Evaluation of a Self-contained, Microcomputer-controlled, Above-knee Prosthesis

by

Gregg Kiyoshi Motonaga

Submitted to the Department of Mechanical Engineering
in partial fulfillment of the requirements for the degree of

Bachelor of Science

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Author. . . . .
Department of Mechanical Engineering
May 1992

Certified by . . . .
Professor Woodie C. Flowers
School of Engineering Professor of Teaching Innovation
Thesis Supervisor

Accepted by . . . .
Professor Peter Griffith
Chairman, Department Committee

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A series of walking trials were conducted to test the performance of an above-knee prosthesis with the purpose of identifying the dependence of amputee gait on prosthesis performance. In particular, the viscous-damping constants were altered, and specific, gait parameters were measured. The information obtained from the walking trials can be used in the design of sophisticated controllers that will automatically adjust the profile of the knee torques to optimize amputee gait.

Thesis Supervisor: Woodie C. Flowers
Title: School of Engineering Professor of Teaching Innovation

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This thesis is dedicated to

Professor Margaret MacVicar

First dean for undergraduate education and
founder of the Undergraduate Research Opportunities Program at M.I.T.

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Chapter 1

Introduction

Man has tremendous abilities to create advanced machines and devices that perform a variety of tasks. Automobiles, computers, and vehicles that propel humans through space are just a few examples. Development of prosthetic devices that replace the function of the human body has not been as successful, partially because the field of engineering for humans has not been of primary interest, but also because the human body is an extremely complex system whose function and form have been refined through many years of evolution.

The development of an above-knee prosthesis is one such area. Many existing prostheses are deficient in areas concerning weight, energy costs, and dynamic behavior. With advances in computer technology and light-weight materials, future prostheses will be sophisticated devices capable of reproducing proper leg-functions while minimizing the consumption of metabolic energy.

Computer-controlled, above-knee-prosthesis research began at M.I.T. in 1969, when Professor Woodie Flowers designed an electro-hydraulic, man-interactive prosthesis simulator [3]. Flowers' prosthesis, as well as later devices, was developed as a research tool to study amputee gait and methods to better control future, computer-controlled prostheses. Because of their research objectives, however, many employed cumbersome instrumentation that would be impractical for daily-worn prostheses.

The most recent development is a self-contained, microcomputer-controlled, above-knee prosthesis designed by Stuart Schechter [10] and Michael Goldfarb [4]. Because of its portability and low weight, this prosthesis could be worn on a daily basis and has the potential to redefine what the industry now terms a conventional prosthesis. The prosthesis, in its present state, is a laboratory prototype that needs refinement in preparation for commercial use.

The controller is one area that requires further refinement. Since there are many factors affecting amputee gait, the controllers will maintain proper gait by intelligently altering the performance of the prosthesis based on an understanding of the relationship between prosthesis function and desired gait patterns. This thesis intends to provide, for the benefit of future controllers, some insight into the dynamics of amputee gait when using the microcomputer-controlled prosthesis.

The existing controller utilizes finite state control based on specific occurrences during a typical gait cycle, and, therefore, the gait cycle should be briefly discussed to familiarize the reader with some of the terminology. For a more complete discussion of human gait, see Inman [5].

1.1 Normal gait

Human bipedal gait is a harmonious movement of body segments during which the feet are continuously repositioned to support the forward motion of the body. Because of its repetitive nature, gait can be described by distinct occurrences within the cycle. Within a typical gait cycle, each of the limbs will be in one of the two phases of gait known as swing and stance. Figure 1-1 shows a typical gait cycle illustrating the two phases as well as the characteristic landmarks.

The stance phase of gait accounts for approximately 65% of the normal gait cycle, during which the foot is in contact with the ground. The opposite foot may also be in

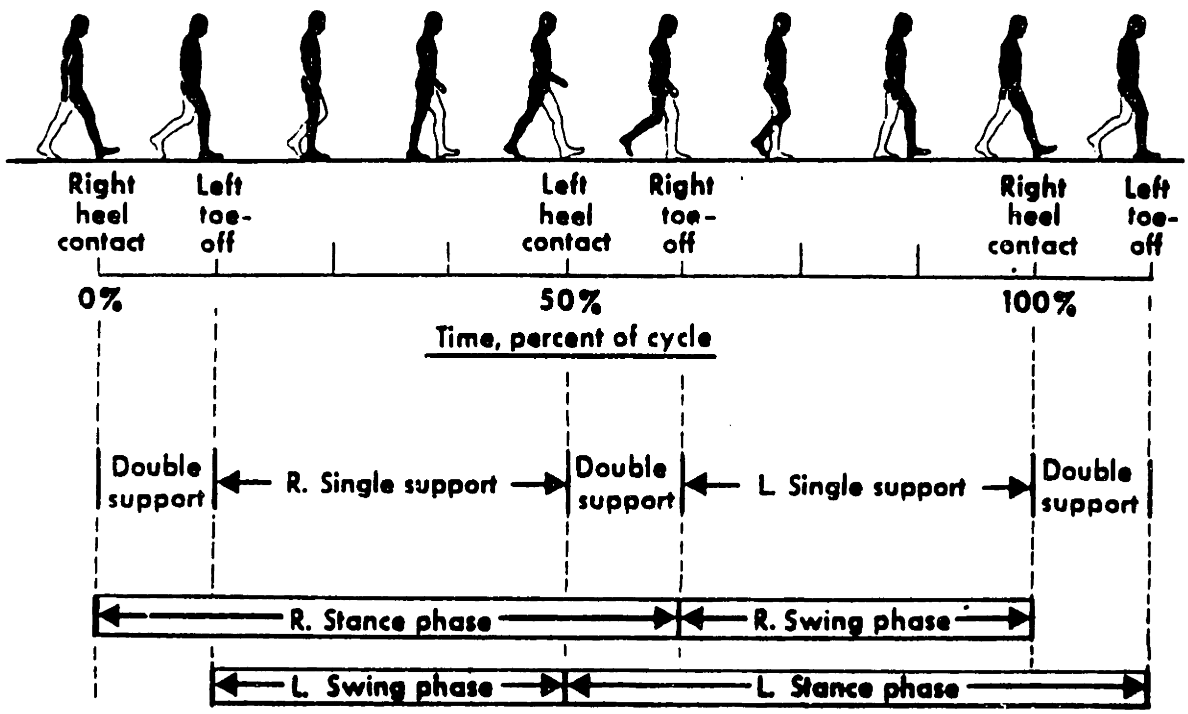

Figure 1-1: Typical gait cycle

contact with the ground, characterizing this portion of the stance phase as double support. The double support phase is approximately 12% of the gait cycle.

The major landmarks that label the stance phase are heel-strike (H.S.), foot-flat (F.F.), heel-off (H.O.), and toe-off (T.O.). Heel-strike is the initiating landmark of the stance phase and occurs just as the foot makes contact with the ground. As the ankle plantarflexes and the foot becomes flat on the floor, the knee flexes slightly to absorb some of the impact from heel strike. As the center of gravity of the body moves over the foot, the knee re-extends, and the heel lifts off the ground. Toe-off is the final landmark of stance indicating when the foot leaves the ground and enters the swing phase.

The swing phase is the portion of the gait cycle when the foot is in the air being repositioned for the next heel-strike. This portion of gait accounts for approximately 35% of the normal gait-cycle. During swing, the flexion of the knee facilitates the passing of the foot beneath the body while preventing the toe from catching on the ground.

Muscle power controlling the knee during a gait cycle can be either positive or negative. The difference between the two is that positive knee power can generate motion of body segments, while negative power can only dissipate the energy of the segments and control their motions. During the gait cycle, positive knee power is found twice, both instances occurring during the stance phase. The first occurs during the flexion/re-extension motion, and the second during heel-off. The knee power during the remaining portions of the gait cycle is negative, providing control of the lower leg through swing and stability of the knee during stance.

