

A SPEED ADAPTIVE CONTROL ALGORITHM  
FOR THE SELF-CONTAINED A/K PROSTHESES

by

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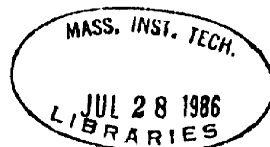
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Archives

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Submitted to the Department of Mechanical Engineering  
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ABSTRACT

Improving the speed responsiveness of the above-knee prosthesis while keeping the control scheme self-contained is the focus of this thesis. A method of predicting the intended walking speed was established. Two types of control scheme, Trajectory Following and Torque Control were developed with the latter one being preferred by amputee subjects during person-interactive experiments. Comparison of the schemes suggested that the smoothness of shank motion weighted heavier than the exact duplication of the normal knee kinematics. The inertial properties of the leg plays an important role in the pattern of gait kinematics.

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DEDICATION

This thesis is dedicated to my father,

Shaoqin Qi

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## CHAPTER 1 INTRODUCTION

### PROBLEM DESCRIPTION

The loss of a lower limb is an extremely harsh experience both physically and psychologically. An above knee (A/K) amputee faces the competitive world with severely reduced mobility and other physiological deficiencies. A prosthetic device must be employed to provide restoration of normal functions and to minimize the amputee's social, physical, and psychological disadvantages, therefore to enable him or her to continue as a contributing member of the society.

It is a uniquely difficult task, however, for the A/K prosthesis to reproduce the elegant lower limb provided by nature. While millions of years of evolution have made the human locomotion system an engineering marvel, thousands of man-years of effort on replacing its missing parts have had very limited success. The current, commercially available A/K prosthesis can approximate very few functions of a normal leg, resulting in a type of gait that lacks the comfort, ease, cosmesis, efficiency and adaptivity of the normal gait. There is a great room for improvement in almost every aspect of A/K prosthesis design.

THESIS OBJECTIVE

The knee joint control of A/K prosthesis is a key issue in attempts to produce a "normal" gait. This thesis is part of an integral effort to design an adaptive, individualizable knee unit using the most recent advances in technology. At present, all of the A/K prosthesis that the amputees wear have only a functionally simple damper at the knee joint. Its oversimplified joint dynamics can not generate the complicated torque profile the normal knee joint does, nor can the torque profile be individualized. For different walking speeds, it can not correspondingly change its dynamics, thus requiring more effort from the amputee to achieve the desired kinematics. All of these suggest that a more sophisticated knee mechanism and a more versatile control scheme should be developed.

A magnetic-particle-brake(MPB)-controlled knee mechanism has been developed by the MIT Knee Group. It can be controlled electronically and thus enables a computer to be incorporated. Several types of design and tests have shown that it has the potential to replace the currently used simple damper. A self-contained microcomputer-controlled MPB prosthesis is being developed.

This research attempts to develop a speed-adaptive control scheme for the new MPB-microcomputer prosthesis. It

requires the controller to recognize the amputee's intended walking speed as well as desired kinematics. The control algorithm is to be designed to achieve these desired features by adjusting the knee dynamics. First, amputee-interactive experiments were conducted to search for a speed indicator. Then walking trials were carried out to evaluate the performance of the control algorithm.

### THESIS ORGANIZATION

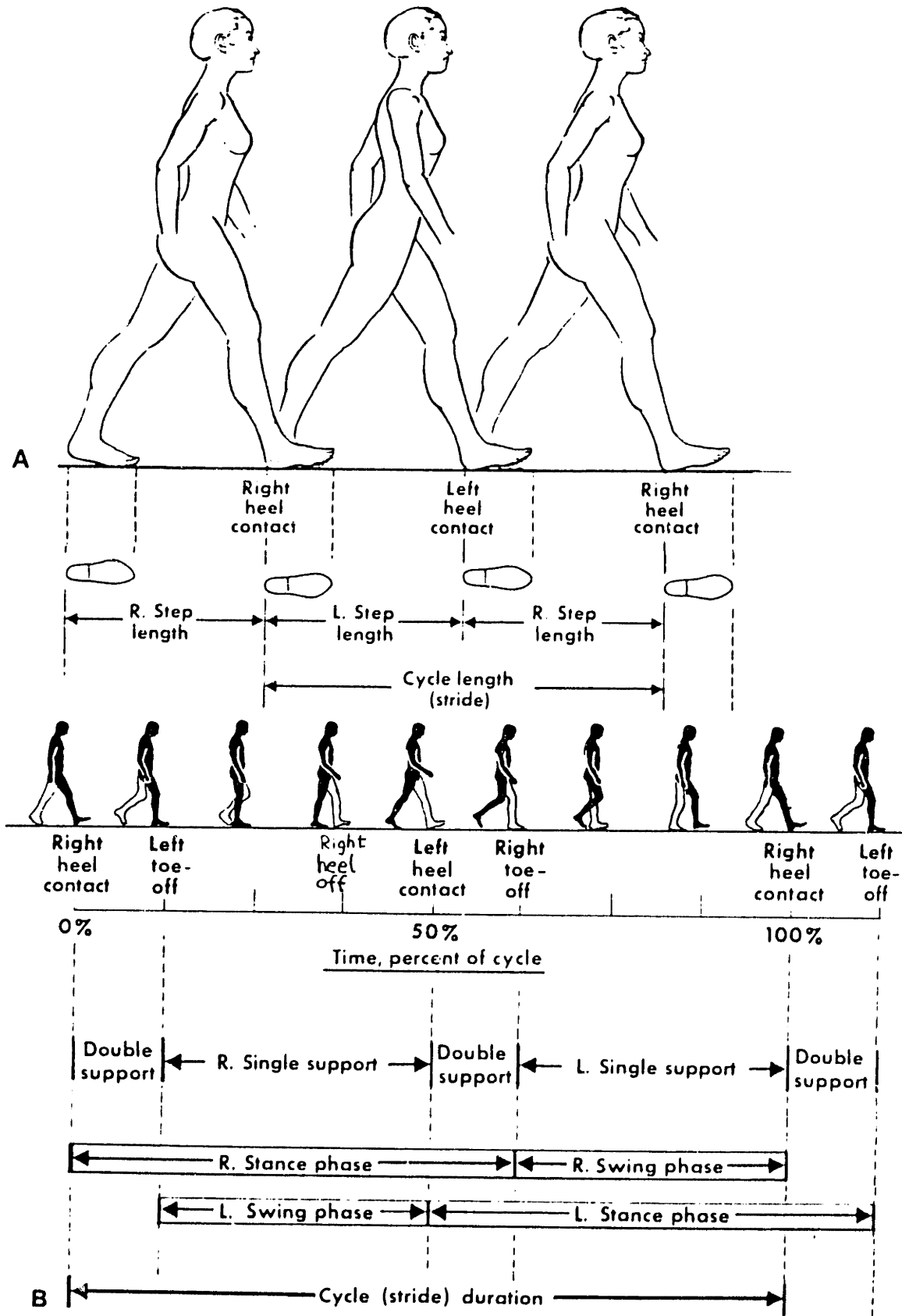
Chapter 2 provides the background information relating to the normal and A/K amputee locomotion, A/K prostheses and research in this area. Chapter 3 contains the analysis and development of the walking speed detection algorithm and adaptive control schemes for knee joint. The amputee-prosthesis simulator system used for experiments is described in chapter 4. Chapter 5 then presents the experiment's results and makes conclusion and recommendations.

## CHAPTER 2 BACKGROUND

### NORMAL GAIT

Human walking is simply the process of locomotion in which the erect, forward-moving body is always alternatively supported by the two legs. In most cases, normal gait is symmetric, smooth and harmonious. Despite this apparent simplicity, the act of walking is actually a very complex procedure involving almost all the body segments in three directions and with three degrees of freedom. It is the large number of degrees of freedom that makes the quantitative description of gait very difficult. Extensive measurement of gait in recent decades has led to a much better understanding of the mechanics of walking. A brief description of major characteristics of gait is presented here.

Walking is commonly identified as a repetition by each leg of a basic pattern known as the walking cycle. Figure 1 illustrates this cycle and its sequence of events. It is based on the motion of one leg and starts arbitrarily at heel contact (HC). The cycle then proceeds to foot-flat (FF), heel off (HO), toe-roll (TR) and toe-off (TO), concluding upon reaching heel-contact again. Further subdivisions can be made by including the events of the opposite leg. Two distinct phases are present in the cycle:



**Time Dimensions of Walking Cycle**

**Figure II-1** Distance and time dimensions of walking cycle. A, distance (length). B, time. [ 8 ]

Stance phase when the foot is in contact with the ground and swing phase when it is in the air. The leg supports the body and propels it forward during stance phase and then is lifted and swung through to prepare for the next stance. Overlap occurs between the stance phase of one leg and stance phase of the other. This is called the double support time, when the body weight is being shifted from one leg to the other.

A walking cycle has two basic dimensions: the time period from beginning to the end (stride time), and the distance the body travels during the cycle (stride length). They are usually expressed by the step frequency, reciprocal of half the stride time, and step length, illustrated in Figure 1. The use of step instead of stride is to reflect the symmetry of gait. The left and right step length will have the same value and will be one half of the stride length if the gait is symmetric. Walking speed is the product of step length and step frequency. Although changes in walking speed can be made by changing either the step length or the step frequency and keeping the other unchanged, normal persons change both. Usually, both step frequency and step length are increased when speed is increased. But each person has his/her own combination of the step rate and step length for a given speed. If he/she is forced to take either unusually shorter or longer strides at any speed, the energy required per meter is increased.



Therefore, the step length-step frequency relationship for each person is consistent under the same walking condition. In fact, they are the two fundamental parameters that largely determines the patterns of the gait.

Figure 2, 3, 4 are the plots of the moment, angle and power in the saggital plane for knee, hip and ankle joint during one cycle of level walking. The knee angle reaches full extention at the end of swing, just prior to HC. Then immediately following heel strike, the knee begins to flex, allowing a smaller vertical displacement of the body than if it remains fully-extended. The knee flexion during stance is also believed to absorb the shock occurred at heel strike. The knee is then re-extended to enable toe clearance for the opposite leg now nearing the end of its swing phase. Near the end of knee extension period, power is generated at the ankle to propel the body forward. The knee then begins to flex again as the hip flexes to propel the whole leg forward. Ankle is dorsiflexed to insure toe clearance. During most of the swing phase, the knee acted like a damped pendulum, preventing excessive heel rise at the flexion period and avoiding impact at the end of extension period. Only exception is at the start of knee extension, when power needs to be generated at the knee joint to initiate the knee extension.

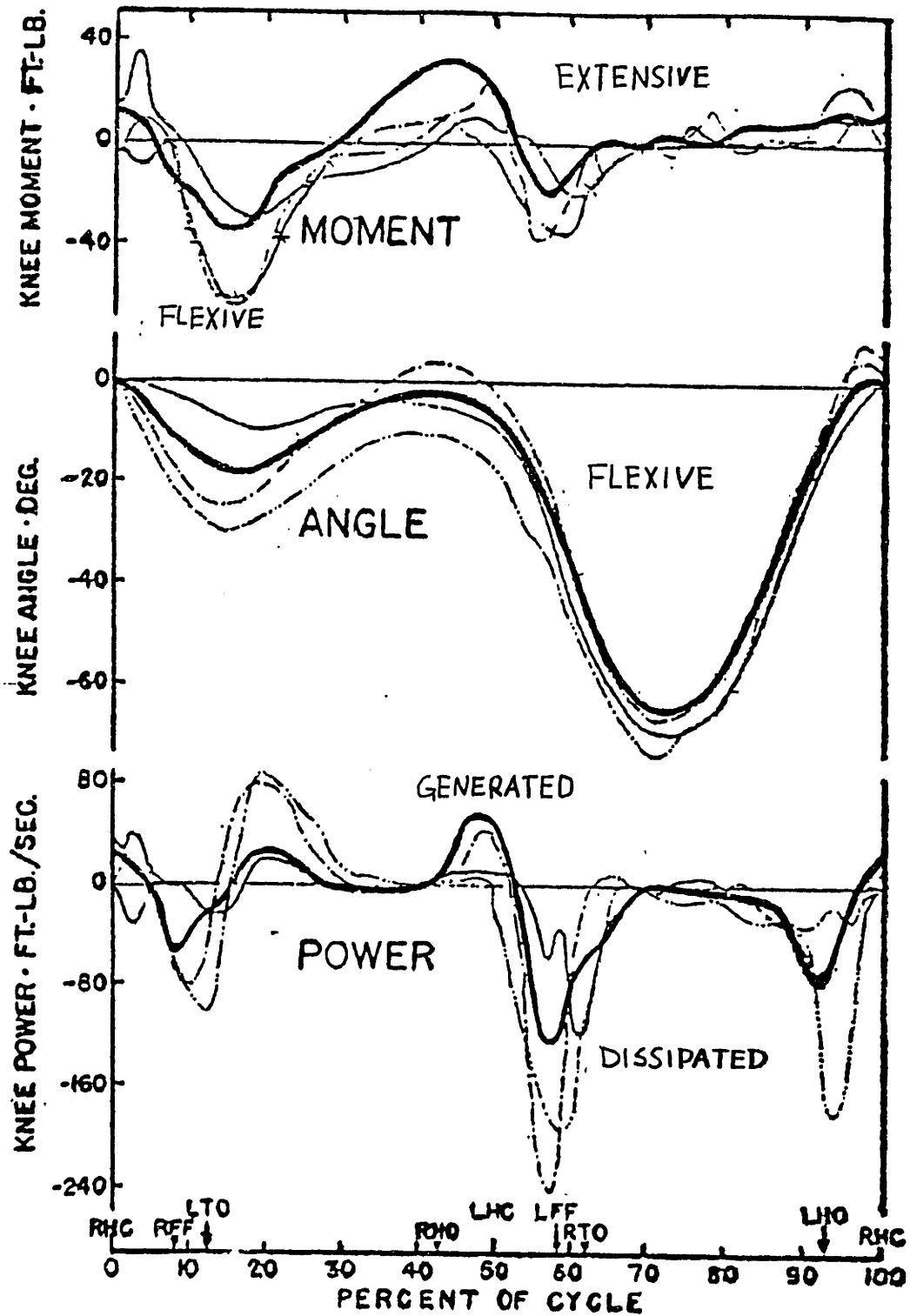


FIGURE II-2  
KNEE MOMENT, ANGLE, AND POWER OF FOUR NORMAL SUBJECTS  
(From reference [2] )

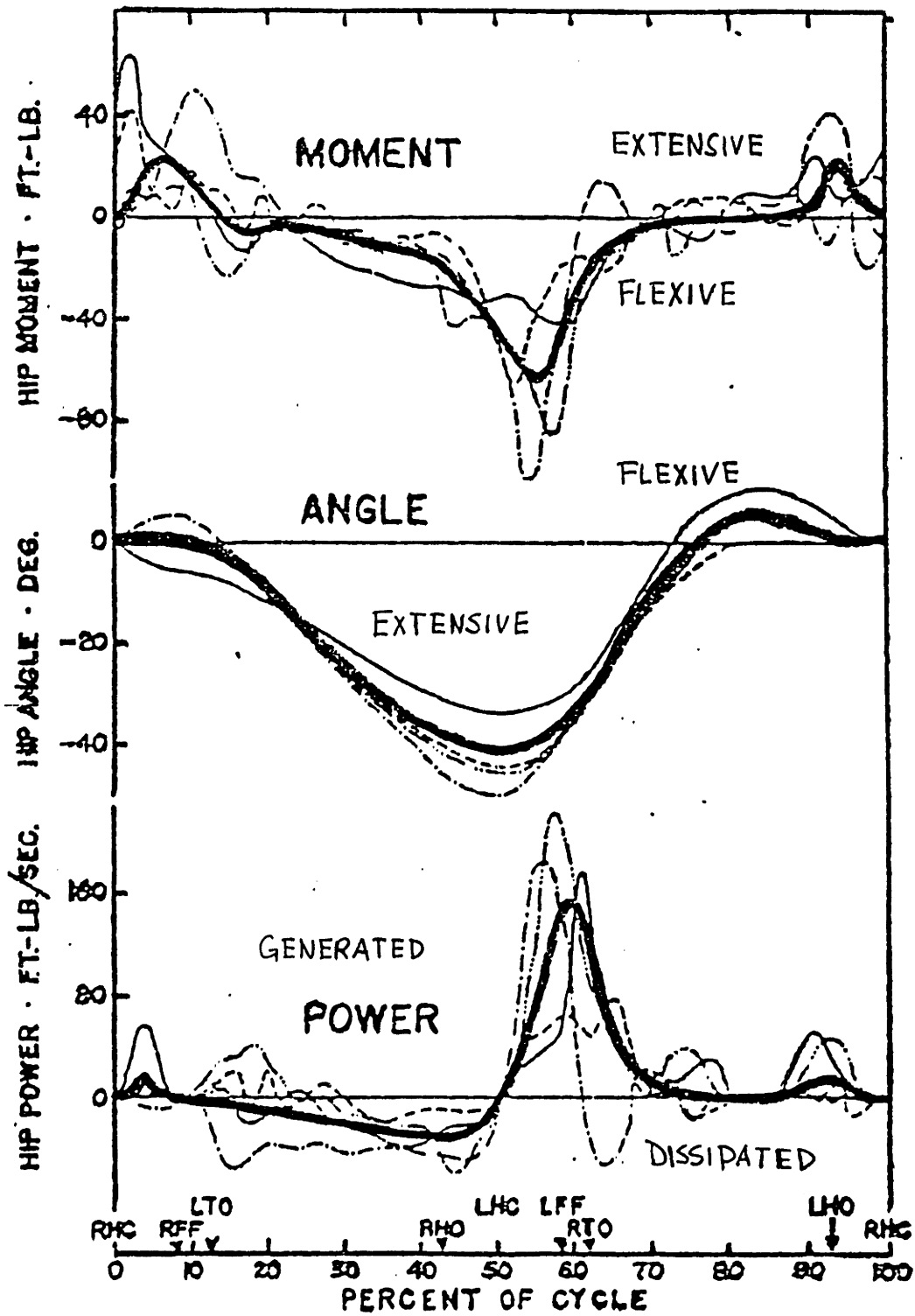


FIGURE II-3

HIP MOMENT, ANGLE AND POWER OF FOUR NORMAL SUBJECTS  
(From Reference [2])

