided by the prosthesis. Normal and prosthetic knee torques are shown in Fig. 1(*b*). In order to insure alignment stability, prosthetic knee torque is extensive through most of stance. If the knee torque became flexive a conventional prosthesis would collapse during stance. Again the largest differences occur during the stance phase. Fig. 1(*c*) shows power requirements for the knee. Prosthetic knee power is zero while the knee is locked in stance. The two dissipative peaks in the last half of the cycle depend upon the damping provided by the prosthesis and can be similar to normal. If knee power is integrated through time over a cycle, the net energy is negative for both normal subjects and for amputees wearing conventional A/K prostheses. For the prosthesis there is no energy required, only energy dissipated. This is what would be expected from a prosthesis whose swing phase is controlled only by a damper since a damper cannot produce a power output. For the normal knee, although the knee actuators must provide energy in some parts of the cycle, the energy that must be dissipated is greater than the energy required. In a typical cycle for a normal, the energy dissipated is 3 to 5 times the energy required at the knee. Vertical body c.g. displacements are shown in Fig. 1(*d*). These c.g. displacements were obtained using a TRACK (Telemetried Real time Acquisition and Computation of Kinematics) position measuring system in the Mobility Laboratory of the Mechanical Engineering Department at MIT [8]. The TRACK system consists of a Selspot position monitoring system, a PDP 11/40 computer, a direct memory access interface, and a data requisition and analysis software package. High points in the c.g. trace occur during mid stance and low points take place during double support. For a normal subject the c.g. pattern indicates the symmetry of the two strides in a cycle and has an amplitude of approximately 2 in. (5.08 cm). The larger c.g. displacements (approximately 3.5 in. (8.89 cm)) for the amputee are an indication of higher energy consumption. Distinct asymmetry between the natural and prosthetic sides is evident.

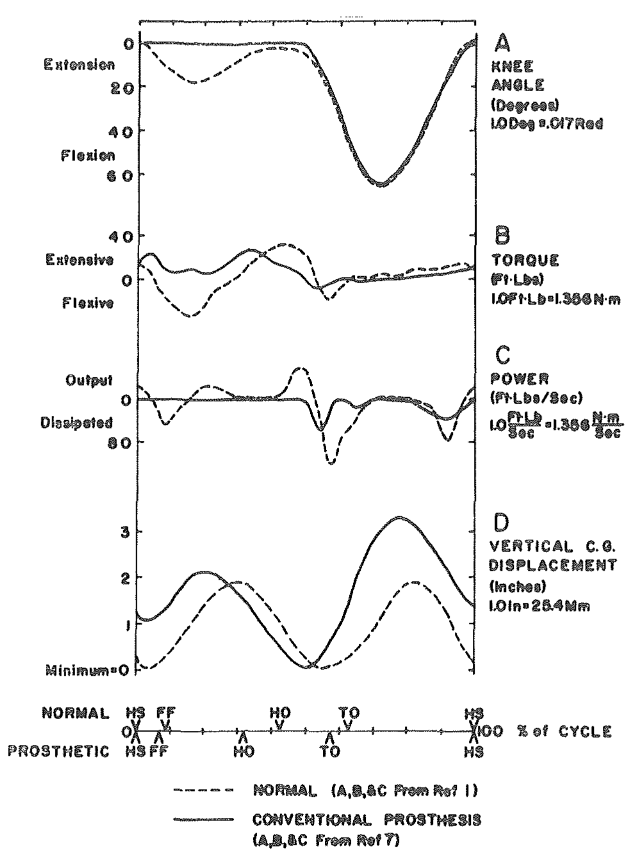


Fig. 1 Knee parameters

These four traces exhibit some of the differences between the gait pattern of A/K amputees and normal subjects. Some of the largest differences occur during the stance phase. The prosthetic knee joint does not exhibit the extend-flex-extend position profile but remains locked in stance. The extend-flex-extend process allows the normal knee to act to absorb energy and is the knee's contribution to minimizing the vertical displacement of the body c.g. [1]. The amputee, however, must vault over his locked prosthesis. This lack of the extend-flex-extend pattern at the prosthetic knee joint has been proposed [2, 3, 4, 7] as a reason for higher c.g. motions and higher energy requirements. This study investigates the feasibility of an active control scheme in which the prosthetic knee joint follows a more normal extend-flex-extend position pattern during stance. Very little research has been directed toward the area of positive power (active) prosthesis control due to the lack of a facility to adequately study such control. This study employs an amputee-interactive prosthesis simulator system [9] which is ideally suited to study positive power A/K prosthesis control.