1.2 Amputee gait

Amputee gait is not as easily characterized as normal gait. There are several reasons why amputee gait is not only different from normal gait, but also widely variant among amputees. One reason is that destroying the symmetry of the body through amputation may cause one amputee to compensate differently from another depending upon the nature of the amputation, such as the location of the amputation along the femur.

In addition, amputee gait depends largely on the dynamic characteristics of the particular prosthesis being worn. There are several available prostheses with dynamic characteristics ranging from a free-swinging, simple pendulum to a hydraulically-damped knee-mechanism and an articulated ankle. Obviously these two prostheses, with disparate dynamic characteristics, will affect the user's gait differently. In addition, attachment of the prosthesis to the amputated appendage can also affect amputee gait. For instance, many amputees alter their gait when their prostheses cause discomfort or pain.

Despite the difficulties associated with describing amputee gait, common phases of gait still exist, regardless of the type of prosthesis used. For instance, before the swing phase of the prosthetic leg, there is toe off of the prosthetic foot. In addition, before heel strike, the prosthetic leg swings forward to full extension momentarily before heel strike. Also, most conventional prostheses are passive devices that do not supply the knee joint with positive power, but rather with negative power, using dissipative elements. Therefore, such prostheses cannot perform the double flexion commonly seen in normal gait, but rather keep the knee rigidly locked throughout stance. The consequence of locking the knee during stance is that the amputee must vault over the locked prosthesis. Many amputees will compensate for this vaulting action by a slight rotation of the pelvis to minimize the vertical C.G. displacements. For amputees using prostheses without knee mechanisms, the typical pattern that is observed during gait is a circumduction of the prosthesis during swing.

Chapter 2

A Self-Contained, Microcomputer-Controlled, Above-Knee, Prosthesis

A microcomputer-controlled prosthesis offers several advantages over conventional prostheses that use strictly mechanical methods of knee control such as hydraulic damping. As was mentioned earlier, a conventional knee-unit has a fairly defined dynamic-pattern, whereas a computer-controlled unit possesses the ability to manipulate its dynamic pattern in response to the varying factors influencing gait. More specifically, controller parameters being used by the on-board microprocessor can be altered to fine-tune the leg to behave optimally for the particular user. The remainder of this chapter provides a brief description of the prosthesis.

2.1 Hardware

The major mechanical components of the prosthesis include the dissipative element, sensors, and the shank. Of these components, the sensors and the dissipative element are the most significant to its intelligent operation. The sensors provide the microprocessor with information about the state of the prosthesis, while the dissipative element provides the resistive torque at the knee. The shank gives the prosthesis its weight-bearing structure and provides structural housing for the weight-bearing sensor. Figure 2-1 shows a photograph of the prosthesis without the socket and the foot.

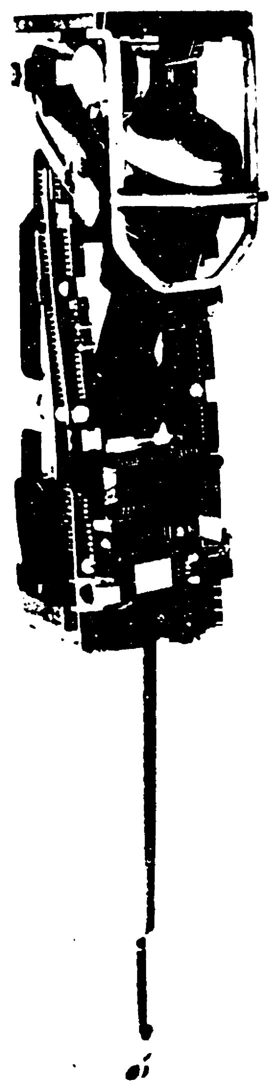

Figure 2-1: Prosthesis without socket and foot (Goldfarb)

A magnetic particle brake is the dissipative element that was chosen for this prosthesis because of its fast response, light weight, and compact size [4]. During stance the brake supplies enough torque to lock the knee, and during swing the brake provides the necessary torque to allow proper knee flexion and extension before heel-strike. The particle brake is mechanically linked to the knee joint by a lead-screw such that the inner disc of the brake is rotated as the knee joint is flexed. The current flow through the brake regulates the applied torque. Being a passive device, the brake lacks the ability to supply the positive power to the knee necessary for the flexion/re-extension motion found during the stance phase of normal gait.

Other necessary components to the performance of the prosthesis, but not inherent to its design, are the socket and the prosthetic foot. The socket provides the interface between user and prosthesis, and the foot supports the distal end of the prosthesis while absorbing shock at heel-strike. Most sockets are suction sockets that create a vacuum around the amputated appendage. Although the socket and foot are only attachments, they nonetheless play an important role in the overall performance of the prosthesis. The suction socket and the Seattle foot, currently attached to the prosthesis, were fitted and aligned by Amtower Biokinetics.

The prosthesis is powered by a 14 volt, rechargeable, Nickel-Cadmium battery-pack which has sustained the prosthesis for over four continuous hours during a walking session with the subject.

2.2 Controller

An on-board microprocessor receives information from the aforementioned sensors that measure position, velocity, and torque about the knee, and axial load in the shank. Based upon the combination of sensory inputs, the controller determines the state, or mode, of the prosthesis within the gait cycle and outputs the proper damping control at the knee.

The controller categorizes the gait cycle into five distinct modes based on the characteristic landmarks of gait . The breakdown of the gait cycle into modes is illustrated in figure 2-2. Modes one and two correspond to single support of stance, during which the brake locks the knee. Mode three allows a slight flexion of the knee to initiate swing. The transition from mode three to four is made at toe-off, when all of the weight is removed from the prosthesis. Mode four represents the majority of the swing phase, for which the particle brake modulates the torque to appropriately control the leg. Mode five heavily damps the prosthesis during the final degrees of extension preceding heel-strike.

The parameters of the mode-four algorithm are described in further detail because they are particularly relevant to the experiments. The knee torque for swing is determined by the following equation:

$$T_{knee} = \beta * \omega * \left[\frac{t_o}{\Delta t + t_o}\right]^2, \tag{2.1}$$

where β is the damping constant, ω is the angular velocity of the knee joint, and the squared quantity is the speed-adaptive factor.

The damping constant used by the controller is not a true damping constant, but rather a proportional scaling-factor for the knee torque. The values for the damping constants range from 0 to 255, where 255 is the maximum damping factor. The damping constants will be given as a percentages of the maximum value of 255.

The speed-adaptive factor changes the knee torque as a function of walking speed. The duration of the stance phase, Δt, provides a measure of the walking speed, and the parameter, t_o, controls the sensitivity of the factor to speed. As the amputee increases walking speed, the stance time decreases, resulting in a higher value for the speed-adaptive factor and, consequently, higher knee torques.