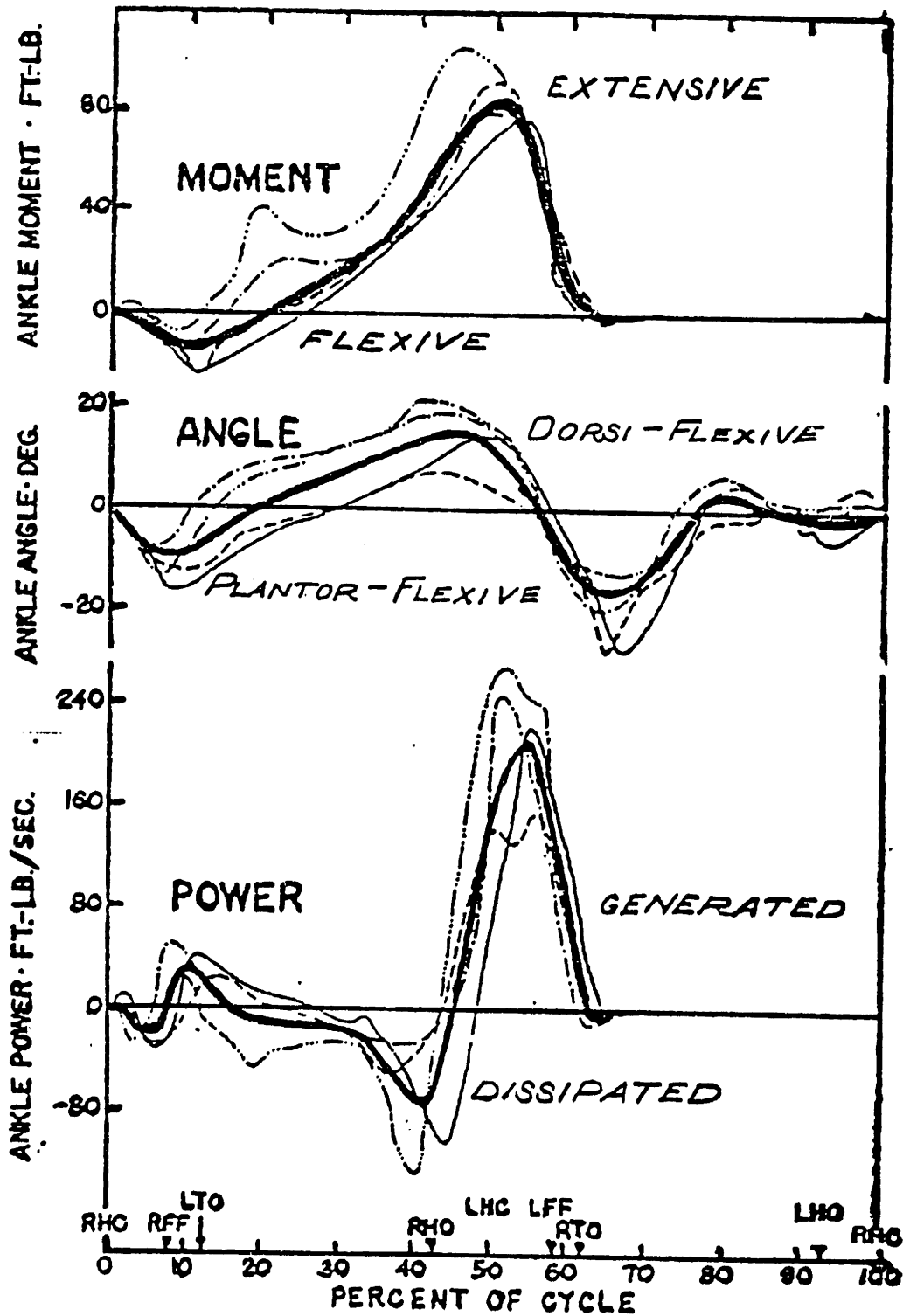


FIGURE II-4  
ANKLE MOMENT, ANGLE AND POWER OF FOUR NORMAL SUBJECTS  
(From reference [2])

Figure 2, 3, 4 also showed that there is a tremendous individual variation in gait. Figure 5, 6, 7 showed the speed variations in gait kinematics. Note that as the walking speed changes, the gait pattern remains about the same but variation in magnitude is apparent. Major differences are that the hip flexion angles increases to increase the step length and knee flexes more during stance to absorb more shock as the walking speed increases.

#### THE CONVENTIONAL A/K PROSTHESIS

Countless attempts have been made over the last few thousand years to develop a prosthesis capable of restoring the normal functions of the natural leg. As a result, the number of commercially available A/K prosthesis is large. But despite this wide selection, all of these units are functionally simple compared to the normal leg. Only one degree of freedom is provided at the knee joint. No power can be generated. Little voluntary control of the prosthesis is available.

Figure 8 shows the typical prosthesis worn by the majority of the amputee population. It consists of a socket, a knee unit, a shank, and a SACH foot (Solid Ankle Cushion Heel ).

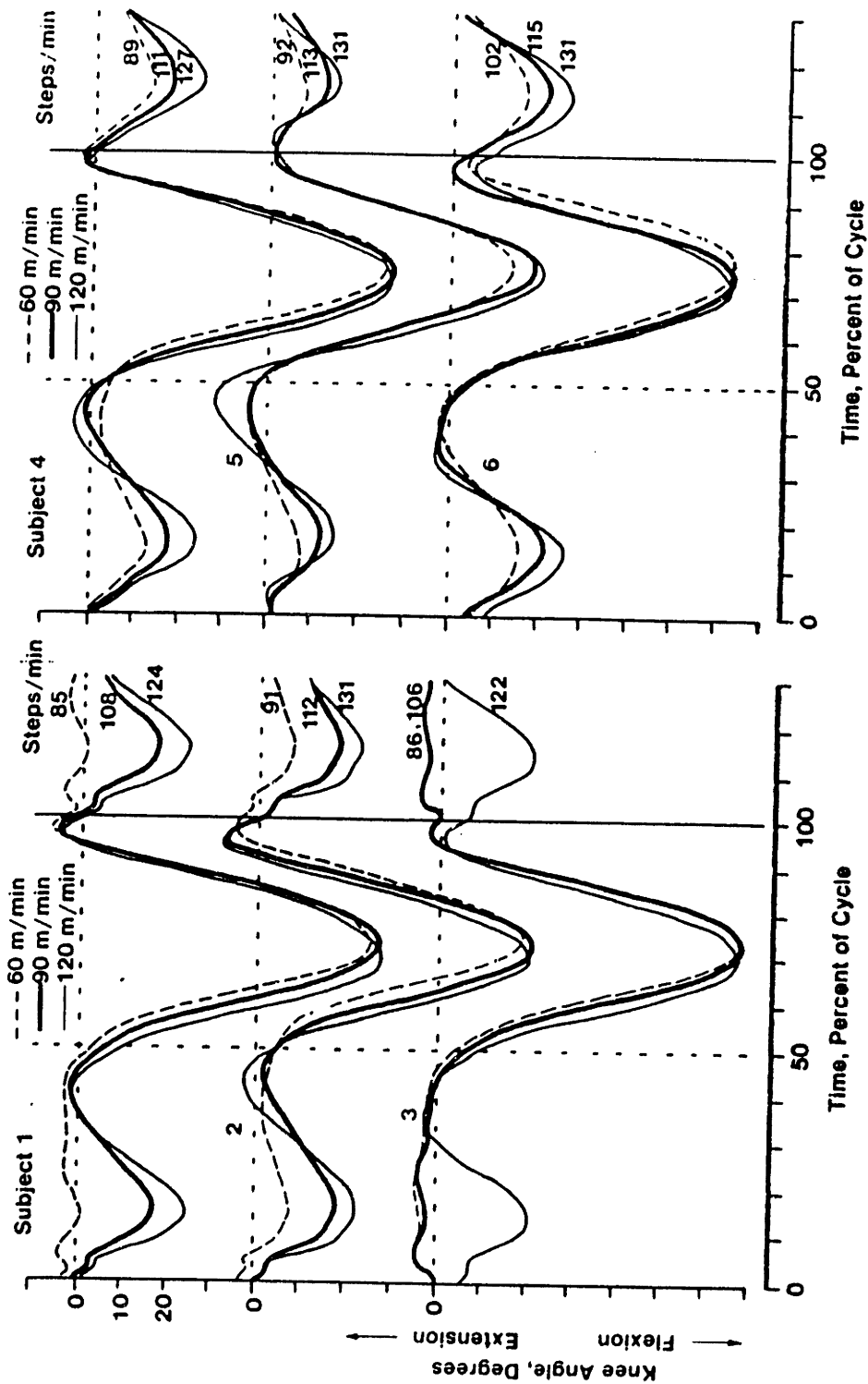


FIGURE II-5

Effect of walking Speed on Knee Angle. Angle at Stanging Position is Zero.[8]

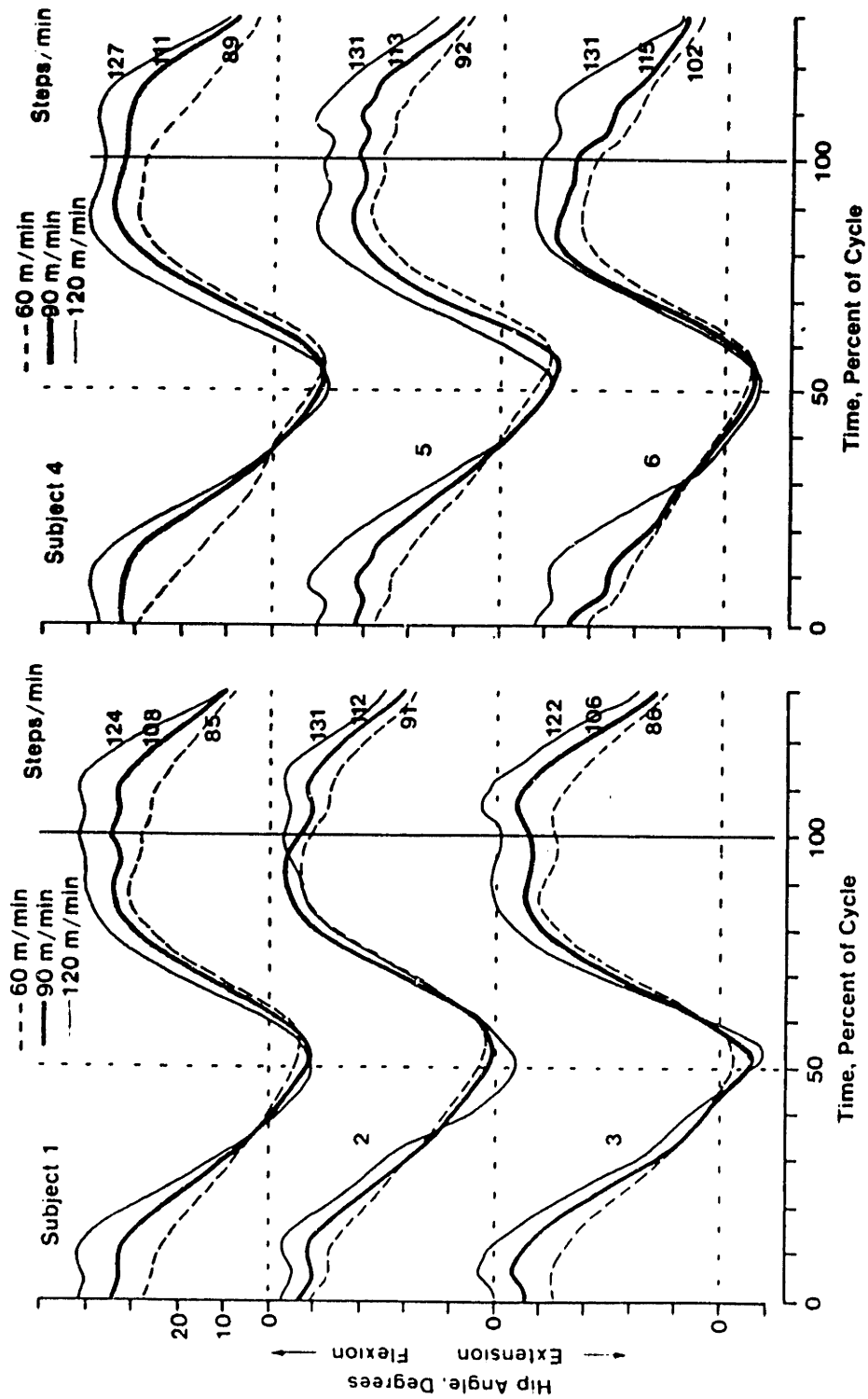


FIGURE II-6

Effect of walking speed on hip angle. Angle at standing position is zero. [8]

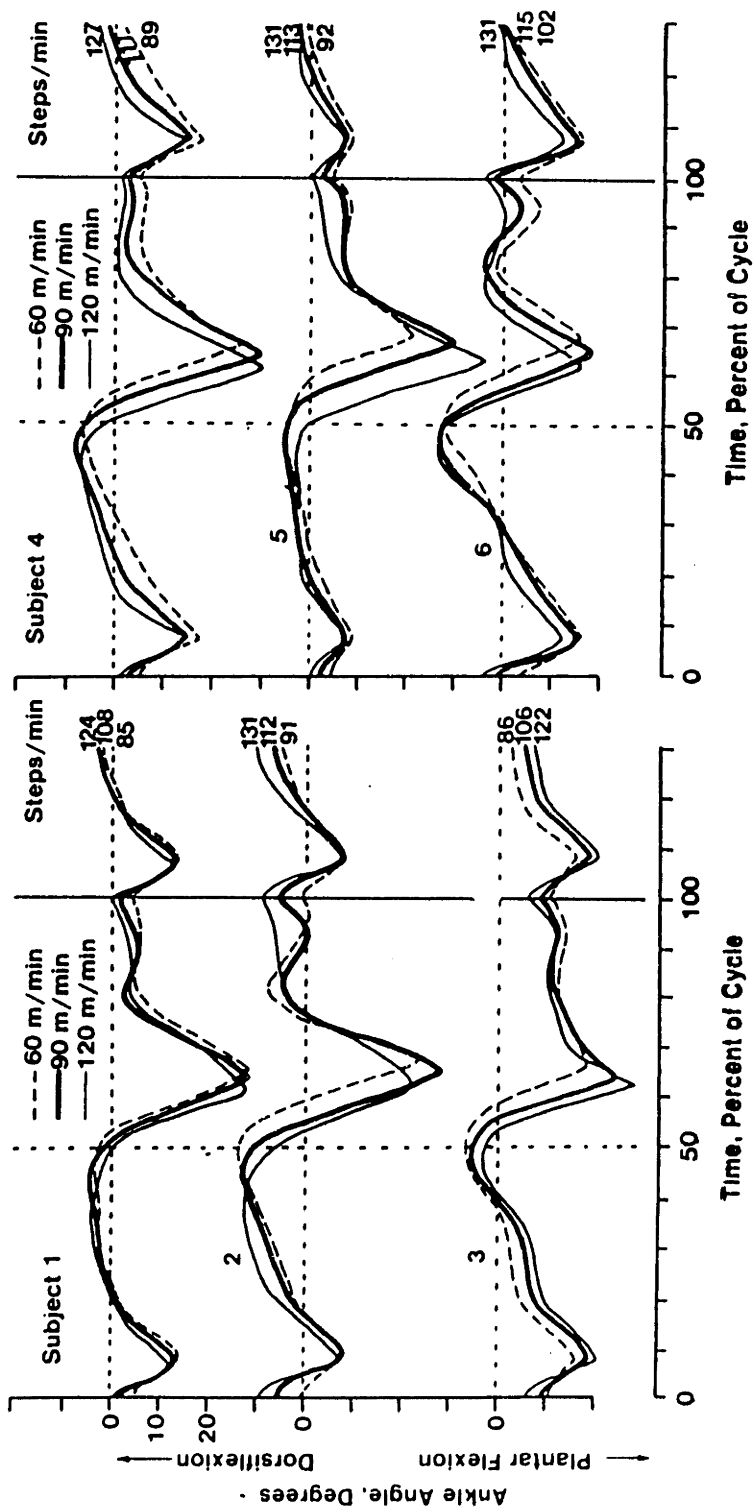


FIGURE II-7  
Effect of walking speed on ankle angle. [8]