An Amputee-Interactive Prothesis Simulator System

Man-interactive prosthesis simulation has been introduced as an alternative to conventional design or typical computer simulation [10]. Such a technique employs simulation of a part of the prosthesis but requires the human to provide the complex natural reactions and inputs to the prosthetic system. In the case of A/K prosthesis simulation, an actual prosthesis, the dynamics of which are electronically controlled, is worn by the amputee as he/she performs prescribed tasks. Unlike typical simulation techniques, the human is retained in the system to provide the natural interface with the prosthesis, human flexibility, and subjective feedback about prosthesis performance. Mathematically modeling these factors for pure computer simulation is very difficult. As opposed to traditional hardware design, the dynamics of the prosthesis can easily be controlled and changed to facilitate flexibility, both in the revision of an idea and in the number and scope of ideas tested. Also the prosthesis simulator is instrumented to allow more thorough quantitative analysis of a prosthetic design. Although the system does not incorporate field hardware design, it can be of great use in de-

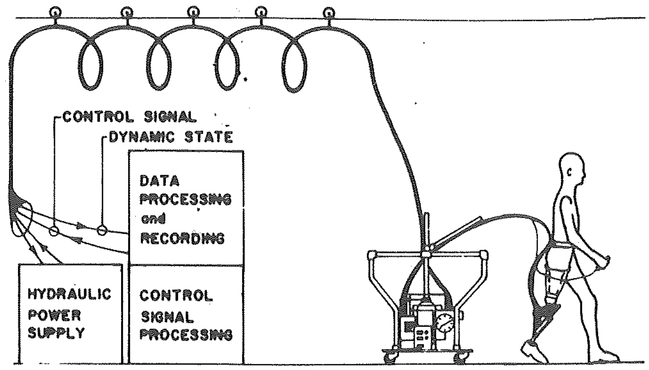


Fig. 2 Overall system layout

termining what should be included in that design. Without the time and money costs inherent in traditional design, development, and manufacture, man-interactive prosthesis simulation allows more detailed feasibility studies of a number of ideas in an attempt to determine what functions a prosthesis should and/or can provide for an amputee.

The A/K prosthesis simulator system used in this study is an electrohydraulic servo-system. A hydraulic system was chosen because it has the speed, accuracy, and power necessary to perform most leg functions while the actuator is consistent with size and weight limitations of A/K prostheses. A stationary pump and relief valve provide a 1000 psi pressure source (Fig. 2). The pressurized fluid then flows through hydraulic lines to a cart near the prosthesis. There, the fluid is filtered, an accumulator damps possible pressure transients, and the fluid pressure is measured. Also on the cart as a safety measure is an amputee controlled abort valve. The amputee carries an emergency abort switch which he can, if necessary, depress to lock the knee. Barring activation of the abort switch, the hydraulic fluid then flows through a small umbilical to the prosthesis. Within the frame of the prosthesis, fluid flow to the actuator is controlled by electrical current to a servovalve. Finally flow in the cylinder controls the velocity of the piston. The prosthesis is also instrumented to measure torque, position, and timing information (Fig. 3). Knee torque is measured by a load cell positioned to measure loads through the piston rod axis. The measured load is proportional to knee torque since the piston rod axis has a constant lever arm and angle with respect to the knee axis. Angular position is measured by a potentiometer. A socket is bolted to the top of the prosthesis and a SACH foot is secured to the distal section. Foot/floor timing information is measured by two strip switches on the bottom of the shoe. With this arrangement it is possible to measure heel strike, foot flat, heel off, and toe off. The voltage signals from these parameters are processed at a remote site to yield additional information. Angular position is differentiated resulting in angular velocity. Knee torque and angular velocity are multiplied to yield knee power. With these methods the quantities are available in real time in contrast to previous calculation methods which required many hours of man and computer time. This allows the signals to be used for feedback and control and permits immediate review of the data.