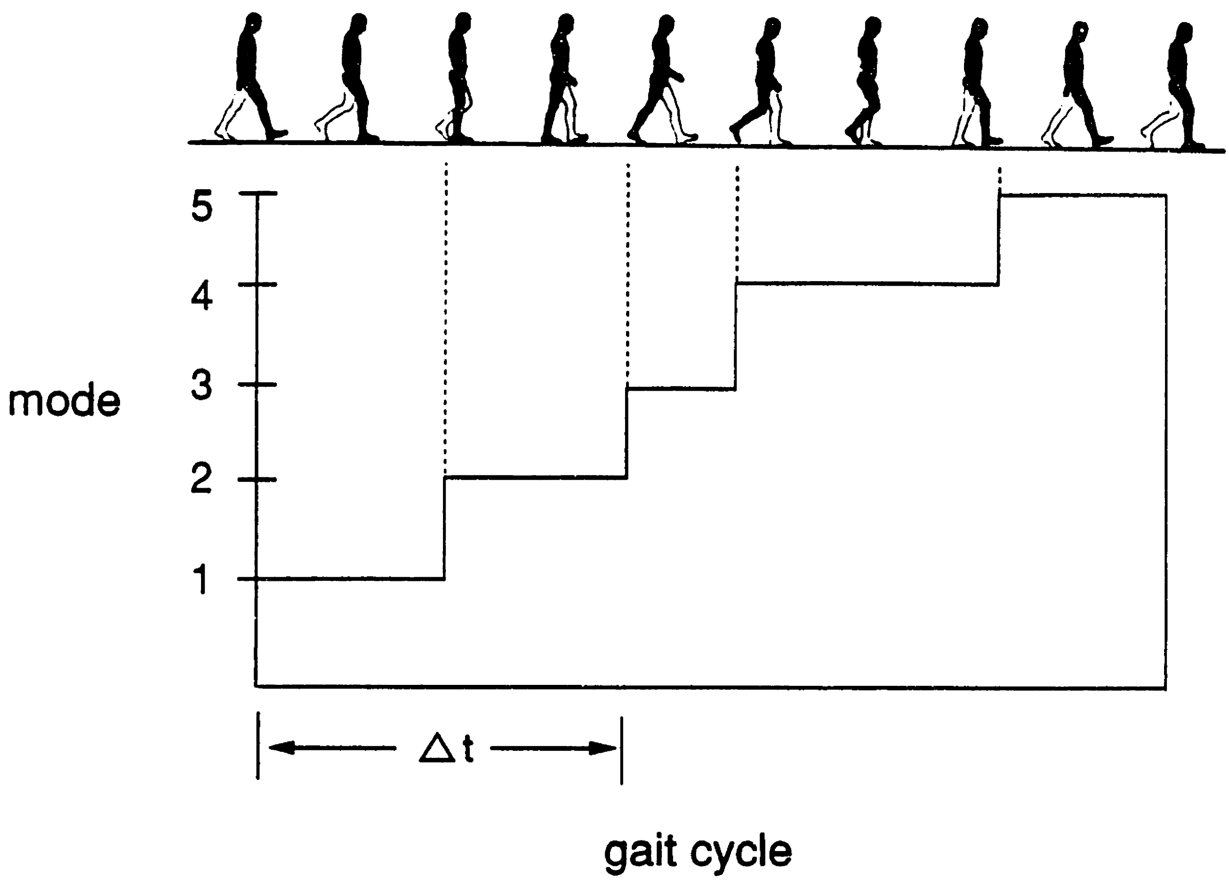

Figure 2-2: Diagram illustrating the controller modes for a typical stride (Goldfarb)

2.3 The Host Computer

Simple connection to a host computer facilitates interaction with the prosthesis. Communication between the computer and the prosthesis allows the controller parameters to be reviewed and altered. Figure 2-3 shows the prosthesis connected to the host computer. Several adjustable parameters are used by the microprocessor to customize the behavior of the knee, and the values of these parameters can be adjusted from the host computer. The process of altering the parameters and downloading them to the prosthesis takes only a few seconds.

In addition to the adjustment of controller parameters, the host computer is used to review the data that is stored by the microcomputer. Each of the on-board sensors records their information to a static-ram chip that resides on the prosthesis. The host computer can then port the information from the prosthesis into the computer for inspection. This information, as well as parameter files, can also be stored to disk for future reference.

Figure 2-3: Prosthesis connected to the host computer

Chapter 3

Walking Trials

The controller enables the prosthesis to intelligently adjust its behavior to suit the amputee. Consequently, refining the controller and increasing its sophistication are likely to improve amputee gait. The refinement of the knee controller is not strictly a theoretical pursuit, but one which must consider the experimentally-observed factors that affect gait. In addition, the verbal feedback provided by an amputee helps to identify deficient areas of control. The following experiments examined the interplay between dynamic behavior of the prosthesis and gait.

The subject used for the walking trials was a 55-year-old male who sustained an above-knee amputation of his left leg approximately nine years ago. He stands six feet, one inch tall and weighs approximately 185 pounds. The subject uses his Mauch hydraulic S-N-S unit with an articulated ankle for a few hours each day.

3.1 Experiment 1

The first experiment was conducted to measure the maximum flexion angles of the knee as a function of walking speed. For these walking trials, the maximum flexion angles of the microcomputer-controlled prosthesis, as well as those of the subject's Mauch prosthesis, were measured as a function of walking speed. A direct comparison between the sound-knee angles and the prosthetic-knee angles provides a basis for measuring prosthesis performance. The controller parameters for the microcomputer-controlled prosthesis were adjusted until the subject was satisfied with its characteristics at several speeds.

The protocol for the experiment was to walk between two markers placed 25 feet apart. The subject was asked to walk between the two markers at several walking speeds ranging from slow to fast. The velocity was calculated from the time taken for the subject to walk between the two markers. The protocol was repeated twice, first with the Mauch prosthesis, and second with the microcomputer-controlled prosthesis.

Experimental data for the microcomputer-controlled prosthesis were obtained by storing the knee-position data from the on-board sensor to the static-ram chip. Because neither the sound leg nor the subject's Mauch prosthesis had data storage abilities, goniometers were attached to the legs to record the knee-position data.

3.2 Experiment 2

The second set of walking trials focused on user-selected, speed-adaptive behavior. Therefore, the speed-adaptive algorithm of the prosthesis was eliminated for this experiment so that, at each distinct velocity, the damping level at the knee would be strictly a function of the knee velocity and damping constant. The subject walked between two markers placed 25 feet apart, and the velocity was calculated from the time taken for the subject to walk between the two markers.

Viscous damping constants were altered between 130 (51%) and 155 (61%), and, at each damping level, the subject walked at several speeds until the speed that resulted in the most comfortable gait was isolated. The results of this experiment indicate the subject's preferred damping constants over a range of walking speeds. Figures 3-1 and 3-2 are photographs of the subject walking during this experiment.

Figure 3-1: Subject walking with microcomputer-controlled prosthesis

Figure 3-2: Subject walking with microcomputer-controlled prosthesis

3.3 Experiment 3

The third experiment focused on relating changes in lower-extremity kinematics with alterations made to the damping profile of the knee for a constant walking speed. The damping constant was systematically altered and gait parameters were measured.

The experiment was conducted at the Biomotion Laboratory at Massachusetts General Hospital in order to take advantage of a high-resolution, kinematic-data-acquisition system known as TRACK. The Newman Laboratory for Biomechanics and Human Rehabilitation at M.I.T. also operates a TRACK system; however, the system records data from one direction, whereas the system at the Biomotion Laboratory is actually two, horizontally-opposed systems which allow information from both sides of the body to be simultaneously recorded. The TRACK system has a limited region, known as the viewing volume, in which data can be recorded. In order to take advantage of the small volume, the tests were initially going to be conducted using a treadmill, allowing the subject to walk continuously in the viewing volume. However, it was found that treadmill ambulation changes the dynamics of gait significantly enough to outweigh the benefits of continuous gait-cycles [10].

The TRACK system uses Selspot infrared cameras that view arrays of infrared-light-emitting diodes (irLEDs) to define the six-degree-of-freedom kinematics of the arrays, from which the system converts the array position and orientation data into body segment position and orientation data [9]. Several arrays of irLEDs are secured to the body at the major segments which include the feet, calf, thigh, pelvis, trunk, arms, and head. Figure 3-3 shows the computer model of the body segments in addition to the placement of the arrays on those segments. It is possible that the added bulk of the instrumentation caused an unknown variation to the subject's regular gait pattern.