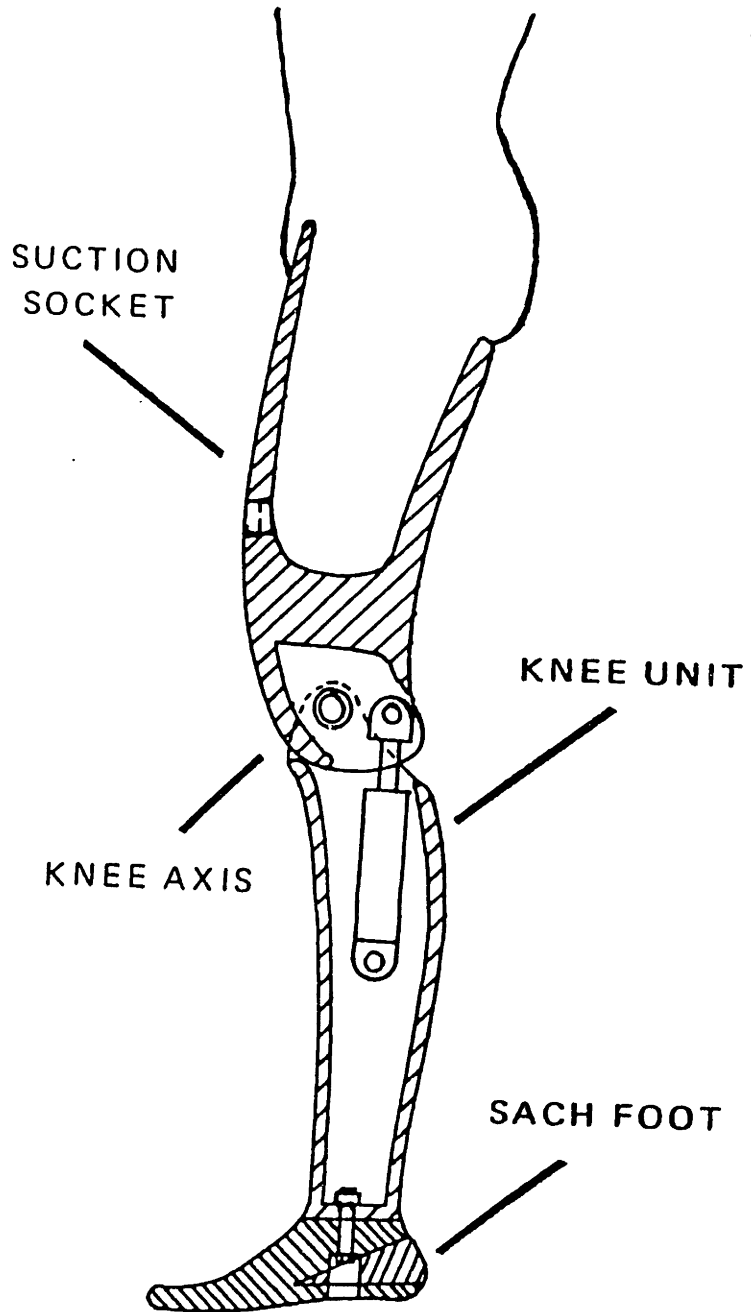
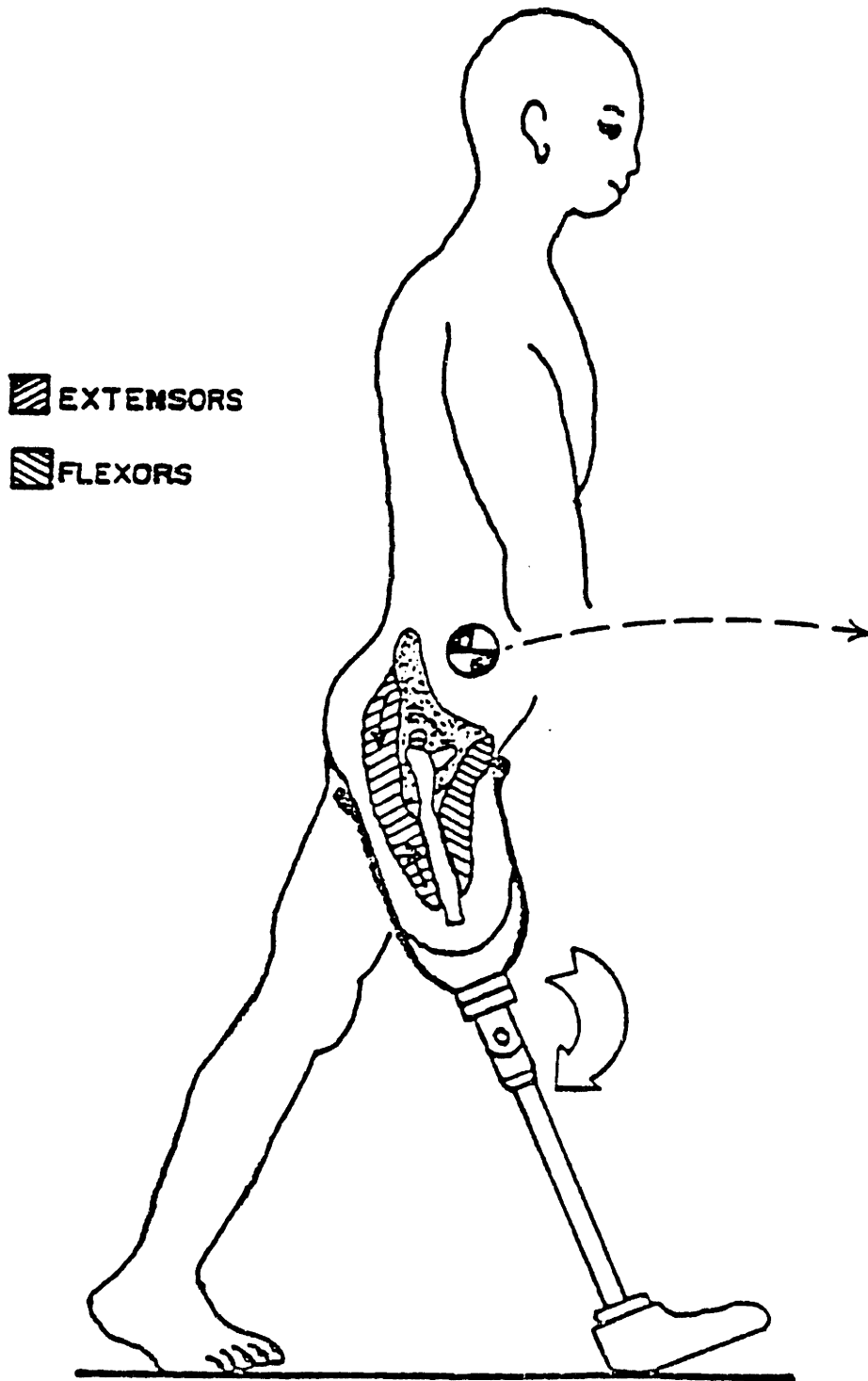


Figure II-6. A Conventional Prosthesis.

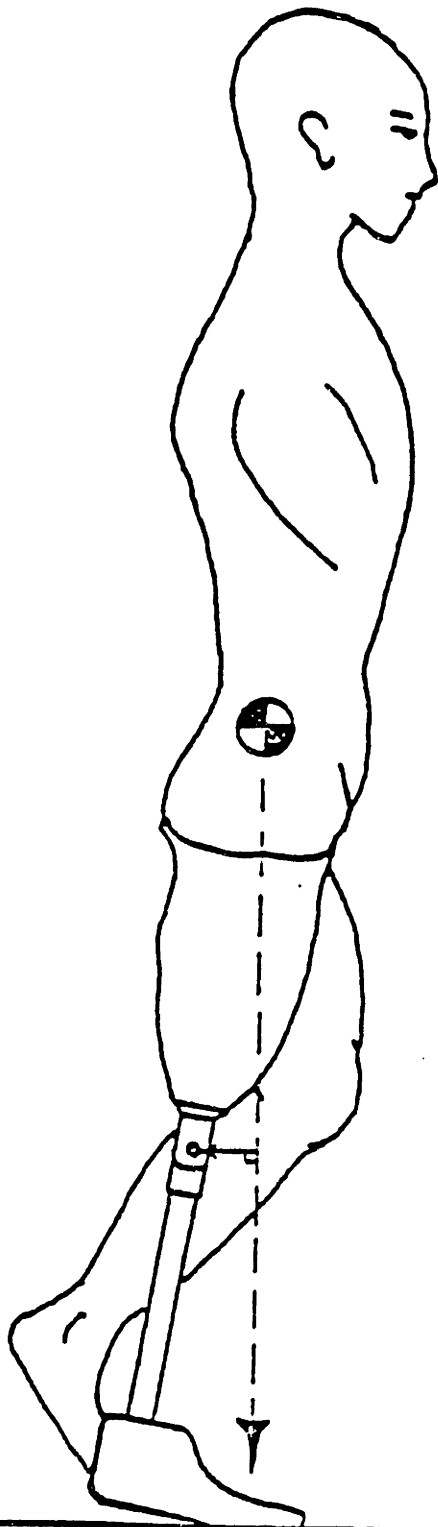
The socket is the interface between the person and the prosthesis system. A suction socket is usually used. The socket is held to the stump by the vacuum formed between them. Therefore, a close fit is very critical. In fact, socket fitting has great influence on the performance of the prosthesis.

The knee unit is the part that distinguishes the prosthesis. It consists of the knee axis and the damping device. Sometimes an extension aid or kick strap is available. The single axis unit is most often used. To achieve the stability during the stance phase, it requires the individual to create a hyperextensive torque against its mechanical stop at the end of swing phase (Figure 9) and placement of his/her body center of gravity in front of the knee axis (Figure 10) during the stance phase. The "amount" of stability can be adjusted by shifting the anterior/posterior position of the knee axis relative to the load line. Other types of mechanism are also used such as the four-bar linkage capable of shifting the center of rotation. The damping device is usually piston cylinder type, either hydraulic or pneumatic. It has a simple characteristics such as viscous damping and only manual adjustment of the damping coefficient is available.



USE OF HIP EXTENSORS TO CREATE  
HYPEREXTENSIVE KNEE TORQUE

FIGURE II-9



BODY WEIGHT BEING USED TO  
CREATE HYPEREXTENSIVE KNEE  
TORQUE

FIGURE II-10

SACH foot (Figure 11) is also an important part of the prosthesis. It is principally designed to provide motion in the anteroposterior plane in simulation of the plantar flexion and dorsiflexion of the ankle and extension of the toe. It can absorb some shock during heel strike and release of this energy gives a little propulsion to the leg. The stiffness of the SACH also affects the knee break at the end of stance.

In summary, the conventional prosthesis provides very limited functions to the amputees. It requires extra effort by the wearer to achieve stability. The device responds to the input motion of the stump with the inherent characteristics rather than to the will of the amputee.

#### RESEARCH IN A/K PROSTHESIS

Historically, the development of the improved A/K prosthesis had been a result of the skill and imagination of the privately working prosthetists. Involvement of scientist and engineer in the use of engineering principles to study the human locomotion system and guide the prosthesis design began after world war II, when the amputee population had largely increased as the result of the war. The first extensive quantitative measurement and the mechanical analysis of the human biped gait was undertaken in the late 40's and early 50's at the University of California at Berkeley[2]. These studies provided a much needed data base

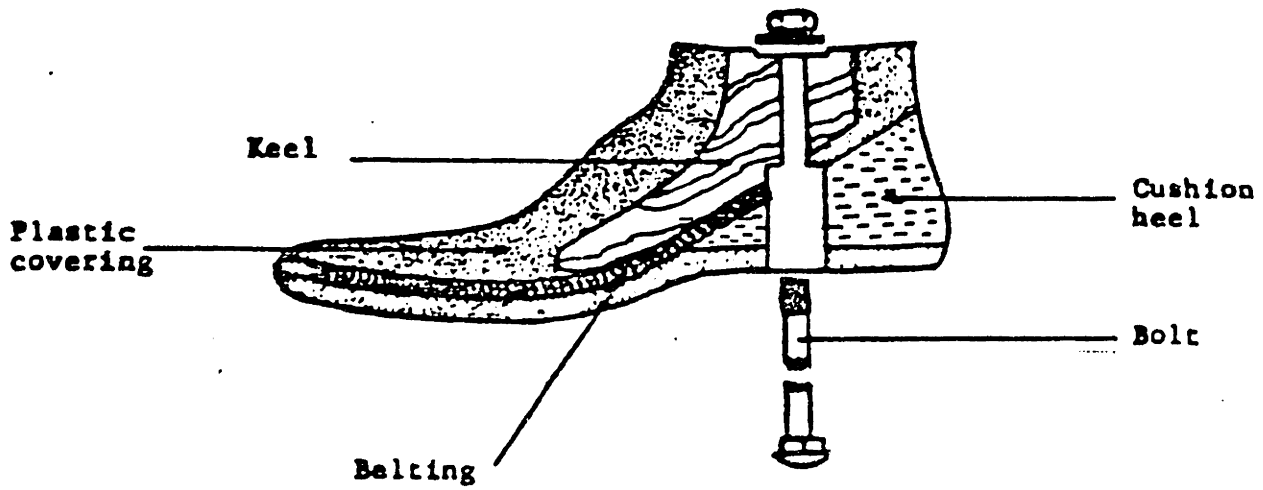


FIGURE II-11

SACH Foot ( Solid Ankle Cushion Heel )

to change the cut and try process of prosthesis design.

To further improve the design process, Flowers[5] and Grimes[7] developed an amputee-interactive simulation system. The system contains the most sophisticated prosthesis to date. Powered by a 1000 psi(6895 kPa) hydraulic pressure source, it is capable of producing torques large enough for ordinary activities like level walking and stair climbing. The unique thing about the prosthesis is its computer controllability, which enables the system to be programmed to simulate any known mechanisms and generate an almost endless list of new ones. The system is a valuable research tool. New strategy can be evaluated and modified conveniently without actually building the hardware. Based on this system, Grimes[7] developed a versatile controller that allow an amputee to climb up and walk down stairs and ramps as well as level walking.

Much of the attention in A/K prosthesis research has also been directed to the knee unit. Continuous research is still being conducted on upgrading the conventional mechanisms. But more effort is concentrated on new types of design that provide more functions and better interact with the amputee. It is unfortunate that within the bound of current technology, a self-contained, powered prosthesis is not possible. But a versatile passive knee unit, which produces resistive torques only, would also bring about significant improvement to the performance of the current A/K

prosthesis.

Using magnetic particle brake as the damping device for the knee unit has shown great promises. The device is small, light weighted and can be powered by a relatively small battery, with the braking torque being controlled by the amount of current passing through the brake. A micro-computer can be easily combined with the device, making the system a versatile and individualizable prosthesis. And the most attractive feature is the whole prosthesis can be made self-contained. Such effort is currently undertaken by the MIT Knee Group and a prototype system has been developed.

With the progress in hardware design, the knee unit control scheme, which largely determines the performance of the prosthesis, is also under development.

The first problem to address is the identification of the amputee's intent through measurable signals and the formation of the command input for controller. One way of reading the amputee's intent is to use myoelectricity (EMG) of the stump muscle. Research in EMG study has shown that proper processing of EMG signal could reveal the intended magnitude of the force generated. In the case of arm prosthesis, EMG has been proven to be a good control signal. The amputee control the arm in the similar way normal person does. But use of EMG for leg prosthesis has several



problems. Since surface electrodes have a very limited pickup depth, the relatively large contributions of the deep muscles of the hip and surface muscles covered by significant fat or fascia can be difficult or impossible to read. Also, accurate interpretation of the EMG signal requires the complicated circuit that is too large to be included into the prosthesis. For all these reasons, EMG control needs more research before it can be practically used. It may, someday, be the most attractive method.

Without the instantaneous intent of the amputee as the control signal, one of the alternative in the past was to use the sound leg kinematics(Echo Control). By mounting sensors on the sound leg, its kinematics were recorded and used as the control signal for the following prosthetic step. In this way, if the sound leg always takes the first step, the gait produced is cross symmetric. Grimes[7] used this method in the stance phase control of the Flowers' Knee[5], which allows the amputee to flex and re-extend their knee joint the same way sound leg does during stance. He also divided the walking process into a finite number of modes and developed predetermined control strategy for each mode. The resulted gait is very close to the normal one, in spite of the fact that there is no direct voluntary control.

Linske[11] used the echo control idea for the swing phase of level walking. He simplified the method by recording only three pieces of information from the sound leg, the maximum knee flexion angle, the time at maximum knee flexion and total time of swing. He then used these data to specify three points of the swing phase knee angle trajectory for the prosthetic side and used cubic splines to reconstruct the remainder of the trajectory. Thus, he formed the target for the controller to follow. His method greatly simplified the echo control algorithm. His walking trials produced a quite symmetric gait during the swing phase of level walking.

Another category of control scheme involves setting the knee torque as a function of instantaneous knee kinematics of the prosthetic side. This can be thought as adjusting the mechanical impedance of the knee joint. The advantage of this scheme is its simplicity. No sound knee goniometer is required. Darling[3] determined swing phase knee torque as a function of instantaneous knee angle and angular velocity. Using the criterion that the prosthetic knee angle best approximates the sound one, iterations were carried to update damping coefficient after each step. Although prosthetic angle trajectory did not match the sound one exactly, some convergence of the damping did allow him to determine the final impedance function. His experiments were conducted in a narrow cadence range of 100--110 steps

per minute. He recommended that several damping functions be found, each suitable in a particular cadence band.

There are other types of control schemes where more kinematic variables are included in determining the knee torque. One such example is the scheme developed by Ishai and Bar([1] and [9]). Dividing the walking cycle into 10 functional stages, the damping level for each stage was chosen. The unique part of their scheme is that the hip angle played an important role in determining the walking stages. This allowed the amputee to control the walking more consciously. The level of damping for each stage, however, is preset and can not change its value as the speed changes. Comparing with the conventional prosthesis, their system produced gait having lower effort requirement from prosthetic side hip muscles for knee stabilization and bending initiation.

In view of the previous schemes, most of them require sensors to be used at places other than the prosthesis itself. This makes the implementation of these schemes in practice very difficult. Darling's[3] scheme is self-contained, but not optimal for a wide range of walking speed. A self-contained, speed adaptive controller is yet to be developed.

### CHAPTER 3 CONTROL SCHEME DEVELOPMENT

#### GENERAL DESCRIPTION

The general guiding principle is that the controller should recognize the intent of the amputee and cause the prosthesis to perform the desired function. It should reduce the effort by the amputee and improve the stability and cosmesis of the amputee's gait.