The final major segment of simulator system hardware deals with processing various inputs to produce a control signal. Knee torque, position, velocity, power, timing information, and myoelectric signals can be used as inputs and processed to produce a signal which controls the dynamic state of the knee. One of the many possible controller models involves the restoration of the extend-flex-extend position pattern at the knee during level walking. However, before any new scheme can be satisfactorily evaluated using the simulator system, it must be determined whether any gait irregularities are caused by the simulator system itself.

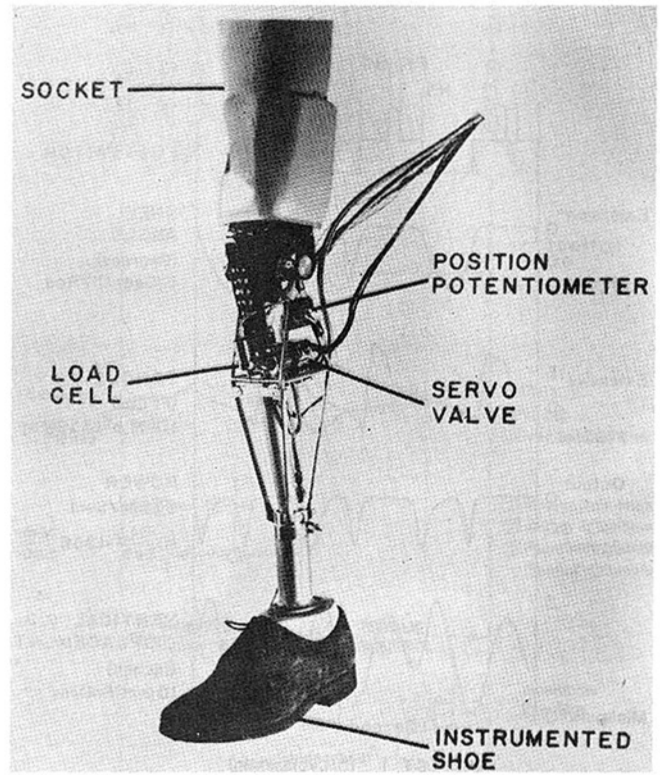


Fig. 3 Prosthesis simulator instrumentation

Conventional Prosthesis Mode Results

To determine whether the umbilical to the prosthesis simulator or other lab equipment introduce any gait anomalies, the simulator system was controlled to approximate the damping in a conventional prosthesis. If walking trials using the prosthesis simulator system in this control mode produce data which are similar to previously published data for conventional prostheses and if amputee comments are in agreement, then the simulator system is adequately modeling a conventional prosthesis and causing no undesired effects. In order to control the prosthesis simulator to behave like a conventional damping control prosthesis a torque feedback loop is used.

With torque feedback the prosthesis moves in response to torques measured by the load cell. In this configuration the prosthesis can be made to behave as a pendulum with viscous (i.e., proportional to velocity) damping ([9], Mode 3). The damping is also a function of the gain of the torque feedback loop. High gains produce low damping and fast response and low gains produce high damping and sluggish response. Although this gain factor can be modulated by any input during any part of the cycle, the control scheme described here uses a constant gain until the final section of the swing phase. Then, in order to avoid impact at full extension and to smooth the transition from swing to stance, additional damping is affected by dividing the torque gain by a damping factor. This damping factor is proportional to knee angle and increases as the prosthesis approaches full extension. Using this control scheme it is possible to tune the simulator system so that the amputee subject perceives the "damping" in the simulator system as he walks to be approximately the same as the damping in his conventional prosthesis.

Walking trials were conducted with the assistance of two A/K amputees. Both amputees are male, approximately 20 yr