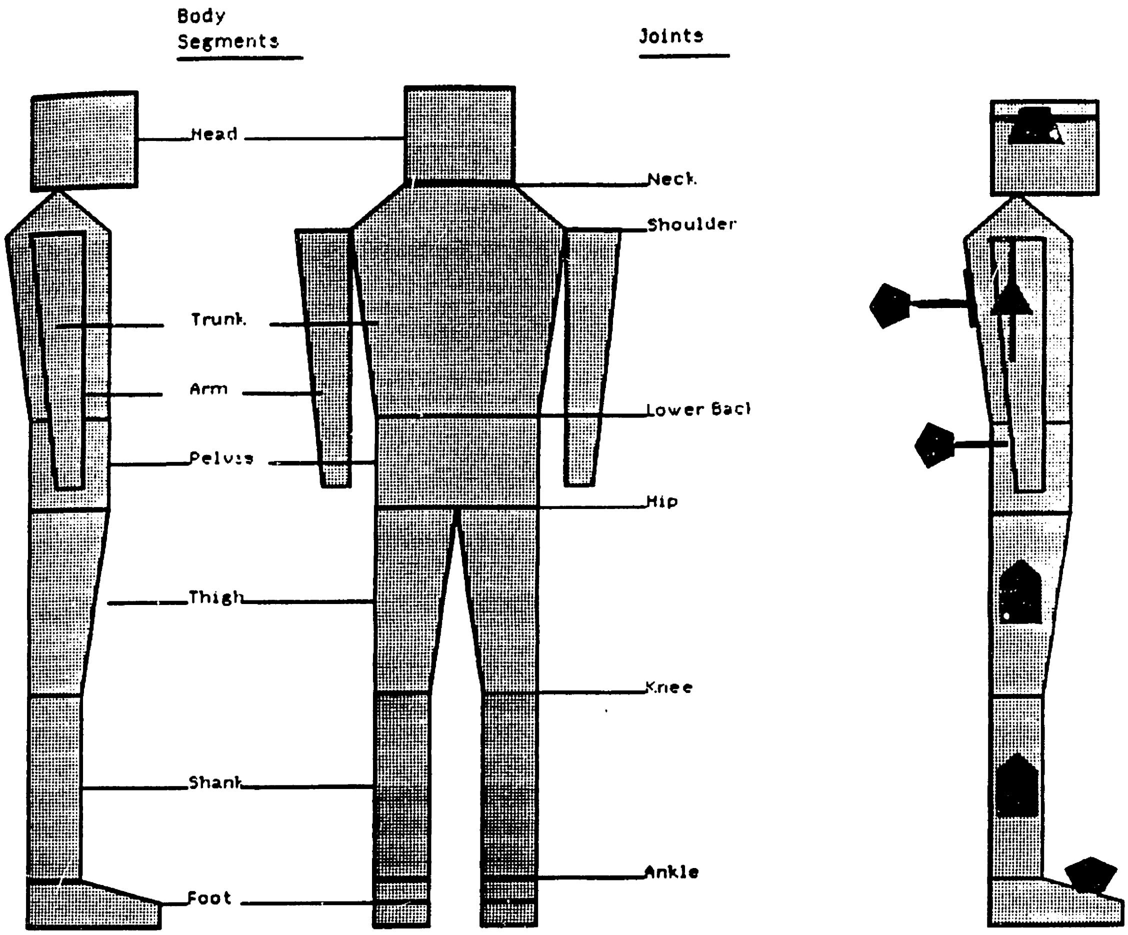

Figure 3-3: Computer model of body segments and array locations

Although a comfortable cadence of 94 beats per minute (BPM) was selected for the experiment, the subject commented that the cadence was too fast for the short walking area in the Biomotion Laboratory, and a cadence of 80 BPM was found to be more acceptable. A metronome was used to pace the subject at 80 BPM, which corresponded to a walking speed of approximately 1 m/s. The subject reported that a damping constant of 135 (53%) was the most comfortable at that cadence. A region of damping constants was bracketed around the comfortable setting beginning at 90 (35%) and increasing to 190 (75%). In addition, data were obtained for the two extreme cases, one during which the power was off, and the other during which the knee was completely locked. Table 3.1 provides a listing of the parameters used in the experiment and indicates their percentage of the maximum damping value of 255. At each damping level, the subject made a total of three passes across the lab. The first pass was used to allow the subject to adjust to the new damping setting. The second and third passes were recorded by the TRACK system.

Toward the end of the walking trials the head array slipped slightly, and the thigh array on the subject's sound leg began to slip downward on the thigh. Both arrays were approximately repositioned at the time the slipping occurred.

Damping Constant Value	% of Max Damping Max Value = 255
90	35 %
110	43 %
125	49 %
130	51 %
135	53 %
140	55 %
150	59 %
170	67 %
190	75 %

Table 3.1: Listing of the damping constants used for experiment 3 and their respective percentages of the maximum damping constant.

Chapter 4

Results and Discussion

4.1 Experiment 1

The results from the first walking experiment are shown graphically in figures 4-1 and 4-2. The sound leg maintained maximum flexion angles between 75 and 80 degrees, and there was a slight increase in the flexion angles as the walking speed increased. The flexion angles of the sound leg appeared to be approximately ten degrees higher than flexion angles typically seen in normal gait. It is possible that the subject was compensating with the sound leg in such a way that increased his knee flexion angles during swing.

The microcomputer-controlled prosthesis achieved angles ranging from 57 degrees at slower speeds to 70 degrees at higher speeds. These angles more closely represent the maximum flexion angles typically seen in normal gait. The Mauch prosthesis has knee flexion angles well below those of the sound leg at slow velocities. Only at very high walking speeds did the flexion angle approach 60 degrees.

The flexion angles of the microcomputer-controlled prosthesis were approximately 12 degrees less than those of the sound leg across the range of walking speeds. The Mauch prosthesis generated a more dramatic rate of increase in flexion angles than either the sound knee or the microcomputer-controlled prosthesis. Although the subject commented that the microcomputer-controlled prosthesis performed well, he did not object to the lower flexion angles of his Mauch prosthesis at slow speeds.