The command input for the controller is the central issue. Although it is the ultimate goal that a signal reflecting the instantaneous will of the person control the prosthesis, such a signal is yet to be found. At present, alternatives must be used. Fortunately for walking, very good ones are available. Walking is a periodic process with same events occurring cycle after cycle. The cycle is further divided to several phases with each phase having its own specific function. The variation in the gait, such as joint angle and joint moment, happens only when the walking speed or the environment changes. The functional requirements for each phase and its speed dependencies are known through gait analysis. Therefore, the controller can be pre-programed to realize the appropriate function for each phase once the walking speed and the various walking events can be detected. This is the general philosophy behind the construction of the control scheme in this research.

Detection of major foot/floor events such as heel strike, heel off and toe off can be made by using a footswitch mounted on the prosthesis shoe. It gives the "on" and "off" signal indicating which part of the foot is in contact with the ground. With these events being detected, stance and swing phase is then determined. The stance phase is further divided into two periods of different function. The weight bearing period begins after heel strike and occupies about 80% of the phase. The knee must be locked during this period to maintain stability. The knee buckling period fills the last 20% of the phase before toe off, where knee joint should be free of friction to allow knee break. Use of a weight sensor on the shank can distinguish these two periods.

The control strategy for stance is then simple. The knee is as long as heel is in touch with the ground. After heel off, the knee remains locked until the axial force on the shank drops below a threshold level (which is chosen to be 30 pounds in this case). Once the shank weight sensor indicates the weight has been shifted to the opposite leg, the knee torque is set to be a small value to allow knee bucking.

For the swing phase, the task is to achieve the normal looking kinematics. The controller must generate the proper torque profile at the knee joint to regulate heel rise and assure smooth knee extension during the desired time of swing. This profile is largely dependent on the walking

speed because the driving force from the hip and the kinematics are speed dependent. Therefore, the controller must be able to predict the intended walking speed and subsequently create the required profile.

#### SPEED DETECTION ALGORITHM

First consider the process during which the amputee decides to change his walking speed. It usually starts somewhere in stance phase. After opposite heel strike, the body weight begins to be shifted to the opposite leg. The hip muscle exerts a flexive torque to initiate the knee buckling and eventually lifts the leg to the air. The faster the person intends to walk, the larger the flexive torque, and consequently, the larger the hip flexion velocity. This information can hardly be used because mounting sensors on the hip joint would not result in a self-contained prosthesis. But since the prosthesis shank is linked to the stump thigh through the socket as a two-link chain system, one should be able to read the difference in the magnitude of the hip flexive torque through the difference in the shank motion, provided that other factors such as knee torque, initial hip and knee angles and angular velocities remain about the same during this period or their variations are known. Intuitively, the larger the hip flexive torque, the faster the shank motion will be.

With above analysis as a clue, an experiment with the amputee was conducted to search for speed indicators. The subject was asked to walk with the experimental prosthesis at three different speed : fast, moderate and slow. For stance phase, the knee was locked during weight bearing period and set free when shank sensor indicates the load is less than 30 pounds. The damping scheme for swing phase is similar to that of his regular prosthesis. For each speed, damping was adjusted "off line" to the best satisfaction of the amputee. Signals on all the sensors were recorded and closely examined afterwards. Special attention was directed to the knee bucking period to see any correlation between those signals and the actual walking speed.

Figuer III-1 shows the shank axial load and the knee angle during a typical walking cycle. The load reaches its peak at the middle of the stance phase and decreases towards the end of the stance. Toe-lift time ( $\Delta t$ ) is defined as the time in the late stance when the shank axial load (or weight on the leg) is less than 30 pounds. During this period, most of the body weight has been shifted to the oppsite leg. The knee starts buckling during this period. The rate of toe-lift can be defined as the threshold value (30 pounds) divided by the toe-lift time. The walking speed was not measured directly. Instead, the stide time, which is the inversal of the cadence, was used as reflection of speed.

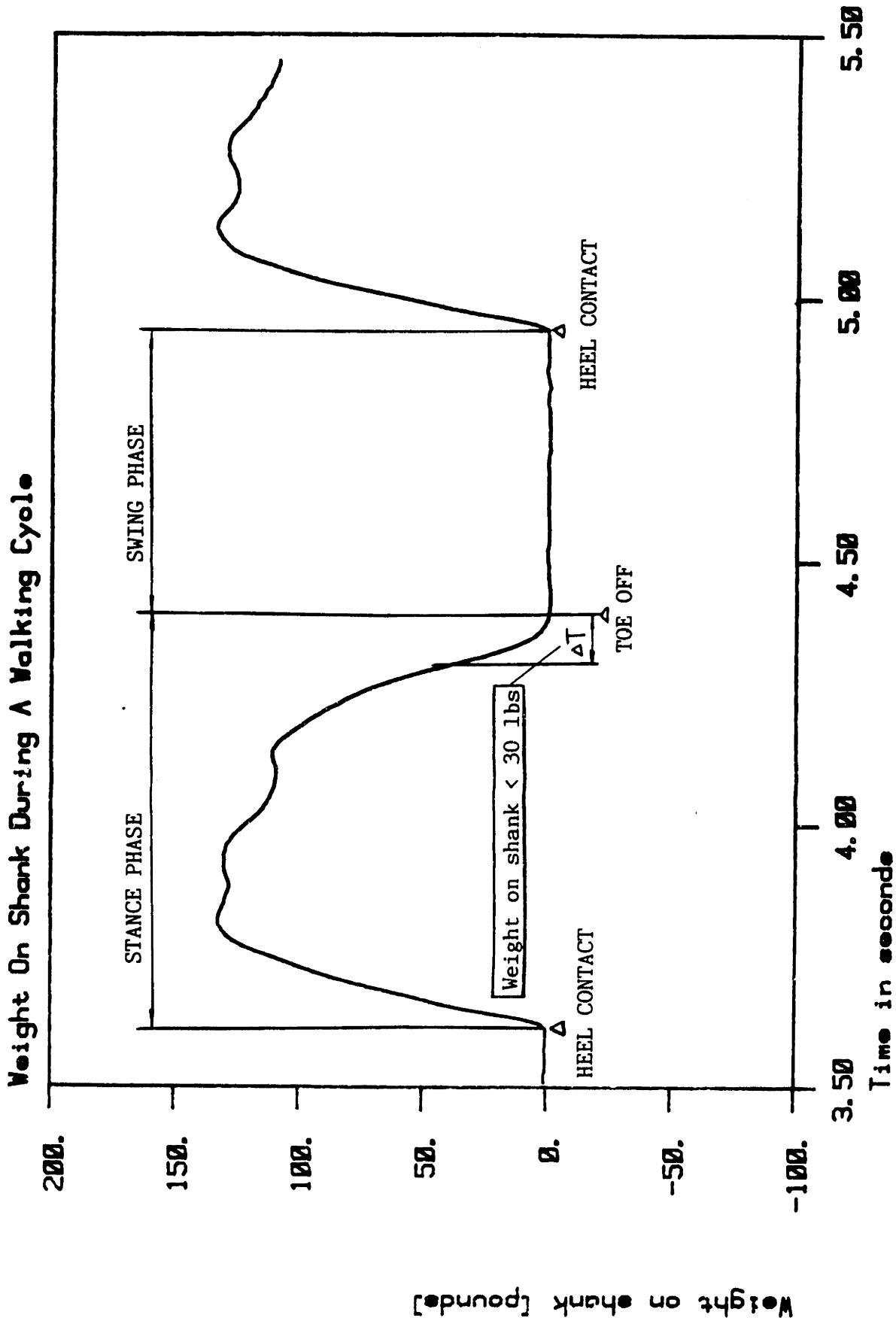


FIGURE III-1 Weight on the shank during a walking cycle



Table III-1 shows how the toe-lift time varies for three different speeds.

TABLE III-1

Slow	STRIDE TIME	SWING TIME	TOE-LIFT TIME
	1.42	0.60	0.12
	1.43	0.60	0.12
	1.44	0.61	0.12
Mean value:	1.43	0.60	0.12
Medium	STRIDE TIME	SWING TIME	TOE-LIFT TIME
	1.20	0.56	0.09
	1.25	0.57	0.09
	1.25	0.57	0.10
Mean value:	1.22	0.57	0.09
Fast	STRIDE TIME	SWING TIME	TOE-LIFT TIME
	1.12	0.54	0.06
	1.12	0.55	0.06
	1.10	0.53	0.06
Mean value :	1.11	0.53	0.06

One can see that as the stride time decreases with increasing forward speed, the toe-lift time and swing time decreases. Figure III-2 plots the relationship between stride time and toe-lift time. The trend was consistent for tests on different days. No effort was made at this point to determine the exact relationship between the toe-lift

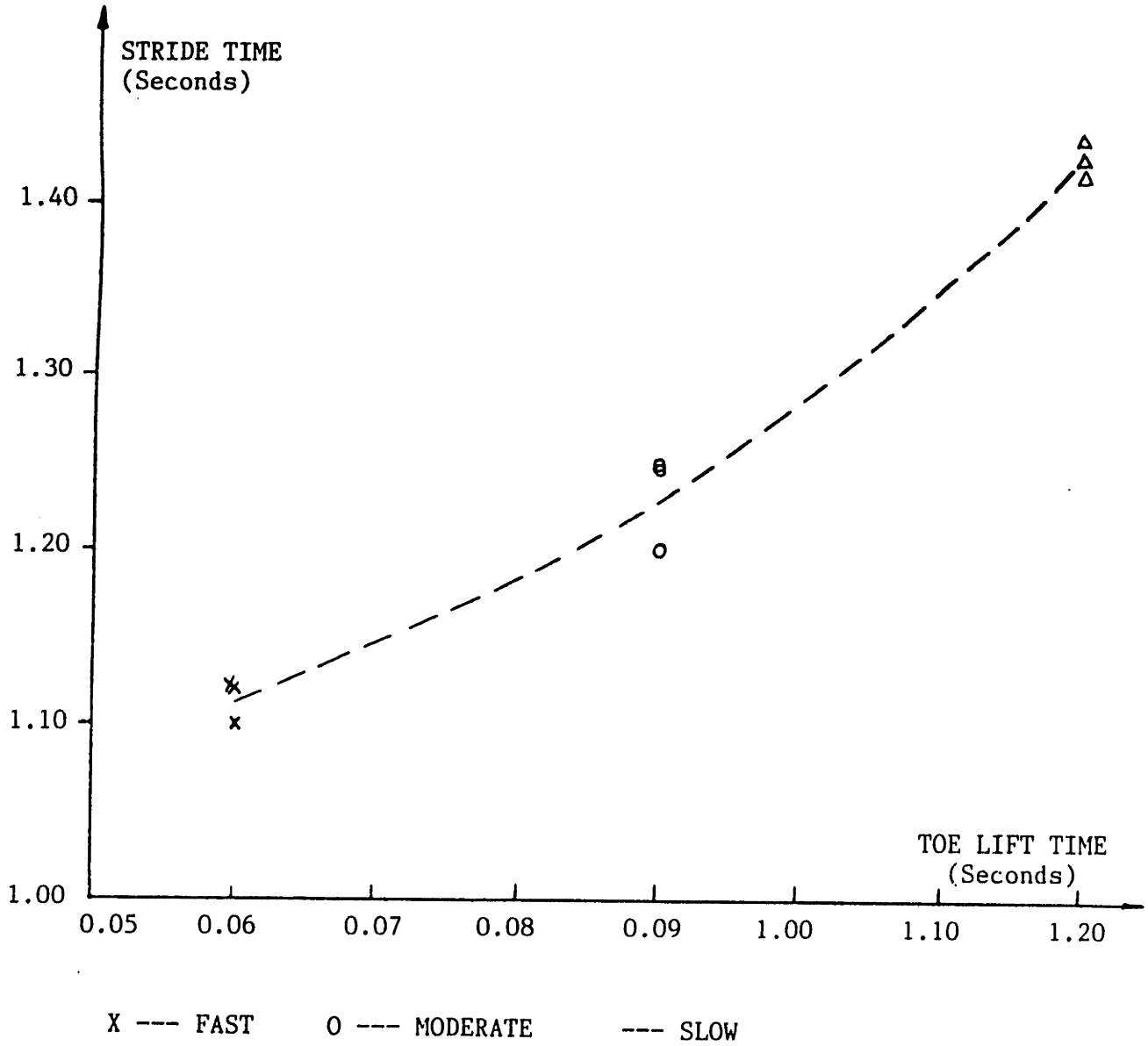


FIGURE III-2 STRIDE TIME vs TOE LIFT TIME

time and the swing time for the entire range of speed. The main concern here is to find a speed indicator. The general trend in figure III-2 does lead to the conclusion that rate of toe lift is a good indication of the intended walking speed. It can be used to predict the time of swing phase and consequently the desired kinematics for that period. The exact prediction function will be sought in later experiments.

#### TRAJECTORY CONTROL SCHEME

The idea of this scheme is to set the controller to control the motion of the leg to track a certain trajectory that has the characteristics of the normal gait, therefore to improve the A/K gait cosmesis.

The major characteristics of the knee angle trajectory in swing phase includes :

1. Proper amount of flexion in the first part of swing phase to guarantee ground clearance.
2. Smooth extension in the second half of swing with zero velocity at full extension.
3. This flexion-extension should be executed in the proper elapsed time.

The above requirements essentially define the maximum knee flexion as well as the time at which it happens, the total time of swing, the knee angle and angular velocity at the end of swing. With the initial condition of knee angle

and knee angular velocity, the positions of three points in the trajectory and their corresponding slopes are defined. The remainder of the trajectory can be generated by a chosen criterion, which should be aimed at helping to accomplish the desired characteristics and should be within the capacity of the controller. In Linsk's scheme, a linear acceleration curve fitting was used. Two third order polynomials (cubic spline) were fitted to the flexion and extension trajectories with their coefficients being determined by boundary conditions : ① initial knee angle and angular velocity, ② maximum knee flexion with zero angular velocity, ③ zero knee angle and angular velocity at the full extension. Figure III-3 further illustrates this method. Calculation of the splines' coefficients is as following:

$$\text{The spline : } \begin{cases} \theta = a*t^3 + b*t^2 + c*t + d \\ \dot{\theta} = 3*a*t^2 + 2*b*t + c \end{cases} \quad (1)$$

where  $\theta$  : knee angle

$\dot{\theta}$  : knee angular velocity

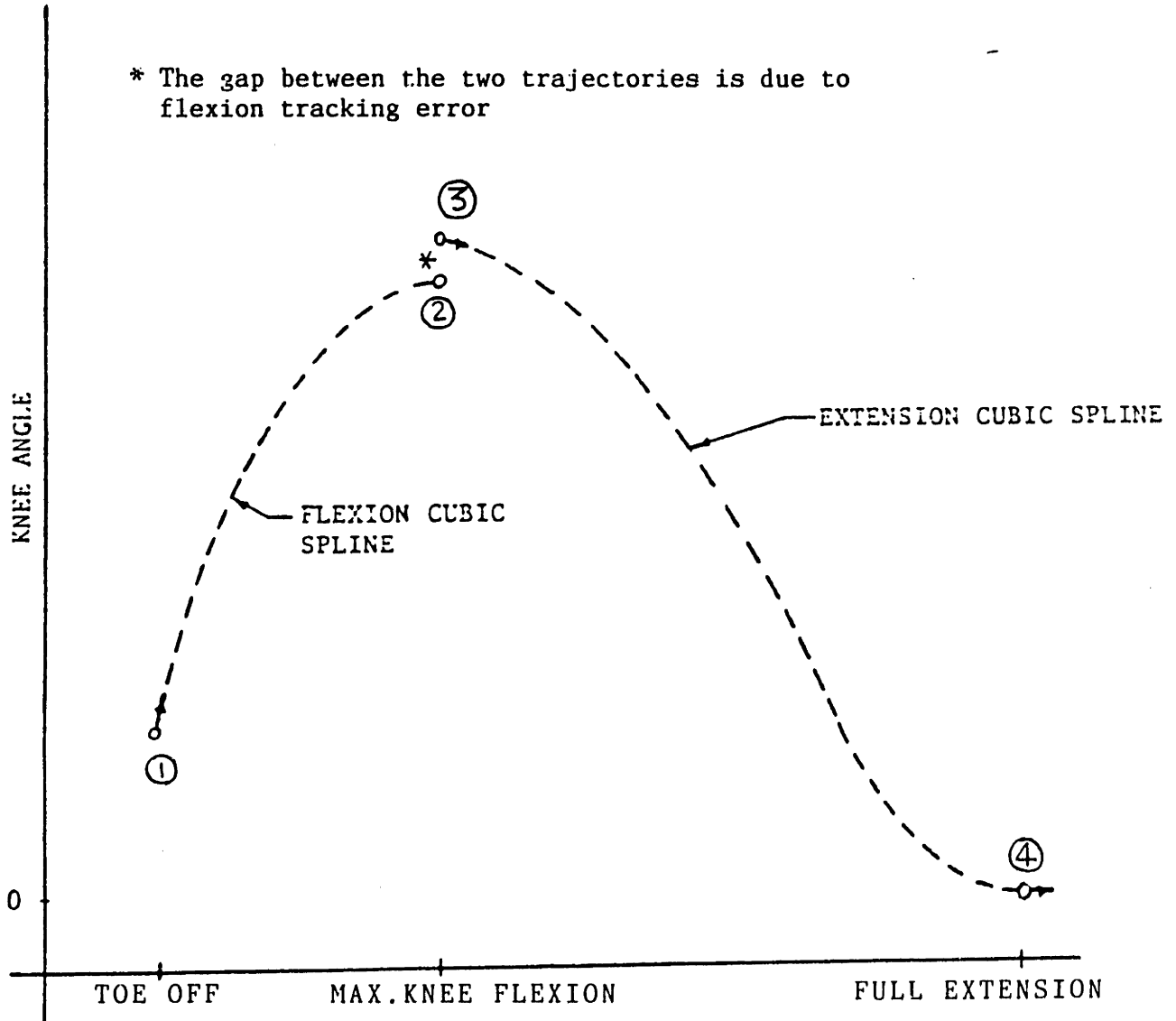
t : time

a, b, c, d : spline coefficients

The knowns : at toe off,  $t=0$

$$\theta = \theta_0$$

$$\dot{\theta} = \dot{\theta}_0$$



- ① Starting point : Value = Knee Angle @ Toe Off  
Slope = Knee velocity @ Toe Off
- ② Ending point : Value = Desired Max. Knee Flexion  
Slope = 0
- ③ Starting point : Value = Knee angle @ M.K.F.  
Slope = knee velocity @ M.K.F.
- ④ Ending point : Value = 0  
Slope = 0

FIGURE III-3 Construction of the swing phase target trajectory [11]

at max. knee flexion:  $t = t_m$

$$\theta = \theta_m$$

$$\dot{\theta} = 0$$

at full extension:  $t = t_n$

$$\theta = 0$$

$$\dot{\theta} = 0$$

Substitute these conditions into (1), we have

For flexion spline :  $a_1 = \dot{\theta}_0/t_m^2 - 2(\theta_m - \theta_0)/t_m^3$

$$b_1 = -2\dot{\theta}_0/t_m^2 + 3(\theta_m - \theta_0)/t_m^3$$

$$c_1 = \dot{\theta}_0, \quad d_1 = \theta_0$$

For extension spline:  $a_2 = 2\theta_m/(T_s - T_m)^3$

$$b_2 = -3\theta_m/(T_s - T_m)^3$$

$$c_2 = 0$$

$$d_2 = \theta_m$$

With the target trajectory chosen, a closed loop control system must be formed to execute the task. Figure III-4 shows its structure. It is a trajectory control system with both position and velocity feedback. Such a system was chosen to insure that the shank follow the trajectory smoothly, passing every point with the desired velocity as well as the desired position. Position feedback alone would have provided less rigorous control of velocity.