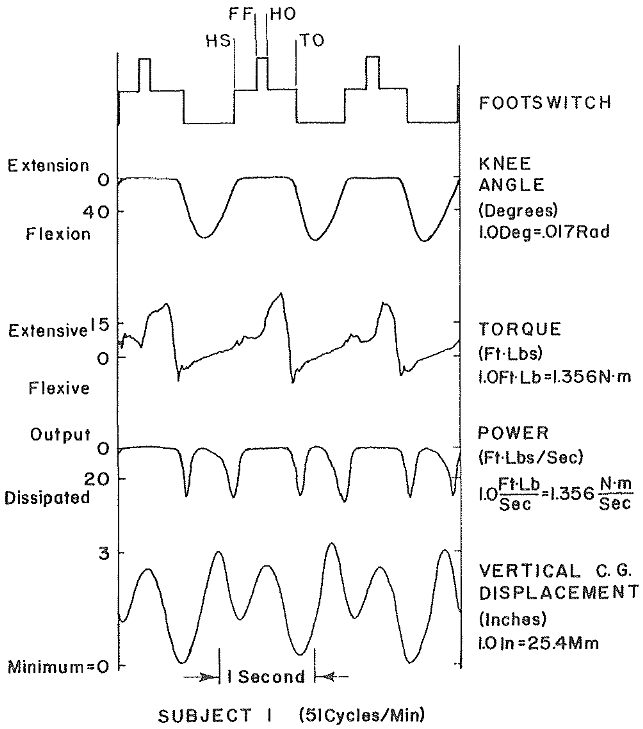


Fig. 4 Gait parameters: conventional prosthesis model

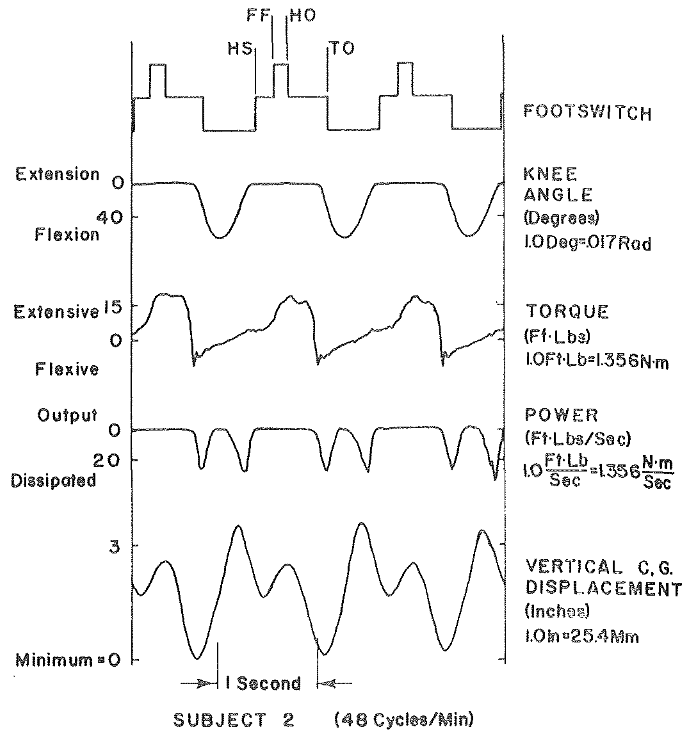


Fig. 5 Gait parameters: conventional prosthesis mode

old, active, and without other health complications. After several practice trials an amputee subject was asked to walk a straight path of approximately fifty feet while data from the simulator and c.g. data were being recorded. Data from each subject are shown in Figs. 4 and 5. These data are very similar to the conventional prosthesis data in Fig. 1. Comparison of Figs. 4 and 5 with Fig. 1 illustrates:

1 timing information from the footswitch is similar;
2 knee positions are characterized by no position change in stance and show a maximum flexion of about 70 deg;
3 knee torque is extensive in all three curves through the first 50 percent of the cycle, clearly showing the amputee's desire to achieve alignment stability during weight bearing;
4 knee power curves show two dissipative peaks in the final half of the cycle and never become positive (damper controlled);
5 all three c.g. traces exhibit the same asymmetrical patterns.

In addition to these results each amputee also reported that the prosthesis simulator felt like his conventional prosthesis. These comments, along with the similarities in the parameters in Figs. 4 and 5 and Fig. 1 lead to two conclusions. First, the prosthesis simulator system in "conventional prosthesis mode" provides a good approximation of a conventional damping control prosthesis. Second, the prosthesis simulator system does not introduce any undesired gait irregularities. The second conclusion is necessary before any new control scheme can be meaningfully evaluated using the amputee-interactive prosthesis simulator system.

The Active Control Scheme

The active control scheme in this study uses two separate control modes to approximate the normal knee position pattern on the prosthetic knee joint. During the stance phase a position feedback loop controls the prosthetic knee joint while torque feedback is employed during the swing phase. By controlling the postion of the knee during the stance phase, stability during weight bearing and the desired position output are attained. Fig. 6 is a simplified model of the system with a position feedback loop. This particular method of position control was chosen because it produces the desired output while requiring only simple analog circuitry.