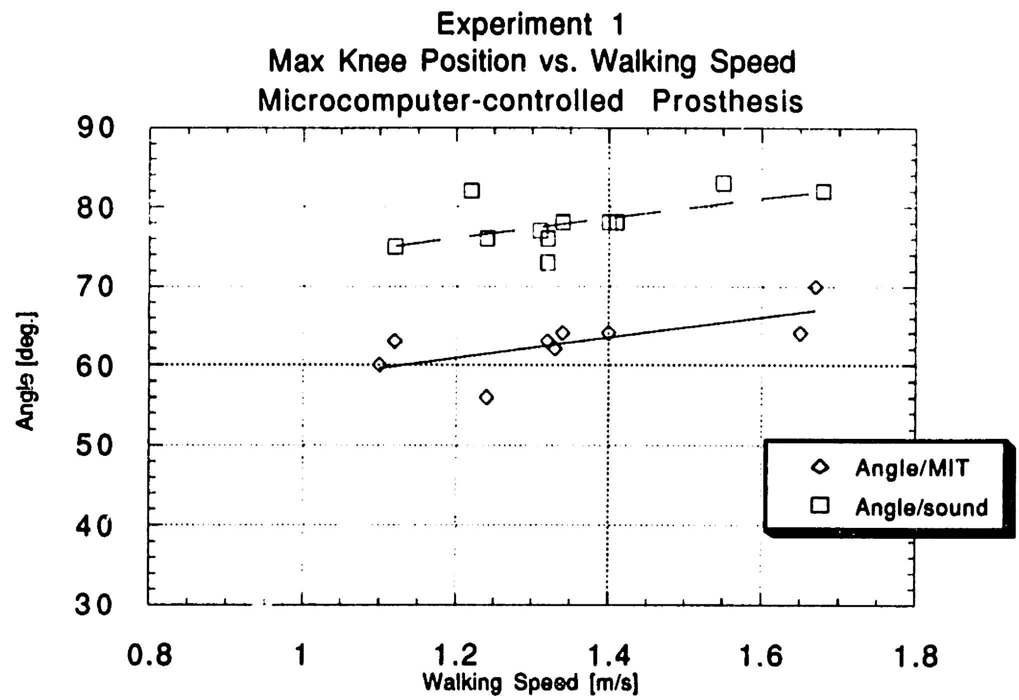

Figure 4-1: Maximum knee-position using the microcomputer-controlled prosthesis

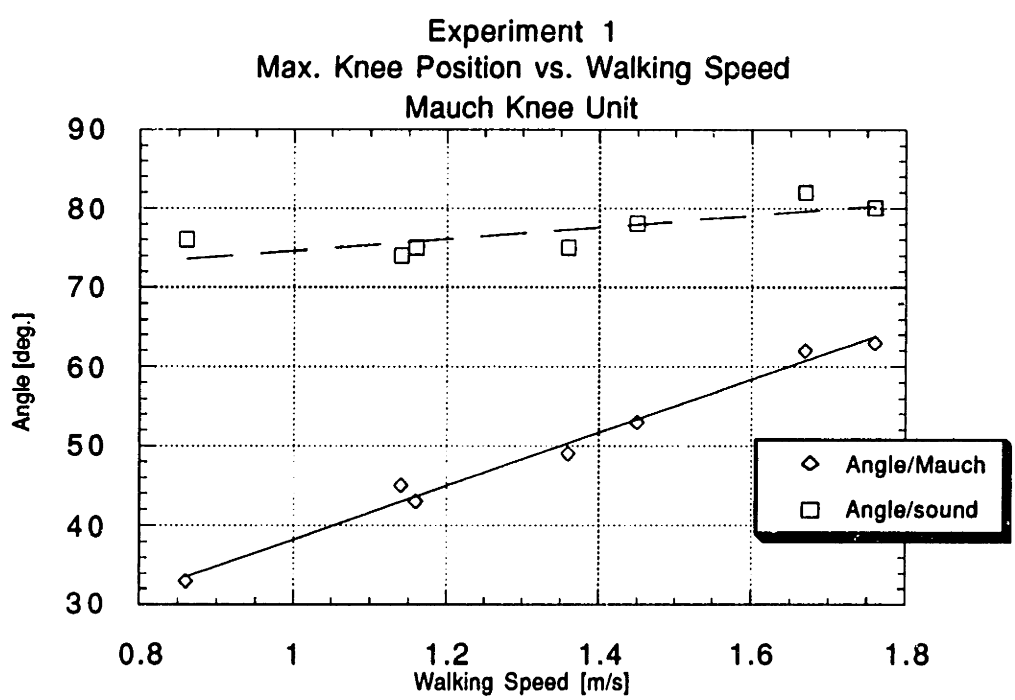

Figure 4-2: Maximum knee-position using the Mauch prosthesis

The subject reported that excessive heel rise at high walking speeds was causing a delay in his gait cycle such that he had to wait slightly for the leg to complete the swing phase. This excessive heel rise was reported in the walking experiments conducted by Goldfarb, and a heel-ramp algorithm was implemented in addition to a turbulent-damping algorithm for swing.

After the protocol was completed, the subject experimented with the turbulent damping and heel-ramp routines. The subject commented that the prosthesis responded well at all cadences with a noticeable improvement at higher walking speeds.

4.2 Experiment 2

The results from the second walking experiment are shown in figure 4-3. It appears from the graph that there was a range of damping constants from 130 (51%) to 140 (55%) that suited a walking speed near 1.2 m/s. As the damping constants were increased to above 145 (57%), the velocity to achieve comfortable gait began to increase more dramatically. The overall curve suggests that the relationship between walking speed and the damping constant is non-linear and must be determined empirically by the amputee.

4.3 Experiment 3

A large amount of data was taken using the TRACK system, which can not be completely reviewed within the scope of this thesis. In general, the kinematic data of the upper extremity were not examined, and in fact, only a small percentage of the lower-extremity kinematic data was considered.

Due to unexplained errors in the data, many measurements of the gait parameters were not presented bilaterally. In particular, prosthetic-side, kinematic data were occasionally being cut off at the edges of the viewing volume with the consequence being that a full gait cycle was not obtained. Figure 4-9 shows four examples of the data represented in body-segment form. When the system recognized inconsistent data, it deleted the corresponding body-segments as the top two figures indicate. The bottom two figures show complete, body-segment representation.

There is a difference between desired walking speed achieved by cadence-matching the subject's steps to a metronome and that achieved by allowing the subject to walk several times across the lab until the desired speed is achieved. For this series of walking trials, the cadence matching method was chosen for its simplicity. A primary concern with the cadence matching method, however, was that the subject's stride length may have been affected by the dynamic alterations of the prosthesis, causing the velocity to change slightly. Average forward displacement of the center of gravity data was examined for each trial, and the velocity of the center-of-gravity is plotted in figure 4-4 as a function of the damping constant. The figure shows that at the established cadence of 80 BPM there does not appear to be a trend in the walking speed of the subject with changing damping-constants.

The flexion angles of the prosthesis and of the sound knee as functions of the damping constants are shown in figures 4-5 and 4-6 respectively. At the comfortable damping setting, the flexion angle was approximately 60 degrees. At higher damping settings, the flexion angle decreased, and at lower damping settings, the flexion angle increased. This result agrees with the comments made by the subject during earlier tests that at low damping levels, excessive amounts of heel rise caused him to wait for the prosthesis to cycle forward for heel strike.

The sound-knee flexion angles were also affected by the changes in the damping level, although not as much as those of the prosthetic side. The angles ranged between 80 and 70 degrees with 76 degrees being the maximum flexion angle of the knee at the comfortable damping setting.