At every sampling moment, the desired and actual position and velocity are compared and if error exists, a control signal (current to the particle brake) is sent to the brake to attempt to drive the error to zero. That exact formula of calculation is :

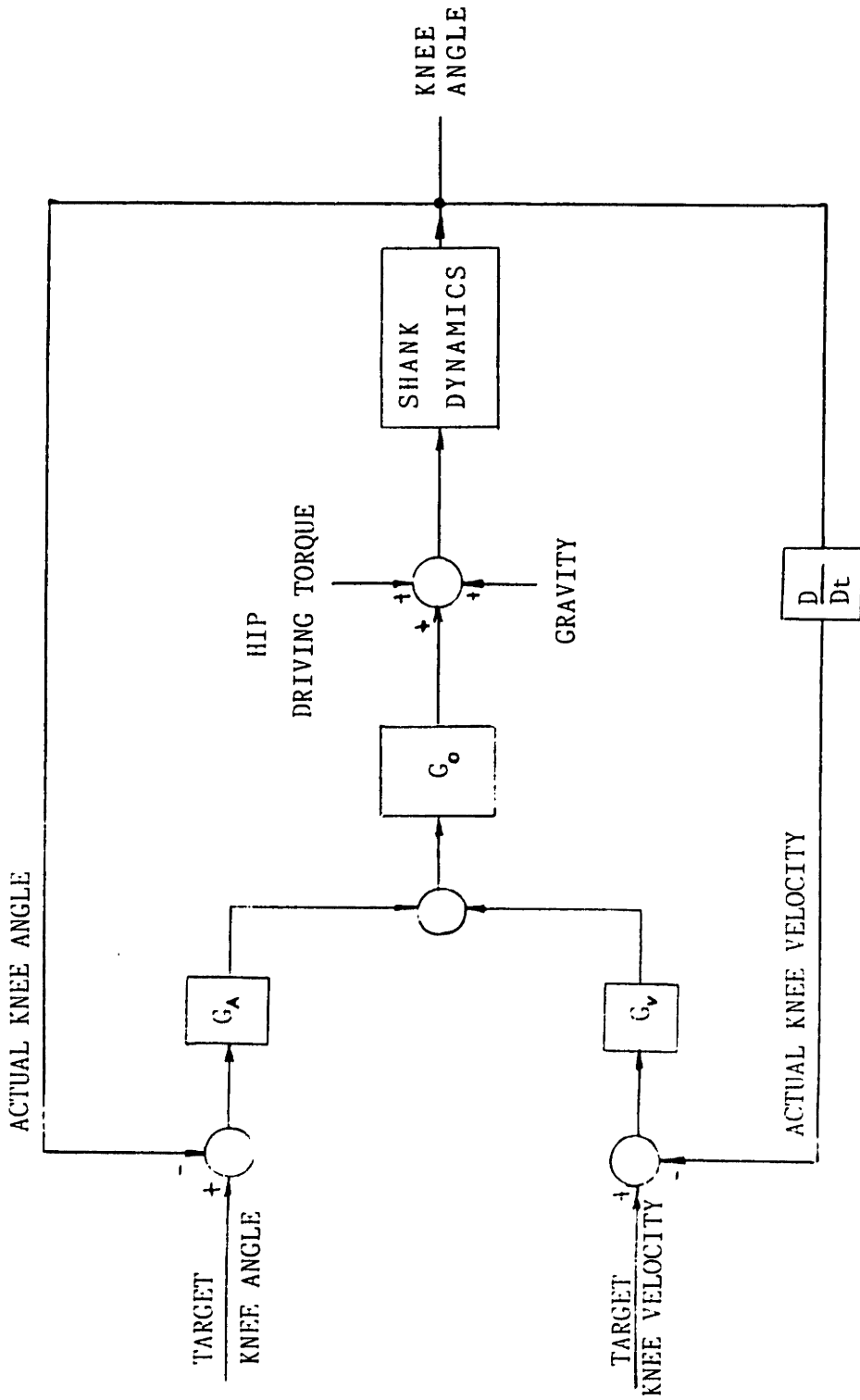


FIGURE III-4 TRAJECTORY CONTROLLER BLOCK DIAGRAM

$$T = G_o * [ G_p * ( \theta_d - \theta_a ) + G_v * ( \dot{\theta}_d - \dot{\theta}_a ) ] \quad (2)$$

$\theta_d$  : target knee angle

$\theta_a$  : actual knee angle

$\dot{\theta}_d$  : target knee velocity

$\dot{\theta}_a$  : actual knee velocity

$G_p$  : position gain

$G_v$  : velocity gain

$G_o$  : overall gain

The driving force for the system comes from the hip muscle. The controller only regulates the power being dissipated at the knee joint. Therefore, the controller, no matter what the values of loop gains are, can only slow down the motion of the shank and can not cause instability. When the tracking error is such that active power is needed to speed up the shank, the best the controller can do is to set the braking torque to zero. In normal walking, power requirement at the knee for most part of swing phase is mainly passive.

The heart of this type of control scheme is the target trajectory, which is created by three parameters, the maximum knee flexion angle, the time it occurred, and the total time of swing. Remember that the value of these parameters depends on the walking speed. And the intended walking speed can be detected by the measured toe lift time. Therefore, The value of the toe lift time should be used to predict the swing parameters, once the exact relationship between them is known.



It was decided to determine this relationship experimentally. An iteration algorithm was created to do the job. Figure III-5 illustrates how this works. It starts with an initial estimate, which can be obtained from the result of the experiment that is done without the close loop control. At the beginning of the swing phase, the controller projects a trajectory according to this initial guess and executes the close loop control during the swing. If the predictions were not the desired ones, the prosthesis user would try to overpower the knee controller. And at this point, if the knee controller is purposefully made less robust than it should be, it is expected the actual values being closer to what is desired by the amputee.

Therefore, the actual values can be used to correct the predicted ones. Once a more accurate relationship is obtained, the process can be repeated until convergence occurs.

#### TORQUE CONTROL SCHEME

The idea of this scheme is to regulate the energy dissipated at the knee directly by making the instantaneous knee torque a function of the instantaneous knee angle and angular velocity only. The general expression of the knee torque is as following :

$$T = B(\theta, \dot{\theta}) * f(\dot{\theta}) \quad (3)$$

where T : calculated knee torque

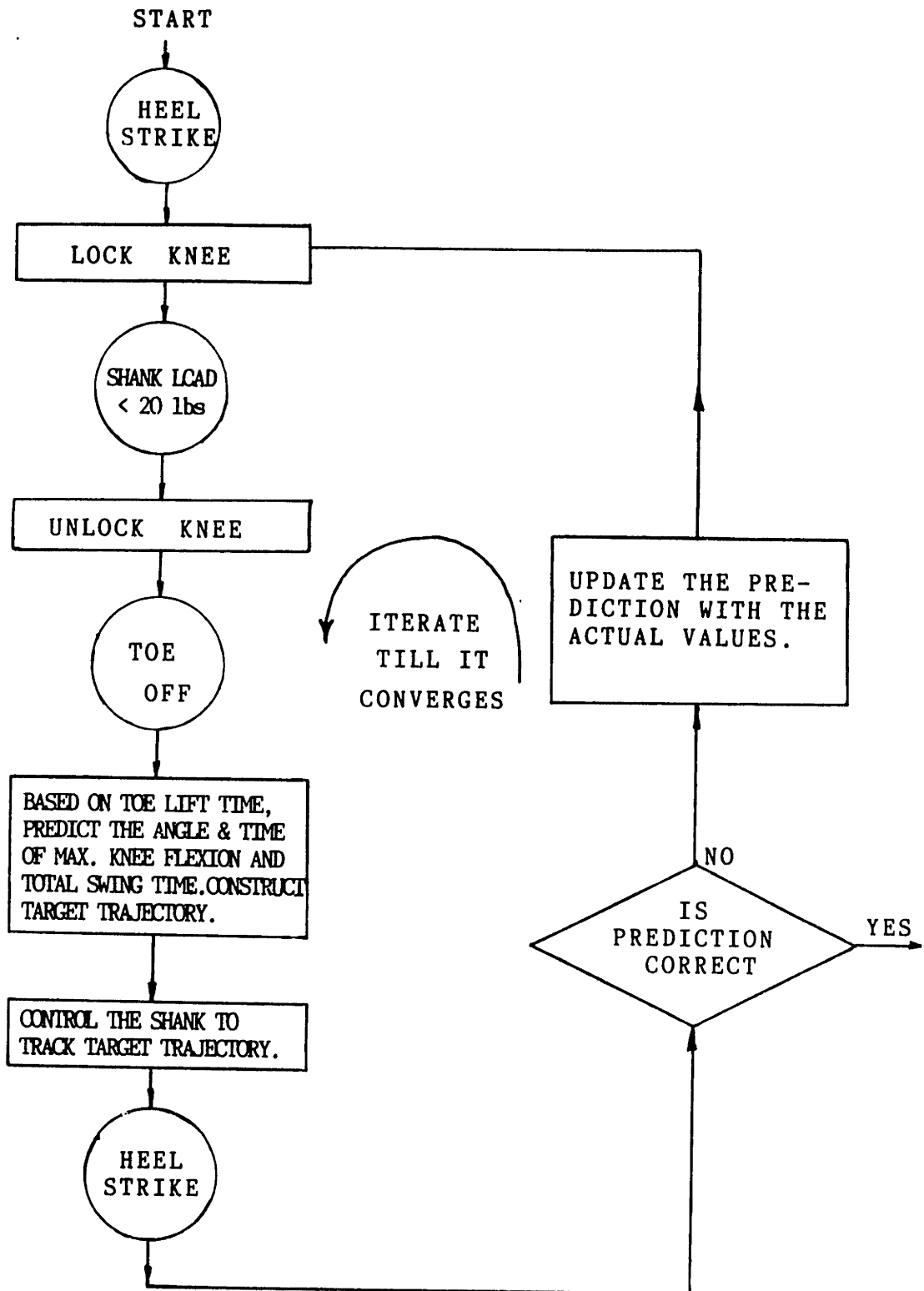


Fig. III-5 Flow Chart of the Iteration Routine

$B$  : damping coefficient

$\theta$ : instantaneous knee angle

$\dot{\theta}$ : instantaneous knee velocity

Note there is no specific target trajectory for the shank to track. The normal trajectory can be achieved by choosing the appropriate form of control functions. This can be considered as regulating the mechanical impedance at the knee joint. The advantage of this type of scheme is that it would give a person a direct and more consistent response any time he swings the leg. This may give an amputee more confidence about the prosthesis. While for the trajectory control scheme, once the target trajectory is chosen, the shank is forced to follow it. Any desired change of target trajectory afterwards is not tolerated.

The key to this scheme is the control function. Many attempts have been made to find the "ideal" damping function. Most of them chose models in which the thigh and shank were regarded as a two link chain system and the damping coefficient  $B$  was inversely calculated from the normal kinematics. The computed damping coefficient is a very complicated function of knee angle and knee velocity and is very sensitive to the variations in the input kinematics. However, these results give only a general idea and can not be used directly in practice because hip torque input to the system by an amputee varies enormously from person to person and can hardly be predicted.

Many experiments have also been performed to develop "ideal" dampers. The most common one is the viscous damping type, with  $T = B * \dot{\theta}$ , where  $B$  is a constant. But with viscous damping, speed adaptivity is only partially achieved. Darling [3] used the form  $T = B(\theta) * \dot{\theta}^2$ , where the damping coefficient was a function of knee angle. The use of knee velocity square was believed to increase the speed responsiveness of the controller. However, his experiments were conducted in a narrow speed range and it was not clear that this scheme would suit for all speeds.

Both theoretical analysis and experimental investigation on the "ideal" damping function indicated the complexity of this function. This is because by making the function depend on the instantaneous knee angle  $\theta$  and velocity  $\dot{\theta}$  only, the desired changes in the damping function as the intended walking speed changes can not be realized directly. With the speed detection method developed, one can simplify this process by two separate steps. First, find the set of the damping coefficient  $B$  for one speed. Then, scale  $B$  by the speed indicator while keep  $B$  the same function of  $\theta$  and  $\dot{\theta}$ . That is :

$$T = B(\theta, \dot{\theta}) * f(v) \quad (4)$$

where  $B$  : Damping coefficient function for speed  $0$

$v$  : Intended walking speed

$f(v)$  : Scaling function

Since intended walking speed is detected before toe lift time  $\Delta t$ , the equation (4) can be expressed as :

$$T = B(\theta, \dot{\theta}) * f(\Delta t) \quad (5)$$

Tasks for both steps were done experimentally in this thesis. Common viscous damping scheme, where B is a constant, is used for the first step. Then scaling function is to be determined from two proposed forms. They are :

$$f_1(\Delta t) = \frac{t_0}{(\Delta t + t_0)} \quad \text{or} \quad f_2(\Delta t) = \frac{t_0^2}{(\Delta t + t_0)^2}$$

where  $\Delta t$  is the toe lift time and  $t_0$  is the constant to be decided. Both forms cause damping to be increased as speed increases (corresponding to  $\Delta t$ 's decrease). The value of  $t_0$  actually determines the damping coefficient for each speed.

#### COMPUTER IMPLEMENTATION

The program architecture is illustrated in Figure III-6. It consists of the main routine, the MODTCR which divides the walking cycle into a number of functional modes and determines the present mode of the leg, and the CONTROL which calculates and sends control signals to the brake according to the mode value and the chosen scheme.

The main routine starts the real time clock and checks the program execution at every sampling period. If the control program execution falls behind the real time clock or the prosthesis user wishes to terminate the program execution because of a stumble for example, the knee will be

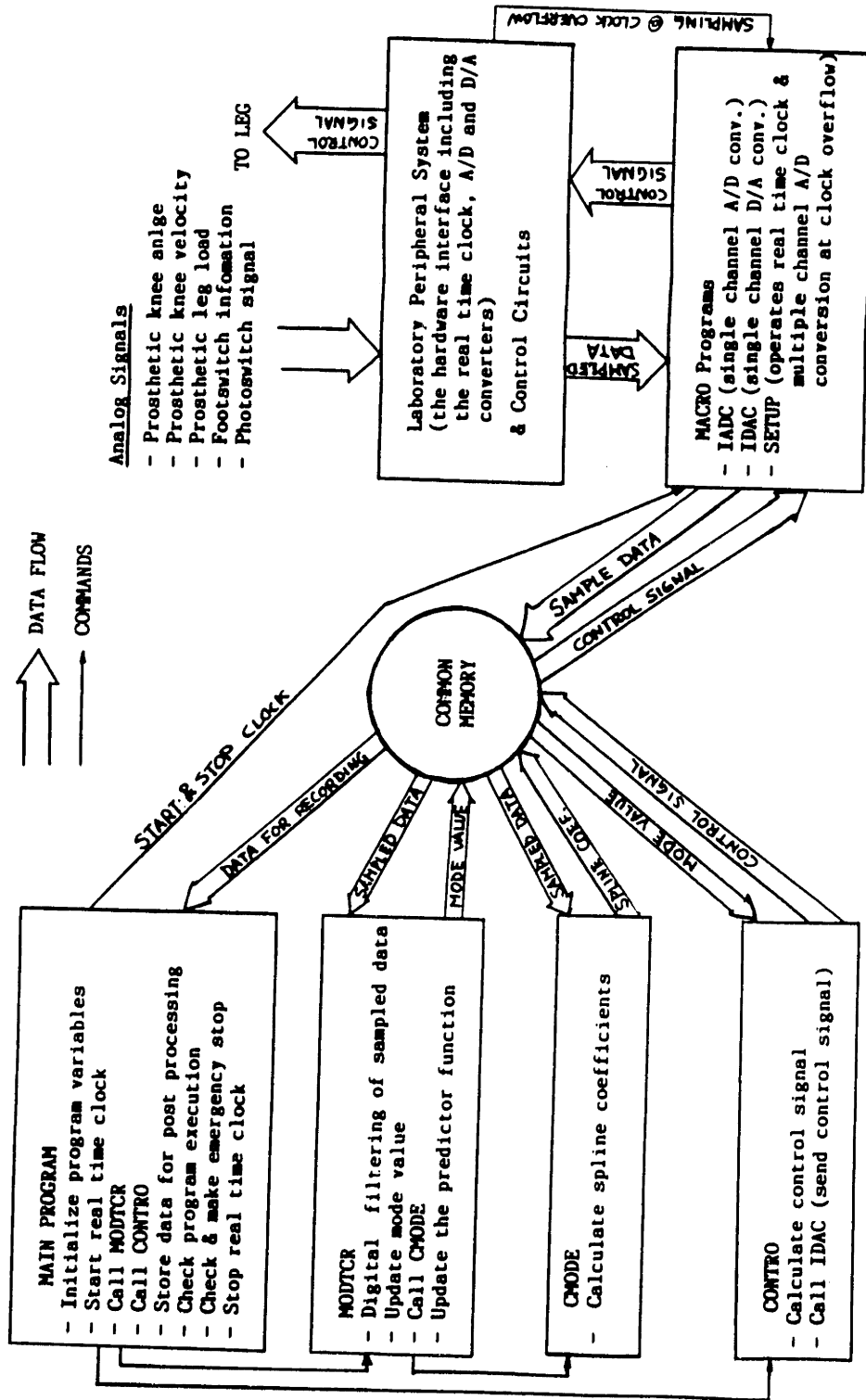


FIGURE III-6 ALGORITHM ARCHITECTURE

locked and program will be stopped. The clock operates at 100 Hz and sampling of the seven A/D channels occurs automatically at clock overflow.