In the model the dynamics of the hydraulic simulator are included in G. With a first order low pass filter in the forward path, the system transfer function is

$$\frac{\theta_{\text{out}}}{\theta_{\text{in}}} = \frac{K/G\tau}{s^2 + \frac{1}{\tau}s + \frac{K}{G\tau}}$$

If the gain and time constant of the filter are properly tuned the dynamics of the filter dominate the dynamics of the hydraulic simulator and G may be approximated as a constant. In this case the transfer function is second order with the undamped natural frequency

$$\omega_n = \sqrt{K/G\tau}$$

and the damping ratio

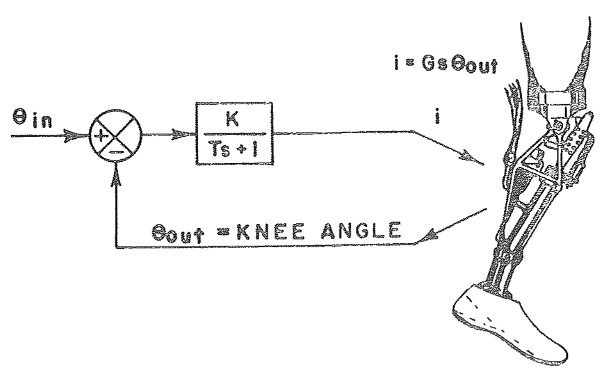


Fig. 6 Simplified block diagram of simulator system under position control

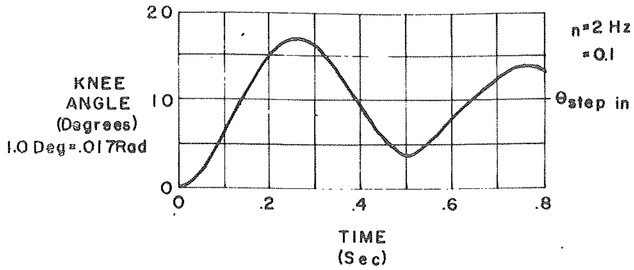


Fig. 7 System response to a step input

$$\zeta = \frac{1}{2\sqrt{K\tau/G}}.$$

System response to a step input can then be used to approximate the flex and extend portions of stance phase in normal gait (Fig. 7). If the step-input position control scheme is activated at heel strike, when the leg is fully extended (extend), it will overshoot (flex) and then undershoot (extend) the step input. This method of stance-phase control provides the necessary stability during weight bearing, has the desired position output without having to store a position profile for the cycle, and can be adjusted to allow implementation of different stance-phase position profiles. Position control through most of stance is followed by torque feedback through the remaining portion of the gait cycle (Fig. 8). The torque feedback scheme used for swing phase control is described earlier in this paper.

If two control modes are to be used (i.e., position control through most of stance and torque control through the end of stance and all of swing), the inputs must again be employed to determine when each control mode is to be in effect (Fig. 9). The first and most obvious timing function in the gait cycle comes from the shoeswitch. The change from torque feedback to position feedback is triggered at heel strike using the heelswitch on the shoe. Position control is in effect whenever the heelswitch is closed. Since the knee begins to flex in anticipation of swing phase between heel off and toe off, the system must at that time be switched back to torque feedback. Attempts were made to switch from position control to torque control at heel off and toe off; but each attempt led to undesirable position and torque outputs, a rough transition between position and torque control, poor gait, and unfavorable comments from the amputee. The method finally adopted for switching from position to torque control employs the plantar switch and the torque signal. While the plantar switch is closed and the torque signal is extensive, the system is in position control mode. When the plantar switch is closed and the torque becomes flexive, the system switches to torque feedback and allows the leg to swing in response to input torques. This method provides proper timing and guarantees a smooth transition between position and torque control since the torque equals zero when switching occurs. A final section of position control is added between heel off and the desired position to torque transition. In some normal cases the torque is flexive between 25 percent and 35 percent of the cycle, immediately following the time for amputee heel off. *If* implementation of this system resulted in the same amputee footswitch timing pattern, and *if* the simulator torque approached the torque of a normal, the knee would buckle immediately following heel off. To avoid this situation, a 100-ms time delay is added after heel off during which the system is in position control. Thus, position control is in effect:

1 when the heelswitch is closed *or*
2 100 ms after heel off *or*
3 when the plantar switch is closed *and* external knee torque is extensive.