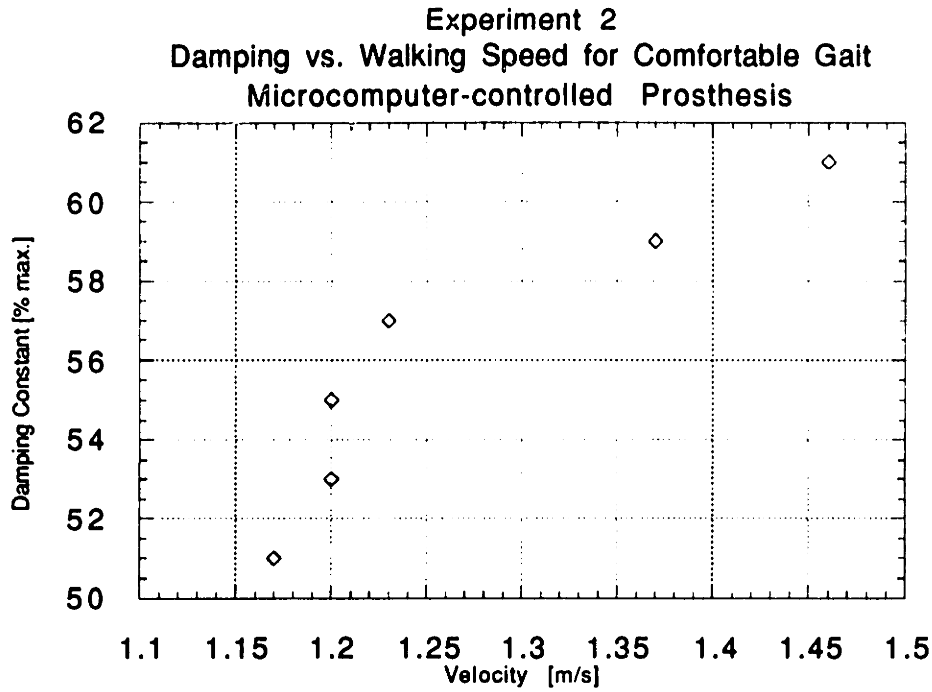

Figure 4-3: User-selected damping vs. walking speed for comfortable gait

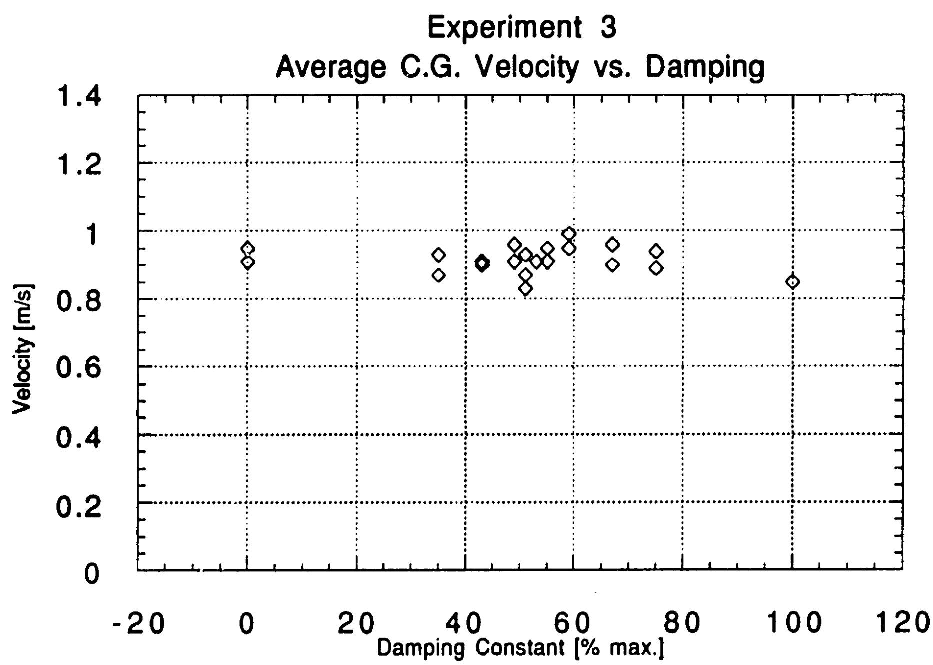

Figure 4-4: Unchanging c.g. velocity with altered damping levels

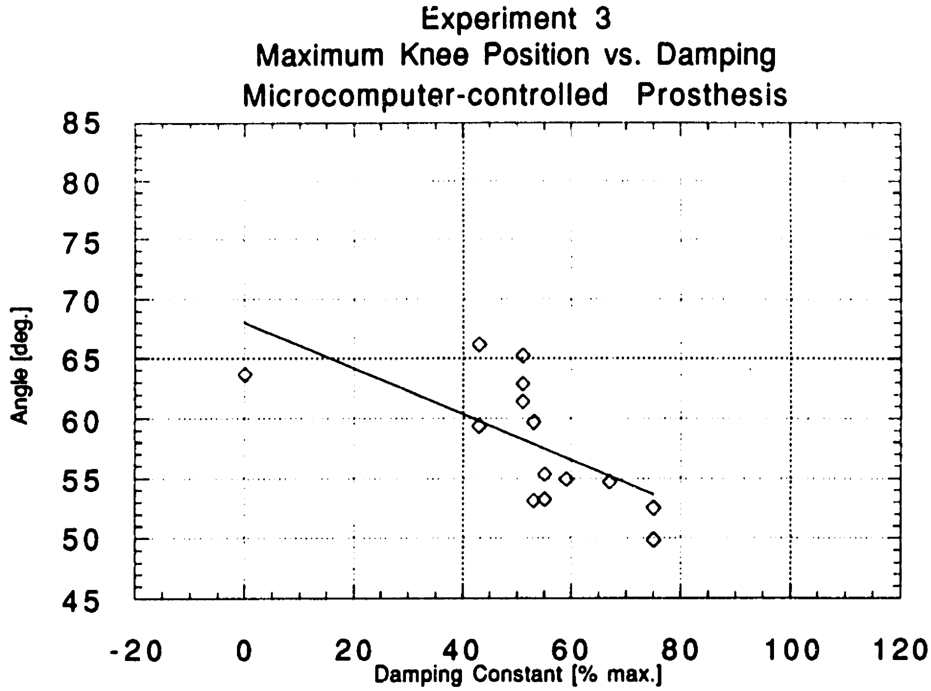

Figure 4-5: Changes in maximum knee angles of microcomputer-controlled prosthesis with altered damping levels

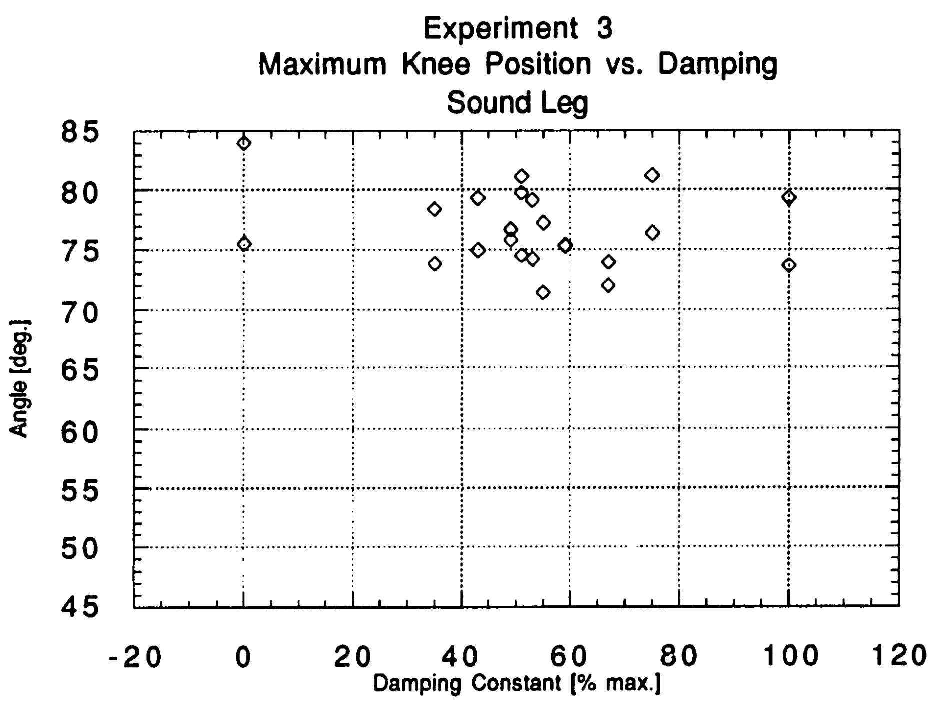

Figure 4-6: Changes in sound-knee angles with altered damping levels

Timing-changes of specific occurrences during the gait cycle indicate compensations made by the subject. It was found that the TRACK system did not record a full gait cycle on the prosthetic side, although it did record a full cycle on the sound side. Because of this, the timing of occurrences was difficult to compare bilaterally.

The period of time from heel-off to heel-strike was measured on the sound side. Figure 4-7 shows this time period as a function of the damping constant. The interesting aspect of this plot is the occurrence of a maximum swing time at the comfortable damping setting. At settings above and below this, the sound-leg swing-time decreased by as much as 13% of the overall gait-cycle, indicating a compensation of the sound side. The implication of a shorter sound-leg swing-time, while maintaining approximately the same knee flexion angles, is that the torques and accelerations about the knee increase. This leads to increased muscle activity and, therefore, increased metabolic energy.

The stance time of the prosthetic leg was determined and is shown in figure 4-8. The stance time of the prosthesis does not show appreciable change as a function of the damping constant.

4.4 Results Summary

An attempt was made to identify gait parameters that provided indications of proper damping levels at the knee. It is desirable to identify these parameters for the prosthetic side rather than the sound side to enable the controller to measure these parameters and employ proper control algorithms. The results from the third experiment show changes in the heel-off-to-heel-strike time for the sound side; however, the same type of measurable changes did not appear on the prosthetic side. Further investigation could lead to the discovery of a characteristic gait-parameter for the prosthetic side.