The heart of the program is the mode detector MODTCR. Nine modes were used here. Table III-2 and Figure III-7 show in detail how the mode value is selected according to the state of the leg. The footswitch plays a major role. In continuous walking the mode value is incremented sequentially from 1 to mode 6. But out of sequence changes, such as sitting down, are also recognized. In this case, the person can lift the leg directly from heel contact to the whole leg off rather than heel off to toe contact to toe off, while mode value jumps from 1 to 7. The mode 8 and mode 9 only exist for one sampling, serving as calculation modes.

Note that the swing phase can end in two different ways. The more common way is that the knee reaches full extension before heel strike, giving the person a more secure feeling. The other is that the heel strike happens first before the knee is fully extended. This could occur when one is speeding up by cutting the swing phase.

Both control schemes share the same structure except for that the trajectory control will call CMODE for spline coefficients calculation at mode 3 and mode 5. In the CONTROL routine, the trajectory control scheme calculates the control signal by calculating the target knee angle and

TABLE III-2

MODE VALUE	KNEE ANGLE	FOOT & OTHERS	CONTROL SIGNAL PERCENTS OF MAX	CONDITIONS FOR MODE CHANGE
1	0	Heel in contact	Maximum (Lock knee)	If heel off, mode=2 If whole foot off, mode=7
2	0	Toe in contact W >30 lb	70 of max. (near lock)	If shank axial load W > 30 lbs, mode=3 If heel contact, mode=1
3	small	Toe in contact	10 (Unlock knee)	If toe off, mode=4 If W >30 lb, mode=2
4	vary	foot in the air	Depends on scheme	If knee velocity change sign, mode=5
5	vary	foot in the air	Depends on scheme	If knee angle < 5, mode=6; If heel strike, mode=8
6	0	foot in the air	Lock knee	If heel strike, mode = 9
7	vary	foot in the air	5 of Max.	If heel strike, mode = 1
8	0	Heel in contact	Lock knee	Immediate Update, mode = 1
9	0	Heel in contact	Lock knee	Immediate update, mode = 1



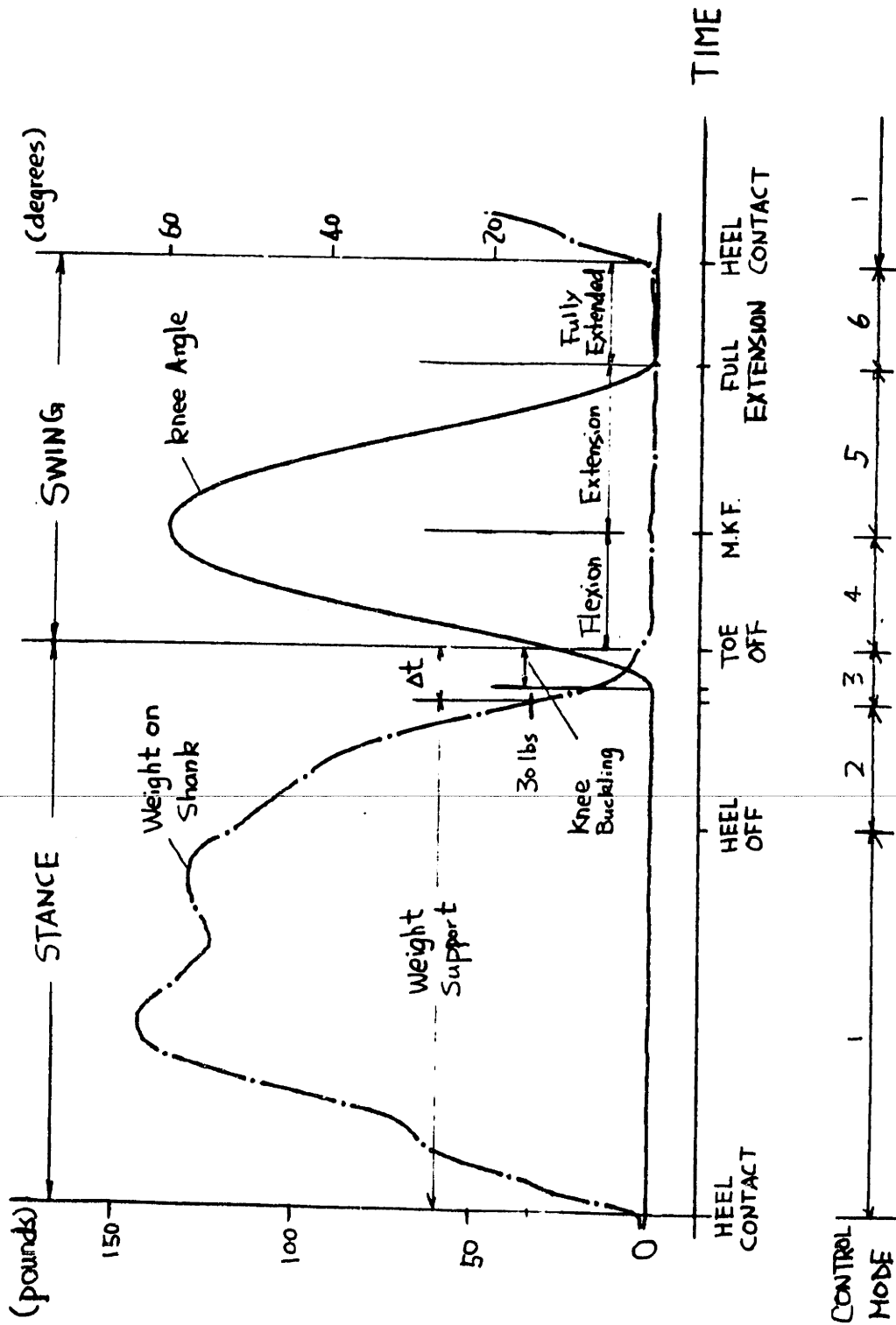


FIGURE III-7 Control Mode Value compared with the knee angle and shank axial load 8,9

velocity first while the torque control scheme calculates it directly from the instantaneous knee angle and velocity. Table III-2 lists the control signal for all mode values.

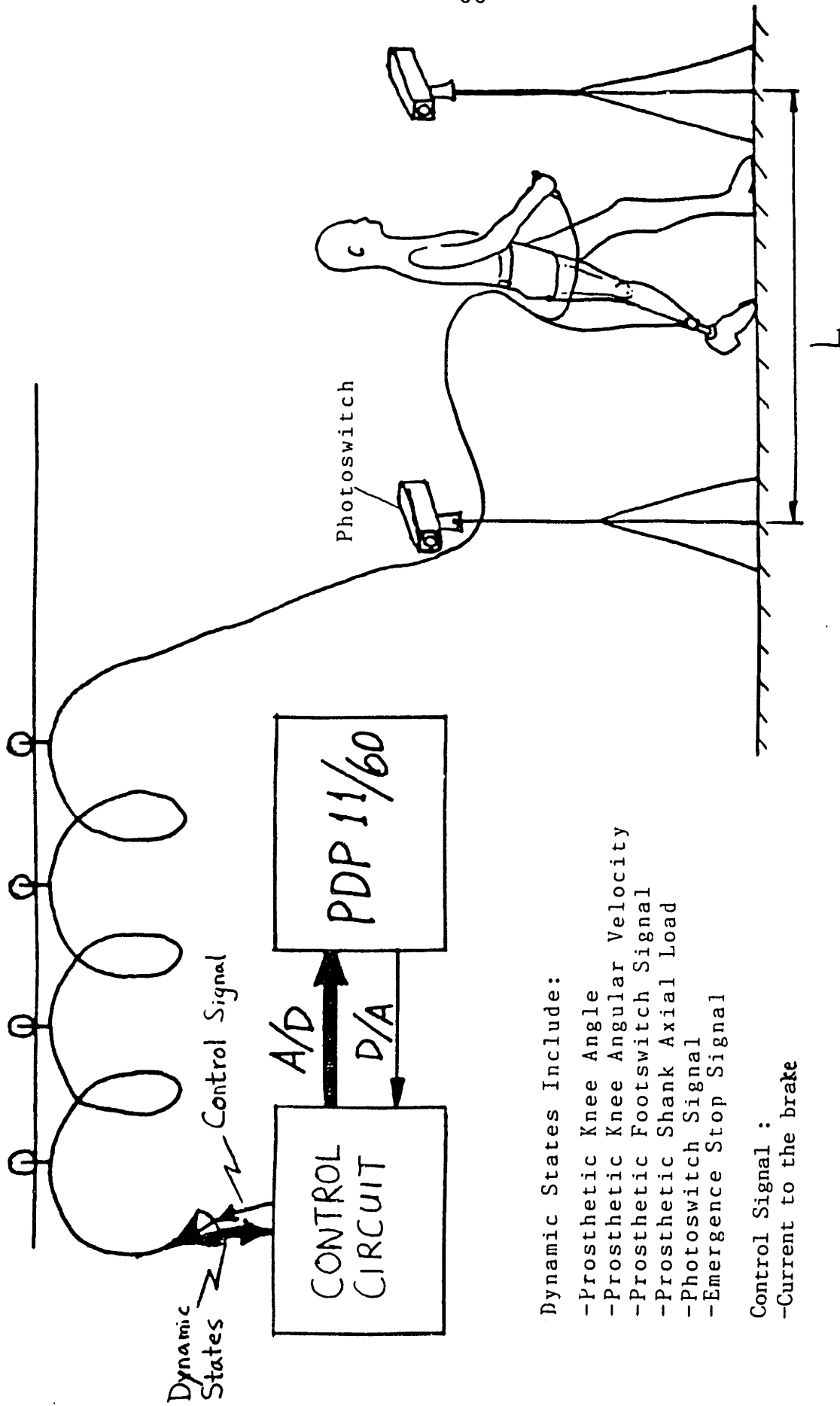
For the trajectory control scheme, the program also updates the swing phase kinematics prediction which includes swing time, maximum knee angle and the time it occurs, from the toe-lift time. It corrects the predicted values with actual ones unless the actual values are the result of an abnormal step (which happens when the prosthesis user terminates the step before it is finished).

## CHAPTER 4 THE WALKING TRIALS

With the control schemes developed, walking trials by two amputee subjects were conducted to evaluate their performance and make any necessary modifications. This process is absolutely required because the input to the prosthesis and reaction to the new control scheme from the amputee are almost impossible to predict. A simulation system was used, which consisted of the subjects, the experimental prosthesis, the computer containing the control scheme, and the control circuits interfacing with the prosthesis and the computer. The system is illustrated in Figure IV-1.

### THE INSTRUMENTED PROSTHESIS

The knee unit (Figure IV-2) was designed and built by Lampe[10] with the magnetic particle brake as its damping device. The braking torque is determined by the electric current applied to the brake. The working principle is that as the current energizes a magnetic field around a drag dish rigidly attached to the output shaft, the magnetic particles packed in the gap between the drag dish and the stationary surface in the housing align in the way that resists the relative motion between the dish and the stationary surface. The amplitude of the resistance torque is dependent on the



Dynamic States Include:

- Prosthetic Knee Angle
- Prosthetic Knee Angular Velocity
- Prosthetic Footswitch Signal
- Prosthetic Shank Axial Load
- Photoswitch Signal
- Emergence Stop Signal

Control Signal :

- Current to the brake

FIGURE IV-1 The Amputee-Interactive Simulation System

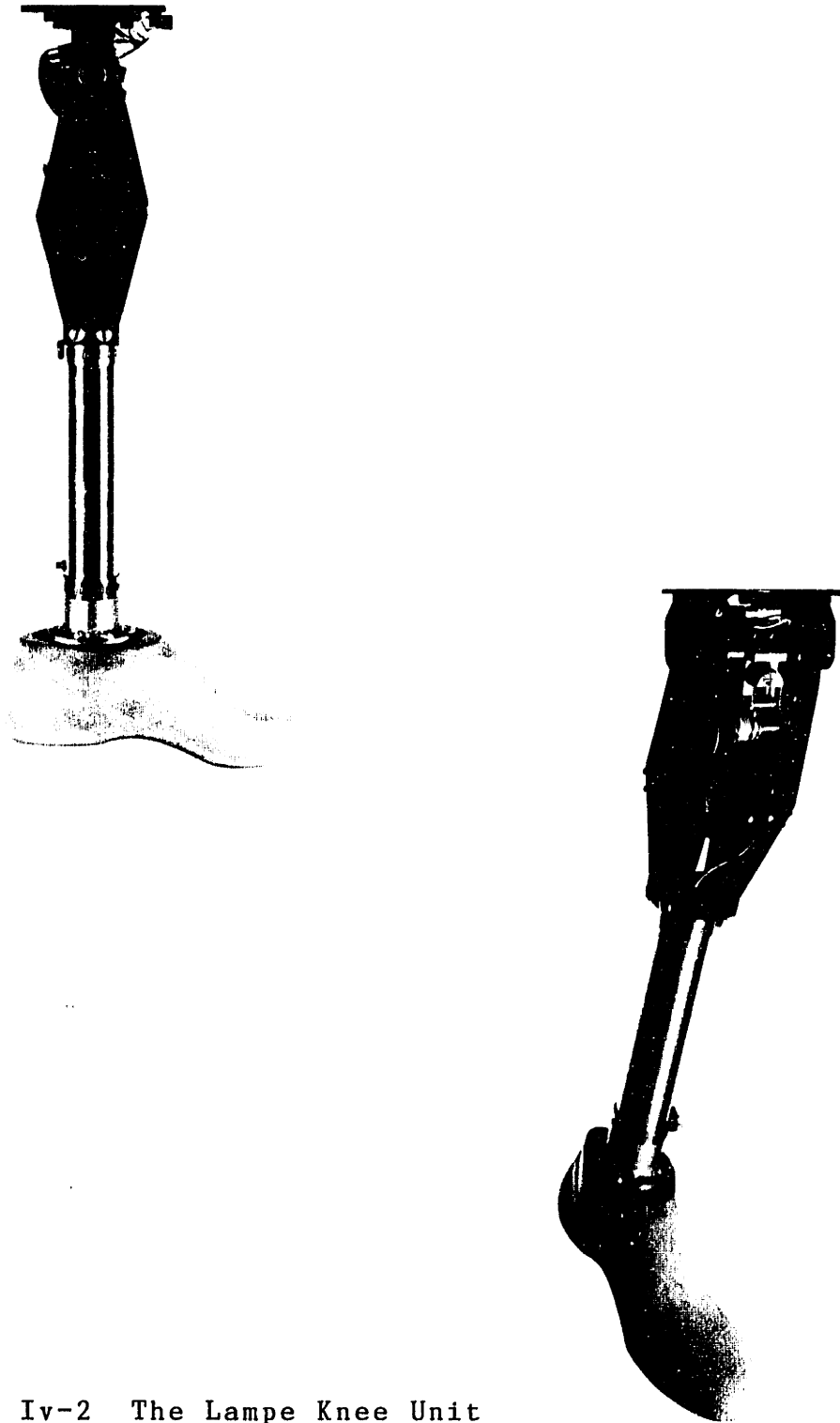


Fig. Iv-2 The Lampe Knee Unit

strength of the field which is related to the current. The relationship is approximately linear except at near the maximum torque region where saturation occurs. The brake used in the Lampe Leg (Figure IV-3) produces torques up to 40 in-lbs(4.52 Newton-meters) at the current of 400 mili-amps. Two-stage transmission with the total ratio of 7.5:1 was used to bring the maximum torque produced at the knee joint to 300 in-lbs(33.9 Newton-meters).

The sockets were custom designed by the prosthetists for each subject and fit well. The properties of the SACH feet such as the size and stiffness were chosen to the preference of the subjects. The whole prosthesis, including the socket, the knee unit, the SACH foot and shoe weighted 9.5 pounds(4.31 kg).