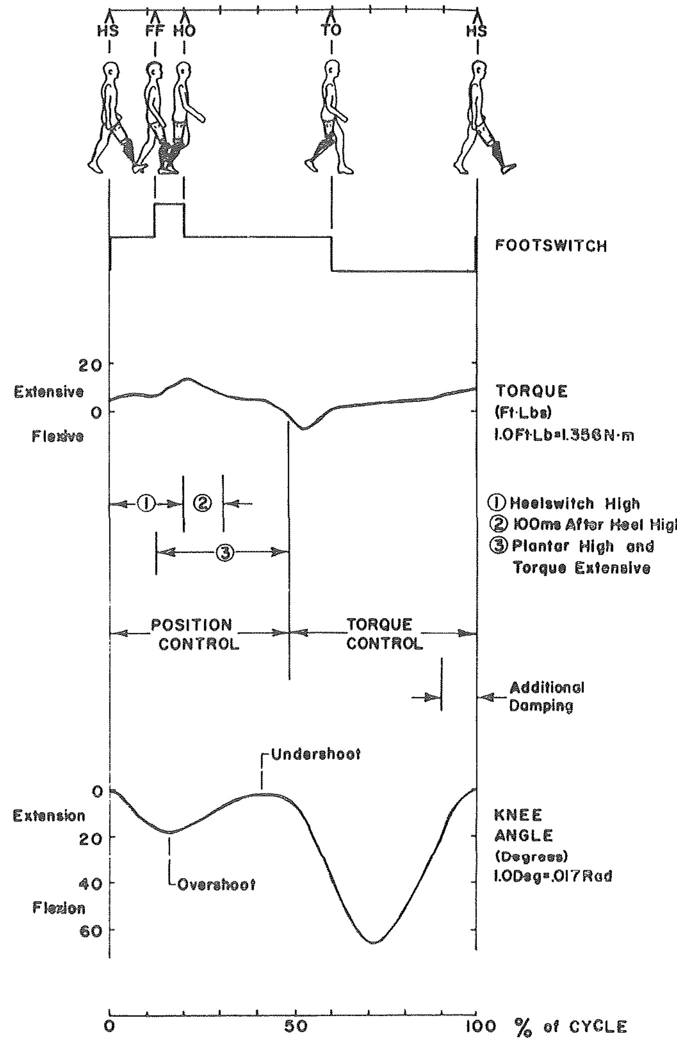


Fig. 8 The active control shceme: cycle timing and control

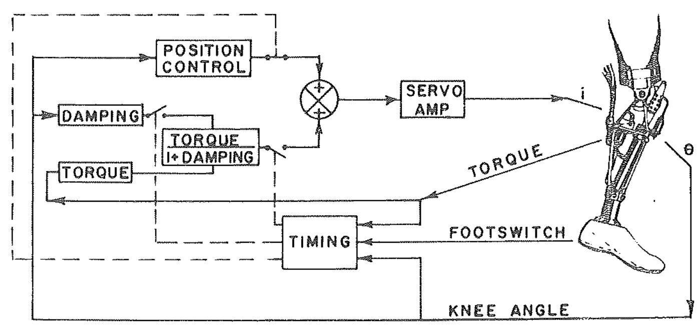


Fig. 9 The active control scheme

When none of these conditions exists the system is in torque feedback mode. In a walking cycle, torque feedback begins when knee torque becomes flexive, at about 50 percent of the cycle, and lasts through swing phase until heel strike.

In review of the cycle (Fig. 8) the system becomes a position controller at heel strike. At that point, the knee is in full extension and a step input is applied to the position controller. The knee angle changes in response to the input signal as it flexes (overshoots) and then extends (undershoots). In this phase the amputee has gone through double support and the weight-bearing phase of gait. Then, while the plantar switch is still closed, knee torque becomes flexive and the simulator switches to torque feedback. At this time the gait cycle begins the final part of stance phase, double support. At the beginning of torque feedback mode the knee starts to flex in anticipation of swing phase. The simulator is in torque feedback mode throughout swing phase. The knee reaches maximum flexion at approximately 70 percent of the cycle and then extends as it is deployed for stance. When the knee is approximately 20 deg from full extension, damping is increased proportionally to knee angle. The shank is decelerated and full extension is reached at approximately the same time as heel strike. The damping smooths the transition between swing and stance, and the entire cycle begins again.