User-selected behavior may be a possible approach to the design of sophisticated controllers that interact with the user to determine optimal behavior. The results from the

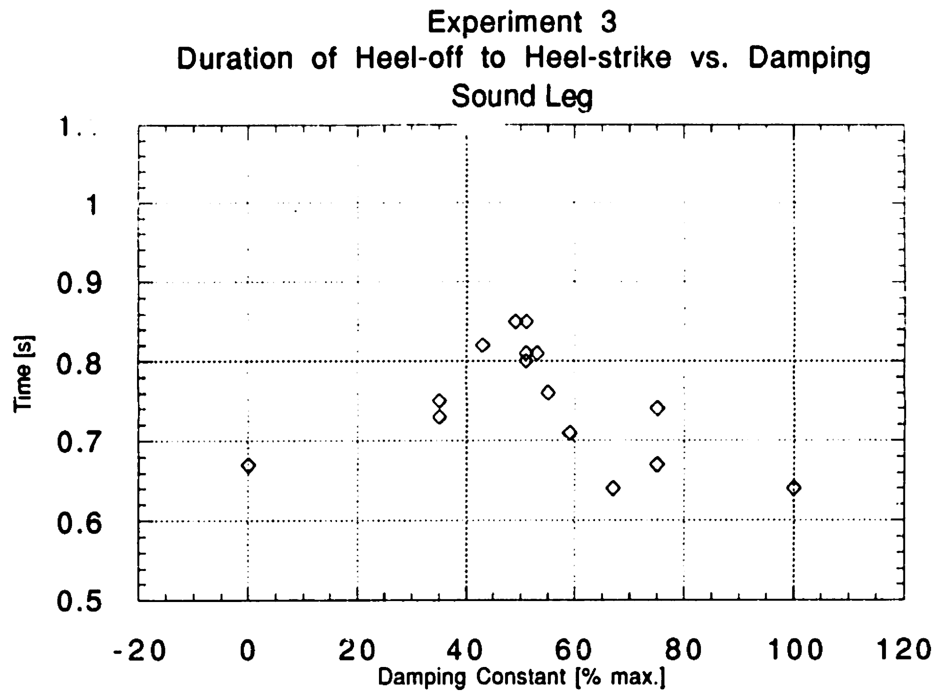

Figure 4-7: Changes in sound-leg-swing time with altered damping levels

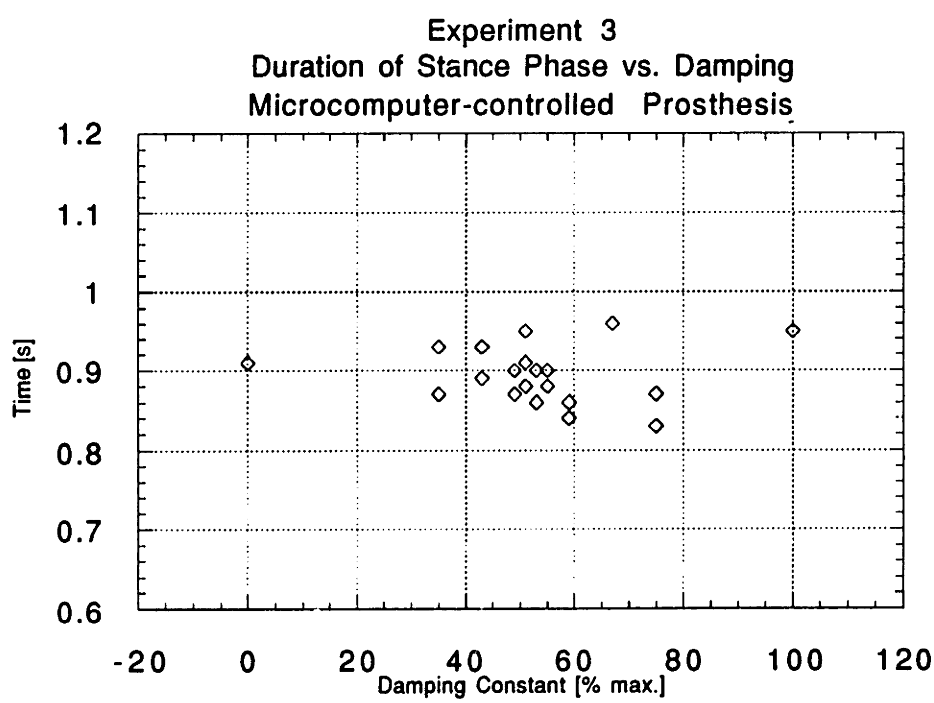

Figure 4-8: Unchanging prosthetic-stance time with altered damping levels

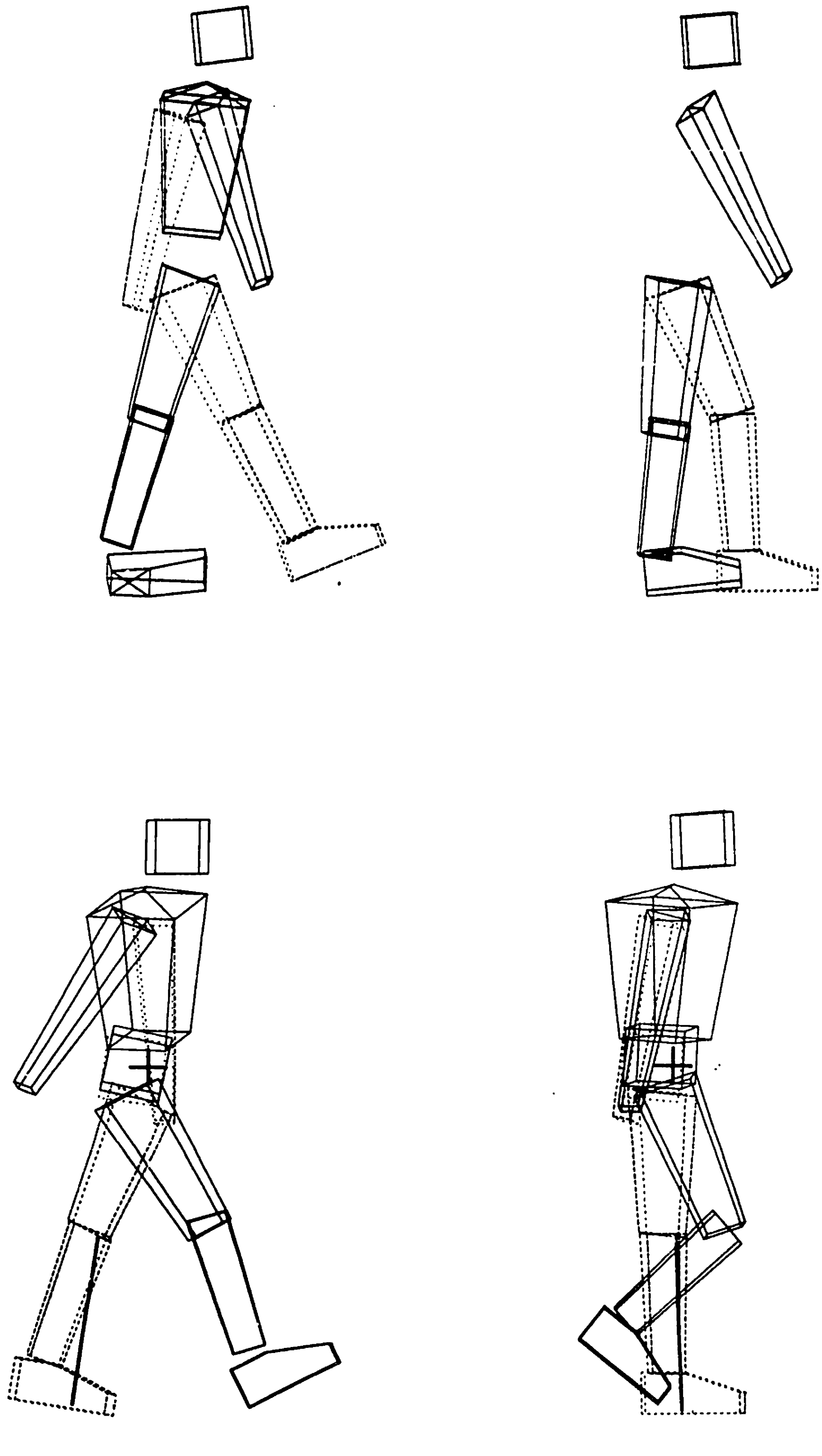

Figure 4-9: Computer generated images of the subject walking. Erroneous data occasionally resulted in missing body segments as seen by the top two figures.

second experiment suggest that this method may be used to determine the non-linear dependence of the damping constant on walking speed. The ability of the amputee to accurately choose proper behavior can be seen from the results of the third experiment. The subject accurately chose the damping constant that led to a maximum heel-off-to-heel-strike time on the sound side, and hence, minimum torques about the knee.