Various transducers mounted on the prosthesis provide the controller with the kinematic state of the leg. A precision rotary potentiometer connected with knee axis through chain transmission serves as a position transducers to monitor the knee angle. A load cell located in the middle of the shank gives the force along the shank. Foot-floor timing signals come from the footswitch positioned at the heel and toe of the shoe. This contact switch, which is similar to the ones used in a push button telephone, needs only a fraction of an ounce force for activation. All sensors are carefully calibrated and worked

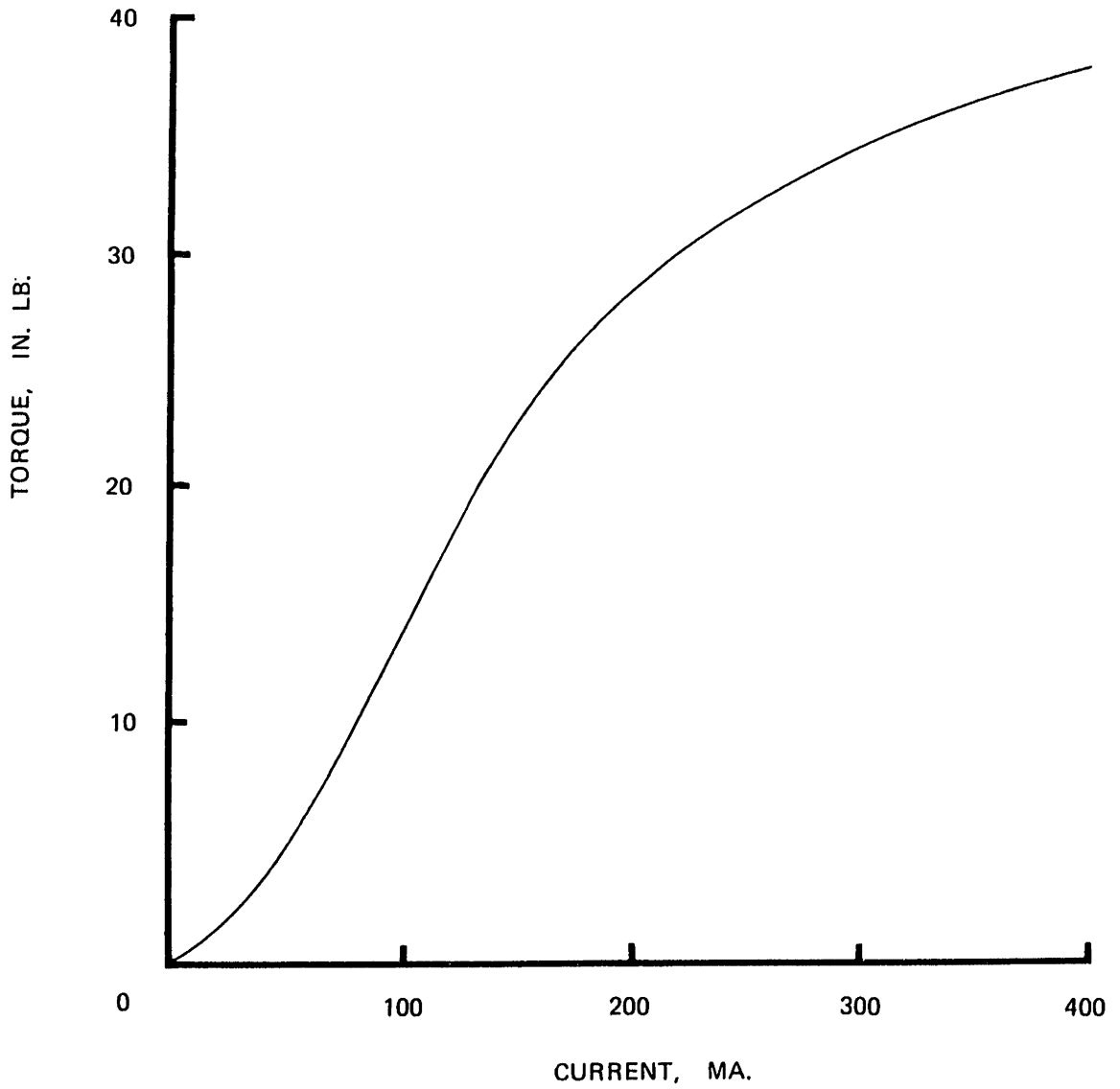
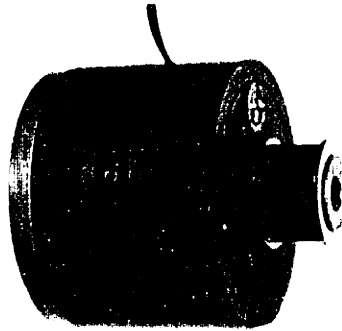


Figure IV-3 The Magnetic Particle Brake

reliably during the experiments.

The average walking speed over a distance of 18 feet was measured by using two photoswitches located at each end of the walkway. The switch both projects a light beam and receives its reflection. It gives an "on" output when reflection is stronger than an adjustable threshold, which occurs when the beam is blocked. Therefore, by setting a clock and monitoring the output of the switch, one can detect the moment when a person is passing by. The elapsed time taken to travel the distance between the two photoswitches is then available for calculation of speed.

#### THE COMPUTER AND INTERFACE BOARD

The interface board is used to produce the signals for the computer's A/D converter. It includes an analog differentiator to produce the needed knee angular velocity signal. It also contains the circuits for the computer to control the current to the brake. The dynamic behavior of the brake, approximated by a first order system with a time constant of 25 milliseconds, requires a voltage controlled current source so that the brake's torque can be updated at the 100 Hz sampling rate.



The PDP-11/60 computer in the rehabilitation laboratory performs the control functions. Its Laboratory Peripheral System (LPS) sets the real time clock and handle the A/D and D/A conversion. Program parameters are determined interactively via a terminal.

#### WALKING TRAILS

Two subjects participated the experiments. Kris had about 15 hours of experience while Steve had only two. They are both young and energetic. Socket alignment were carefully made each time. They were also given a safety button to hold in their hand. When activated, it will lock the knee. Figure IV-4 shows two of the trials.



FIGURE IV-4 Subject A walking with the Instrumented Prosthesis.

## CHAPTER 5 RESULTS, CONCLUSION AND RECOMMENDATIONS

### RESULTS

#### 1. Trajectory Control Scheme:

The results of amputee subjects' walking under the trajectory control scheme were unexpectedly inconsistent. Subjects had mixed reactions toward the scheme. The major problem with the scheme in the trial is that, on many occasions, although the motion of the prosthetic shank followed the predicted target trajectory well, both subjects complained that the prosthesis did not feel comfortable. Sometimes a low frequency chatter would occur in the motion of the shank, which forced the controller to set up a upper limit for the knee braking torque.

Recall the iteration routine that is used to determine the relationship between the key parameters of swing trajectory and the toe lift time. The controller's prediction of the maximum knee flexion angle and the time it occurred agreed well with the actual ones after a series of iterations. But the program had difficulties to predict the value of the time period when the knee flexion and extension occurred.

Figure V-1 shows the breakdown of the swing phase.  $T_t$  is the total swing time that is between toe off and heel contact, while  $T_s$  is the time for shank swing. Note there is a period ( $T_f$ ) when the knee is fully extended but the leg is still in the air. Although this period does not exist in normal person's walking, it takes place in amputee's walking.

At first, without knowing the existence of the  $T_f$  period, the controller expected the shank swing (includes flexion and extension) to occupy the entire swing time and thus the iteration routine was set to correct the predicted total swing time ( $T_t$ ) by the actual one. The final converged value of total swing time is much longer than the period in which shank swing (flexion and extension) took place.

Figure V-2 shows both the predicted and actual knee angle for one walking trial when subject A was asked to gradually increase his walking speed step by step. Initially, the two agreed well for low speed but the last two steps showed the actual shank swing took much less time than predicted. The predicted trajectory was not what he expected or desired. The subject chose to have the knee fully-extended for a rather long time before it touched the ground. The disagreement would not happen if the controller was robust enough. But controller had to set an upper limit for the knee torque to avoid shank chattering.

### DEFINITION OF PERIODS IN SWING PHASE

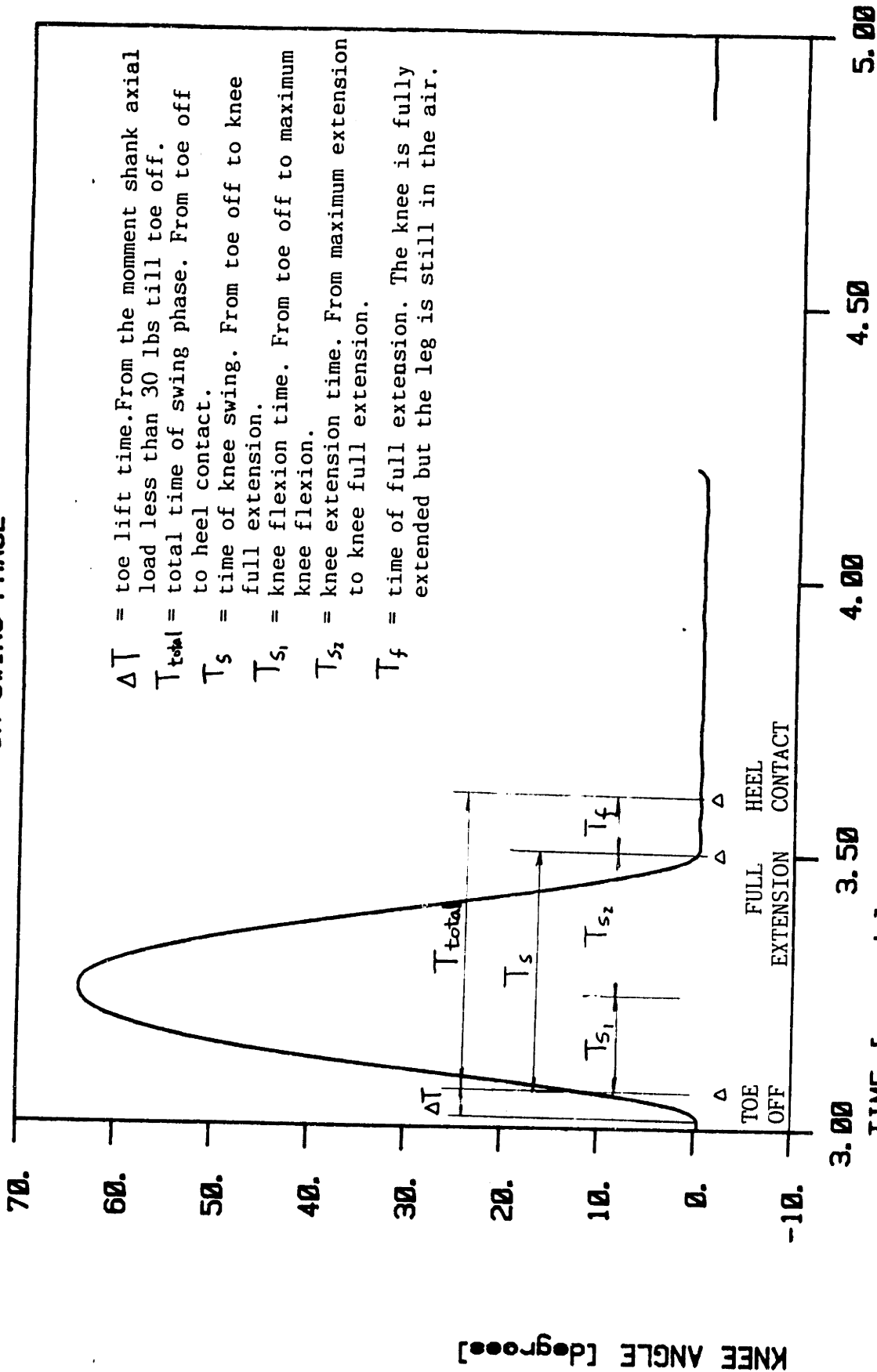


FIGURE V-1 Swing phase breakdown

KNEE ANGLE [degrees]

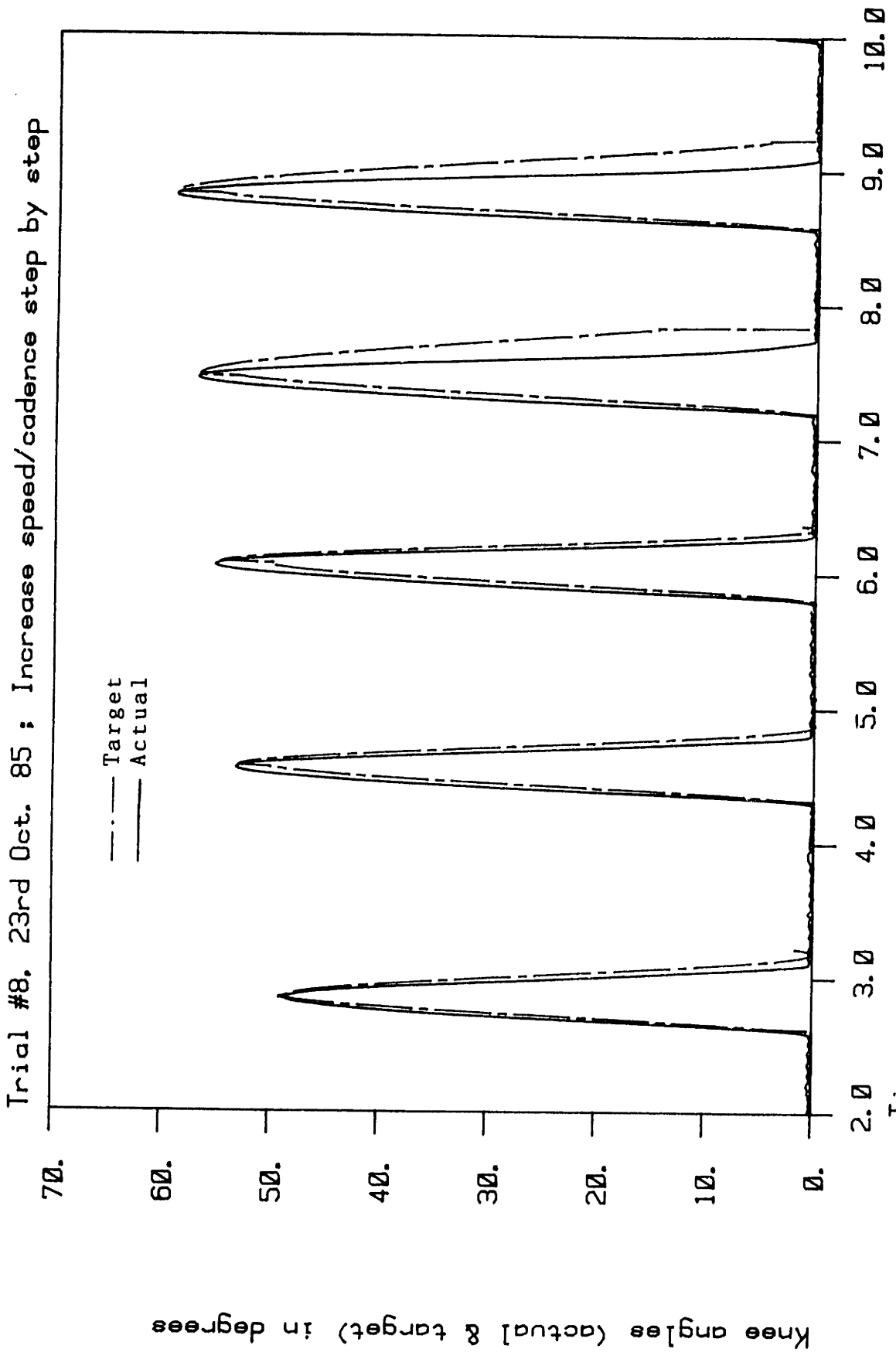


FIG V-2 Knee Angle (Target & Actual) under Trajectory Control Scheme with the predictor function of the swing kinematics being updated after each step

Knee angles (actual & target) in degrees

Surprisingly, Subject A did not seem to be bothered by the discrepancy between target and actual knee angle. He made positive comment about the routine, saying that controller was learning and optimizing itself. The subject also felt that speed adaptivity of the controller was satisfactory.