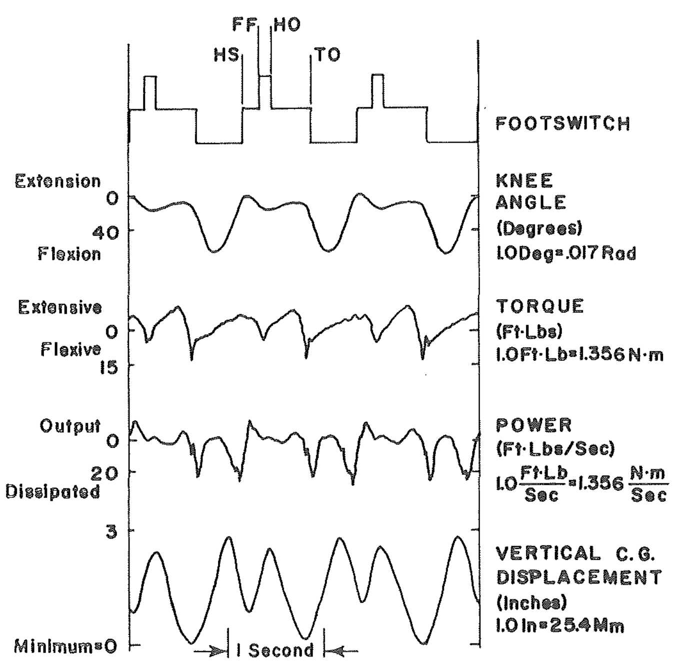
This system approximates the extend-flex-extend position profile in stance phase of normal gait and allows standard swing phase performance. It is cadence responsive with respect to control mode timing, but the step size, natural frequency, and damping ratio in the stance phase must be set prior to an experimental run. One should note that this control scheme was designed solely for the purpose of studying stance-phase control in level walking and might not be applicable in any other mode of locomotion. With these qualifications noted, results are presented in the next section.

Active Control Scheme Results

The active control scheme was tested in walking trials by the same two amputees who assisted in the previously described experiments. Young active amputees were chosen as subjects for these experiments because they belong to that part of the amputee population who might benefit most from active control schemes. At the beginning of each experimental session the active control scheme was implemented gradually to ease the transition between a conventional prosthesis and the simulator with active control. The input step size was slowly increased over a period of approximately ten runs of 50 ft each in order to allow the amputee to adjust to the extend-flex-extend cycle in stance. Both amputees were able to walk smoothly with the active control scheme within a few minutes. No significant difficulties were encountered. After the amputee felt comfortable in the warm-up period, walking trials were conducted and data were recorded.