An issue that merits discussion is the possibility that a familiarity factor had a large influence on the results. The subject normally wears his Mauch prosthesis several hours each day and has become accustomed to its behavior. It is likely that the subject has developed a pattern of gait ideally suited for his prosthesis and will continue to exhibit this pattern while evaluating the microcomputer-controlled prosthesis.

An example of this occurred during one of the initial walking trials conducted by Goldfarb. The subject walking on the computer-controlled prosthesis was experiencing a locking of the knee just prior to toe-off. This problem was extremely uncomfortable for the subject, and the controlling parameters were adjusted in an attempt to eliminate the problem. It was later found, during review of the data, that the subject was kicking back with approximately 15 pounds on the prosthesis, prior to toe-off, in the same manner that he kicked back on his own prosthesis in order to release the knee mechanism from its locked position. The controller, sensing an increase in weight, quickly switched algorithms from swing to stance and locked the knee.

Therefore, if an amputee is comfortable with the performance of his normally-worn prosthesis or has spent many years learning to walk on his prosthesis, it is reasonable to assume that he may tend to praise the performance of the microcomputer-controlled prosthesis as its behavior converges upon that of his normally-worn prosthesis.

Chapter 5

Recommendations

5.1 Controller

Ease and flexibility of programming control algorithms is important during the development of sophisticated controllers because changes need to be made as the factors affecting amputee gait are identified through continued experimentation. The current control algorithms are programmed in assembly code for the NEC 78C10 microprocessor. With the need for more sophisticated algorithms, a higher-level programming language such as C should be used rather than assembly language, which tends to be a difficult language for calculating complex equations. A C compiler for the microprocessor would facilitate the implementation and modification of more sophisticated control schemes

During a few of the trials, the subject reported that the Mauch prosthesis felt "smoother" than the microcomputer-controlled prosthesis. One explanation may be that the normally-continuous profile of the brake's output during swing is subjected to occasionally-aberrant discontinuity. Filtering the output to the brake may insure a more continuous profile, resulting in a smoother response for the user.

Torques in the normal knee have been determined to be functions of gait speed, knee position, and knee velocity. It would not be unreasonable to model the knee torque of the prosthesis after the passive torques of the normal leg. Darling studied phase-plane trajectories of knee position and velocity as a means of optimal control [1]. Another approach might be to allow the torques to be proportional to the swing velocity raised to any power rather than being proportional to just velocity, as in viscous damping, or proportional to the velocity squared, as in turbulent damping.

Presently, the turbulent-damping routine is inadequately scaled and cannot provide sufficient knee torques at high swing-speeds. Therefore, if the current controller is going to be used with the turbulent-damping routine, the scaling factor should be adjusted to increase the sensitivity of torque to swing speed.

5.2 Hardware

The shank length of the prosthesis is not very adjustable. It was initially fixed to the proper length for the subject, but if the leg is to be tested by other subjects, more adjustability should be built into the shank. Before this can be done, the load cell, which occupies most of the shank's length, must be reduced in size, since the existing length of the load cell prevents the shank from being shortened by more than about one inch. Considering that the subject who has been testing the leg stands over six feet in height, it is reasonable to expect the length of the shank to shorten by at least two or three inches in order to accommodate shorter subjects.

The currently-exposed electronics should be more protected. An easily-detachable, protective housing may be a feasible addition. Some of the electronics are delicate and have been accidentally damaged on several occasions, mostly by people handling the prosthesis without realizing its delicate nature. The housing would also protect the electronics if an amputee were to stumble and fall.

5.3 Future Experimentation

The opinion of the user is an important aspect of prosthesis evaluation; however, as was discussed in the previous chapter, the behavior of the subject's normally-worn Mauch prosthesis may influence his opinion of other prostheses. This may support the need for the subject to perform an extended evaluation that would allow the subject to wear the microcomputer-controlled prosthesis in place of his normal prosthesis. After becoming accustomed to the microcomputer-controlled prosthesis, experiments could be performed in the lab.

Another approach to prosthesis evaluation by the amputee could be to create a user-optimized controller that would allow the subject to easily alter the behavior of the prosthesis as he wears it for an extended period of time. When the subject has adapted to the comfortably-adjusted leg, he would return to the lab where walking trials could be conducted.

Finally, an energy-consumption study may provide positive results by confirming that sophisticated control algorithms considerably reduce metabolic energy costs to amputees by intelligently optimizing their gait.

Appendices

A.1 Assembly Code for Controller

The controller code, written by Goldfarb [4], was modified to include the turbulent-damping and heel-ramp routines.

The mode 4 algorithm during swing phase is a viscous algorithm that applies the knee torque linearly with velocity:

$$T_{knee} = \beta * \omega * \left[\frac{t_o}{\Delta t + t_o}\right]^2, \tag{A.1}$$

as opposed to the turbulent algorithm that applies the knee torque as the velocity squared:

$$T_{knee} = \beta * \omega^2 * \left[\frac{t_o}{\Delta t + t_o}\right]^2. \tag{A.2}$$

The heel-ramp routine was implemented to control the amount of heel rise at full flexion. The algorithm provides a ramp function that increases the knee torque linearly beyond a starting angle during flexion. The starting angle and the slope of the heel ramp can be changed through the parameter file on the host computer. The algorithm for the heel-ramp routine is given by:

$$T_{knee} = \beta * \omega^2 * \left[\frac{t_o}{\Delta t + t_o}\right]^2 * [1 + s(\theta_r - \theta)], \tag{A.3}$$

where s is the slope of the ramp, and θ_r is the starting angle.

A.2 Interface Program

The parameter file that is downloaded from the host computer to the on-board microprocessor was modified to include the additional parameters necessary for the turbulent and heel-ramp algorithms. Table A.1 shows the updated set of controlling parameters.

	Parameter	Description
Controller Parameters	Damping Constant	The viscous damping constant for swing phase damping
	Swing-Ramp Angle	The angle at which the viscous damping begins to increase
	Swing-Ramp Slope	The rate at which the viscous damping constant increases in the latter part of swing
	Knee Break Torque	The percentage of maximum knee torque imposed during initiation of swing
	Knee Break Threshold	The axial load in the shank at which the amputee begins knee flexion
	Foot Contact Value	Defines the amount of axial force in the shank that determines when the foot is in contact with the ground
	Locking Torque	Defines the amount of huperextensive torque during stance above which the magnetic particle brake is turned off
	Gait Speed Factor	Affects the amount by which swing phase damping increases with increased gait speed
	Heel-Ramp Slope	The rate at which damping increases near maximum knee flexion
	Heel-Ramp Angle	The angle at which the damping for the heel ramp begins
	Velocity Threshold	The velocity below which the heel ramp turns off
	Damping Type	Determines whether the damping type during swing is viscous or turbulent

Table A.1: Table of controller parameters

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