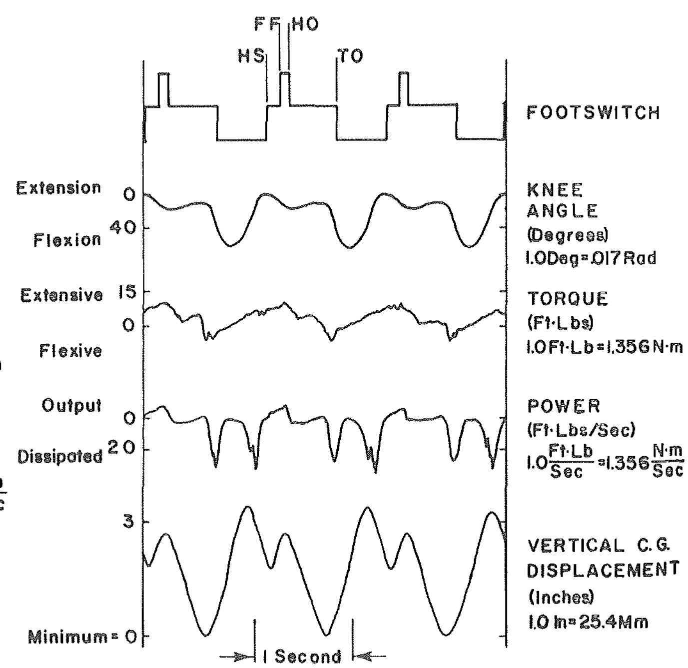
Figs. 10 and 11 show data from each subject. The prosthesis simulator with active control provides a smooth knee position profile which is very close to the position pattern followed by a normal knee. The amputees voiced a definite preference for a maximum flexion in stance of 16 deg to 20 deg and a subsequent extension to 2 deg to 5 deg. Maximum flexions greater than approximately 25 deg led to difficulty in walking and unfavorable comments from the amputee. The footswitch pattern remains similar to the conventional prosthesis pattern except that foot flat and heel off occur earlier in the stance phase. When the knee begins to flex at heel contact the plantar section of the SACH foot is forced downward. As a result, foot flat occurs earlier in the cycle. As the subject proceeds toward midstance, the flexed knee causes the heel of the SACH foot to lift off the floor earlier than if the knee were locked in hyperextension. Stance time/swing time ratios remain the same as for conventional prostheses. Both natural and prosthetic stride lengths are essentially unchanged. Knee torques during the stance phase are considerably less with the active control scheme. These torques are reduced because the lever arm for the foot/floor reaction force vector is decreased by flexing the knee at the end of stance. Knee power is positive only through two short parts of the cycle with the active control scheme. Maximum power output is approximately 15 ft·lb/sec (20.34 N·m/s). The most important feature of the power curves is obtained by integrating power with respect to time to obtain the energy required at the knee. This net energy for an average walking cycle is negative 5 ft·lb (6.78 N·m). Although energy is required in two short parts of the cycle, the net function of the knee is that of an energy dissipator. Not only is the net energy required for a cycle negative, but the gross energy dissipated is approximately four times the energy required. This indicates that 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.*

The vertical displacement of the body c.g. shows no change between gait with a conventional prosthesis and gait with the



SUBJECT 1 (51Cycles/Min)

Fig. 10 Gait parameters: active control



SUBJECT 2 (48 Cycles/Min)

Fig. 11 Gait parameters: active control

prosthesis simulator system in active control. Other A/K gait irregularities mask the effect of the extend-flex-extend knee position pattern in stance. C.G. patterns for both amputees show a higher displacement during natural limb midstance than during prosthetic midstance. These peaks in c.g. displacement during natural limb midstance result from the amputees' attempts to insure prosthetic toe clearance during prosthetic swing phase. Position patterns at the prosthetic knee during stance phase could not be expected to alter this gait irregularity which occurs during prosthetic swing phase. There is also no appreciable difference in the peak displacement of the body c.g. between active control and conventional control during prosthetic midstance. This may be due to hip abduction or the lack of prosthetic ankle dorsiflexion during prosthetic stance phase with the active control scheme. These factors indicate that the extend-flex-extend pattern at the knee may not be as important to energy consumption as previously thought. However, both amputees had favorable comments about the active control scheme and preferred the active scheme to their conventional prostheses. Other gait evaluation techniques should also be employed to help determine changes in wearer performance. In addition to clinical observation, one possible quantiative method of gait evaluation proposes the use of a pattern derived index indicating the relative quality of gait [11].

Summary

An amputee-interactive A/K prosthesis simulator system, which was found to introduce no additional gait irregularities, has been used to investigate a new type of active knee control for A/K prostheses. This active control scheme controls the knee joint of the A/K prosthesis simulator to duplicate the position pattern of a normal knee during level walking. Two amputee subjects had no significant difficulty in walking with the active control scheme and actually preferred its performance to their own conventional prostheses. Power requirements for the prosthetic knee joint show that this active control scheme may be possible without external power sources. However, measurements of the vertical displacement of the body c.g. indicate that no changes have taken place between gait with a conventional prosthesis and gait with the prosthesis simulator system in active control mode for these amputees. In this situation other gait irregularities obscure possible effects of this active control scheme on c.g. motion. Additional gait evaluation techniques are necessary to completely describe gait changes introduced by this active control scheme.

This paper is part of a larger ongoing project aimed at determining the functions an A/K prosthesis should and/or can provide for an amputee. The results presented here have directed a portion of our research activity toward other energy viable active control schemes. For level walking additional swing control may be necessary to abolish the c.g. peak during midstance on the amputee's natural limb. Normal knee kinematic performance from a prosthetic knee joint may not be desirable when a SACH foot is used. Research along these lines is currently being initiated using the amputee-interactive A/K prosthesis simulator system.

Acknowledgments

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