The iteration routine was then set to predict  $T_s$ , the shank swing time. The controller attempted to make the flexion and extension take place in the predicted time. The results of both subjects' showed the trajectory was followed well (Fig V-3 and Fig V-4). But the subjects consistently felt that the action of the knee was not smooth enough. Figure V-5 showed the typical brake current profile for one of such cycles. Notice the long time period at the beginning of the extension when the current is zero. In this region, the inertial properties of the prosthesis is such that the prosthesis simply can not be accelerated fast enough by the gravity alone to follow the target trajectory. Active power at the knee is needed. Remember that the passive prosthesis was used and could not produce the needed power to speed up the shank. The best the controller can do is to set the knee friction torque at minimum. This probably gave the person a feeling of lack of control. He did not feel that he was causing the movement of the shank. The brake torque profile also fluctuated as the tracking error changed its sign, which made the subjects feel the action of the knee was "rough". The iteration did not

Subject A: Oct 21, 1985;

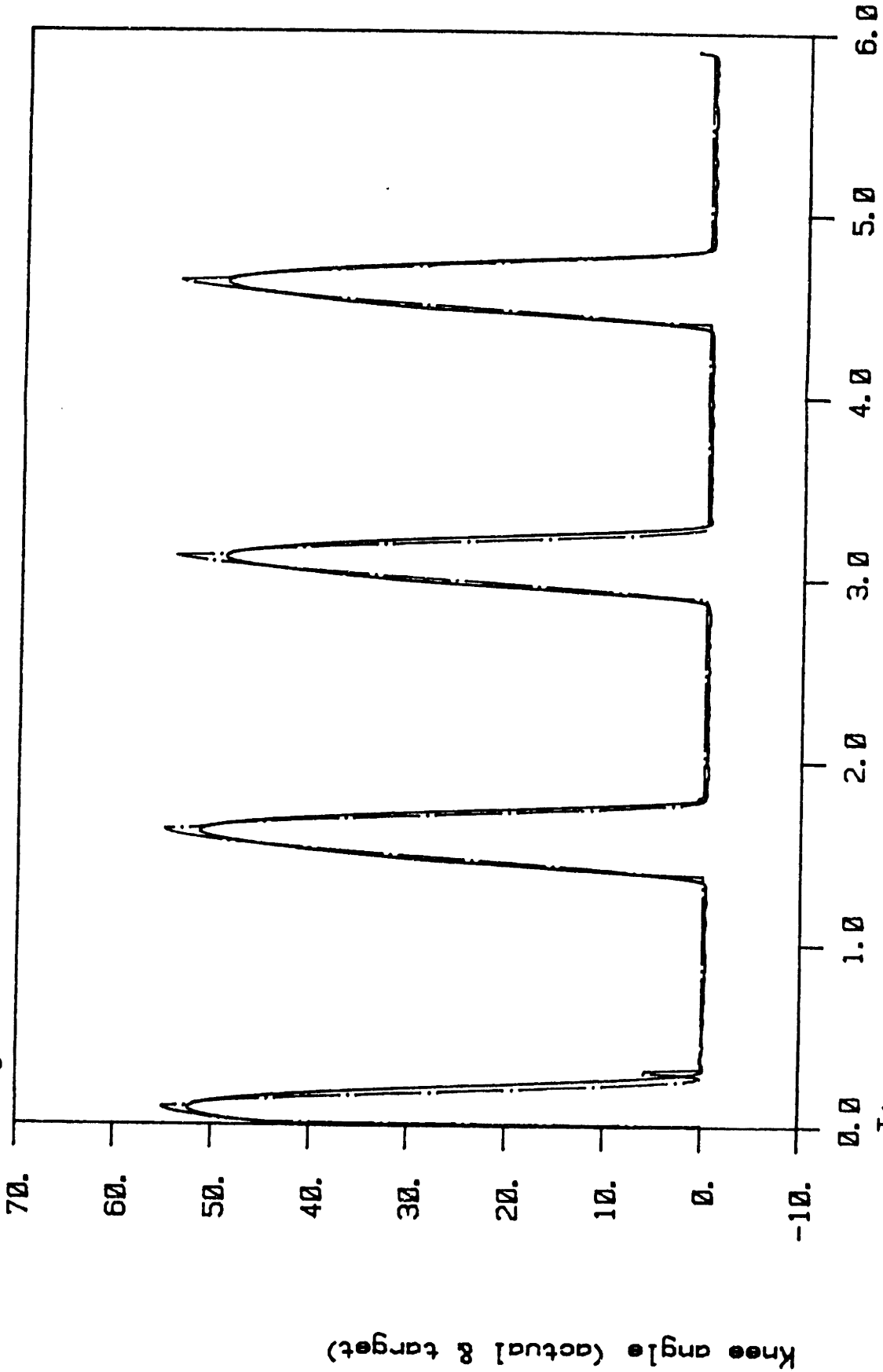


FIG. V-3 Knee Angle (actual & target) with Trajectory Control Scheme  
NO UPDATING OF THE PREDICTOR FUNCTION



Subject B. Dec. 8, 1985 ;

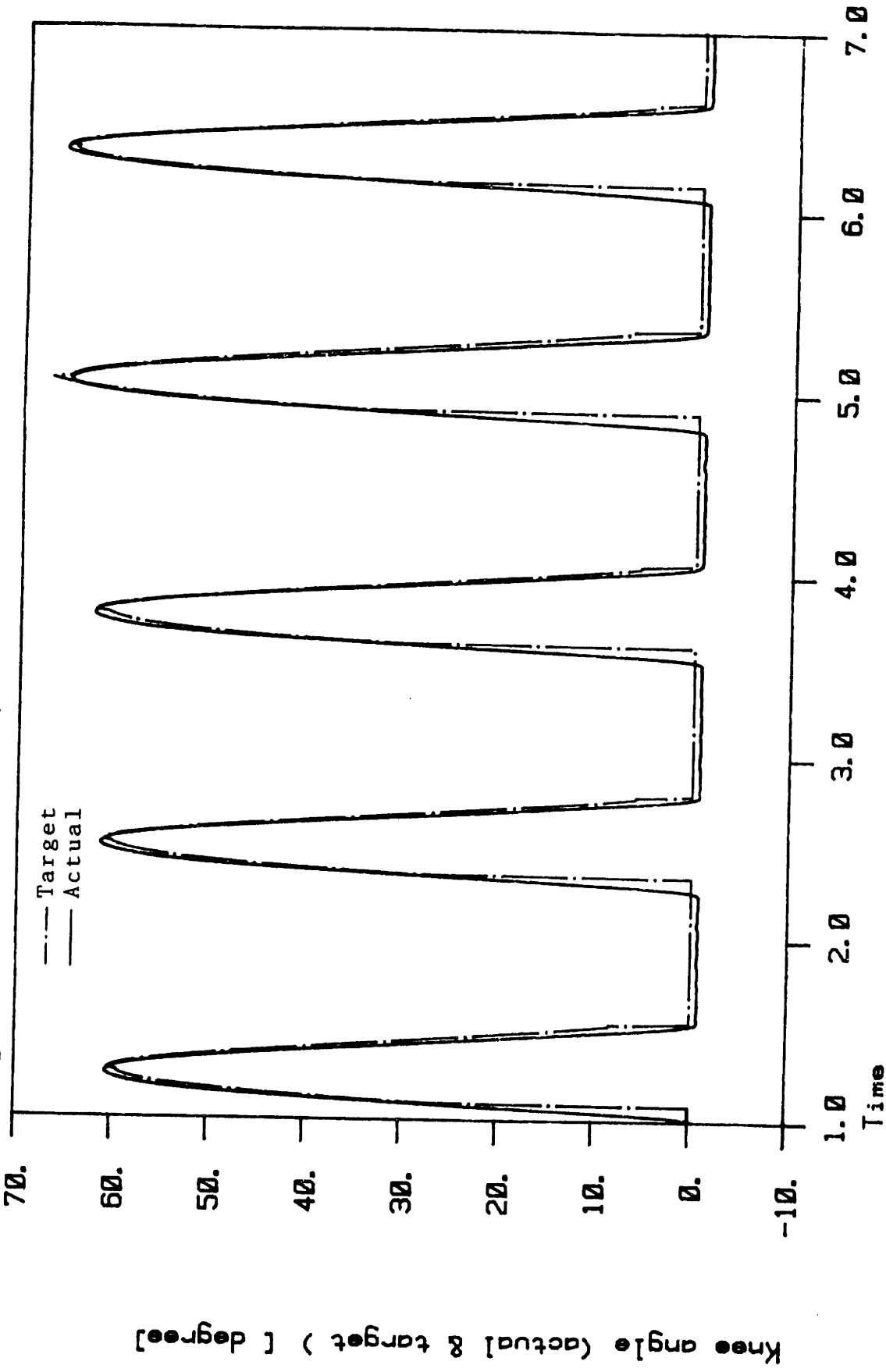


FIGURE V-4 Knee angle (actual & target) with Trajectory Control Scheme.  
NO UPDATING OF THE PREDICTOR FUNCTION

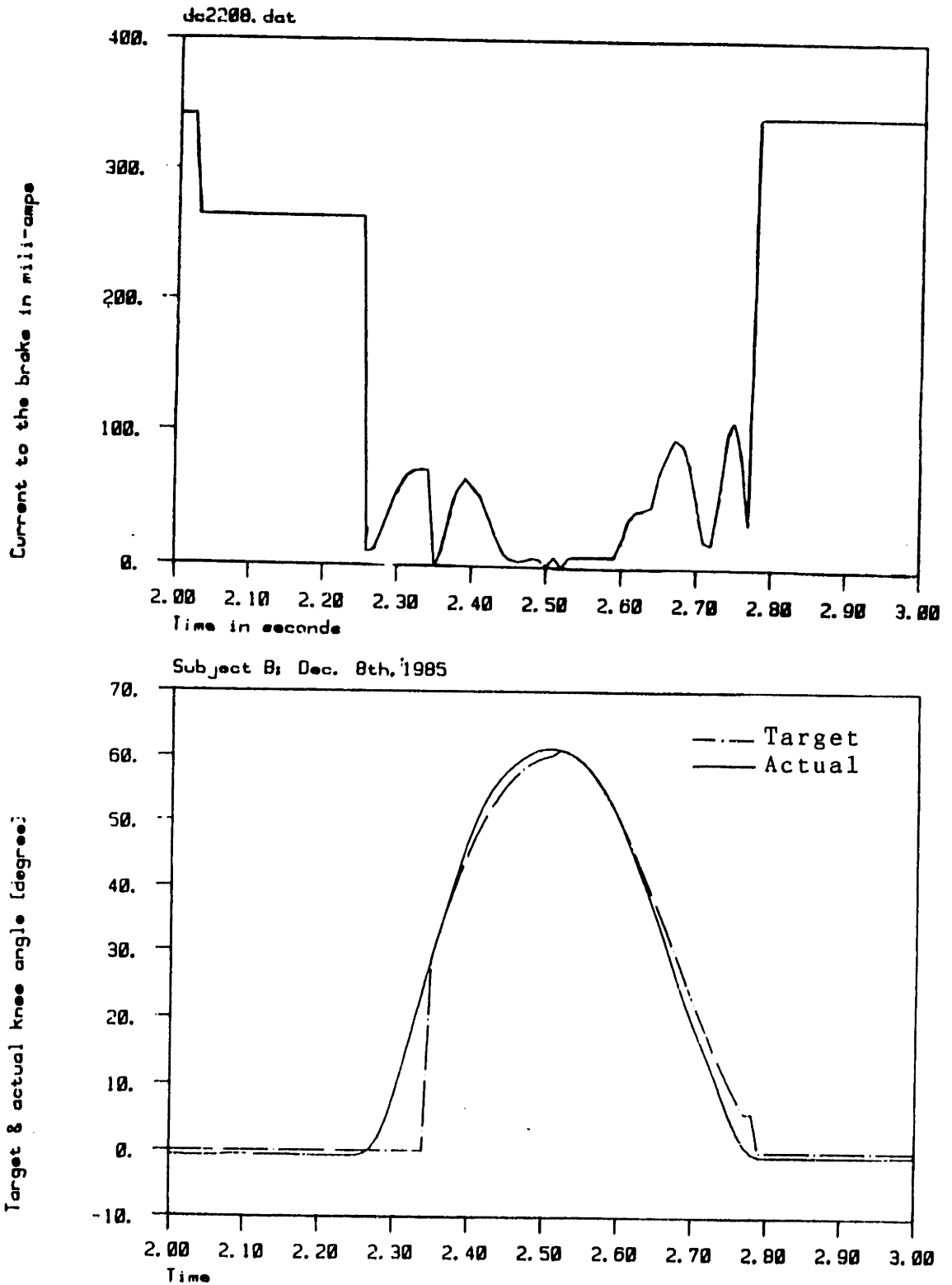


Fig.V-5 Brake current produced under Trajectory Control Scheme. Note its "roughness". But target trajectory was followed well.

converge to an optimal either.

## 2. Torque Control Scheme

This scheme was actually developed after the trajectory control scheme did not perform consistently. The torque control is to improve the smoothness of the shank motion. Both subjects commented positively on the scheme, especially for its smoothness. The gait in general looked good. Subject A who participated the experiment longer and had more experience with the scheme also commented that walking with this scheme for different speed required less effort than his regular prosthesis. He was able to comfortably vary his walking speed from 0.6 m/s to 1.4 m/s.

On the two proposed methods of scaling the damping coefficient  $B$ , the form  $B = \frac{t_e^2}{(\Delta t + t_e)^2} B_0$  was preferred. This scheme worked well and no attempts were made to further scale  $B$  by knee angle.

Figure V-6 through figure V-13 presents the knee angle, knee velocity, mode values, brake current and shank axial load for two different walking speeds by two subjects. Note that the brake current did not has the fluctuations that figure V-5 had.

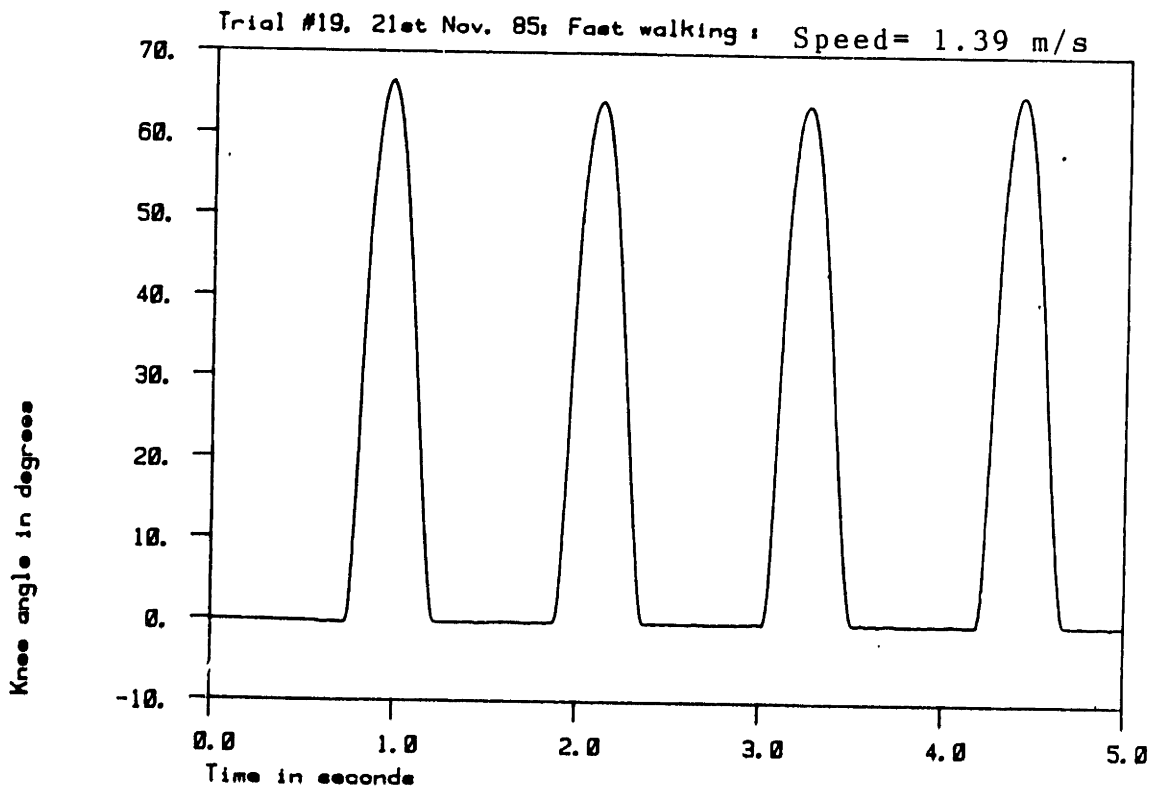
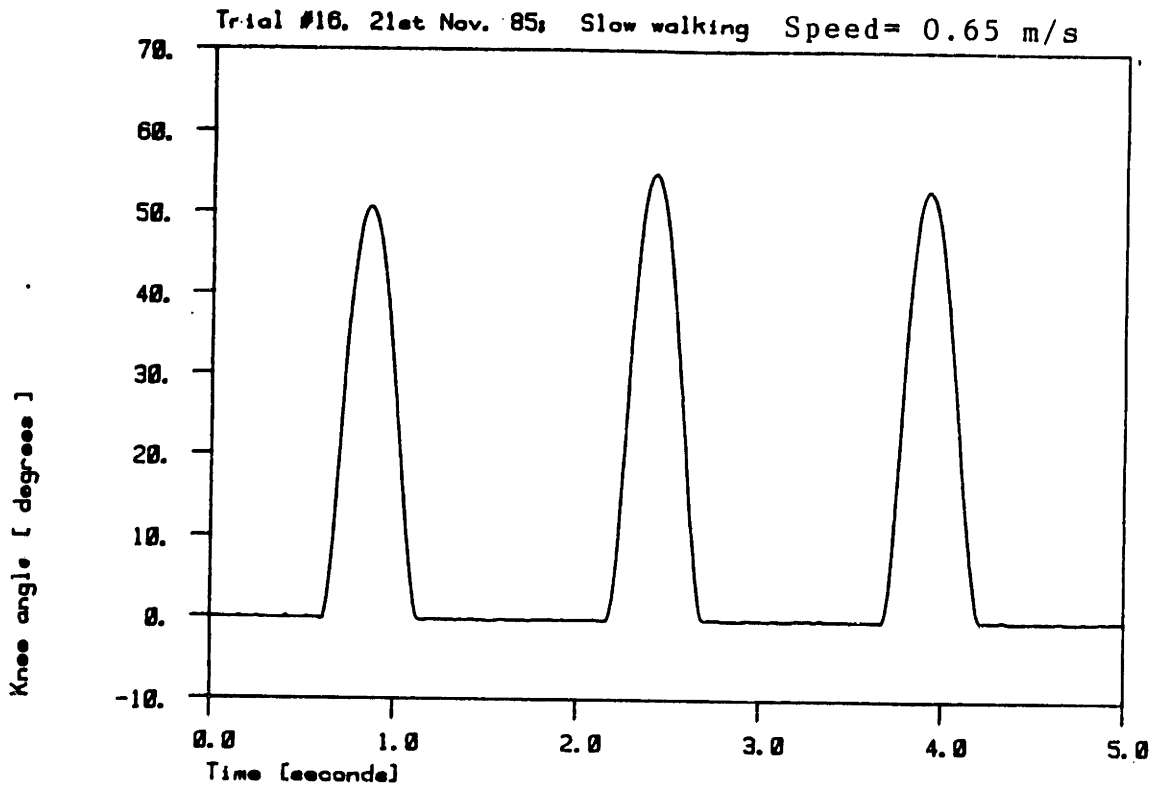


Fig V-6 Comparison of knee angle at two different walking speed. (Torque Control Scheme)

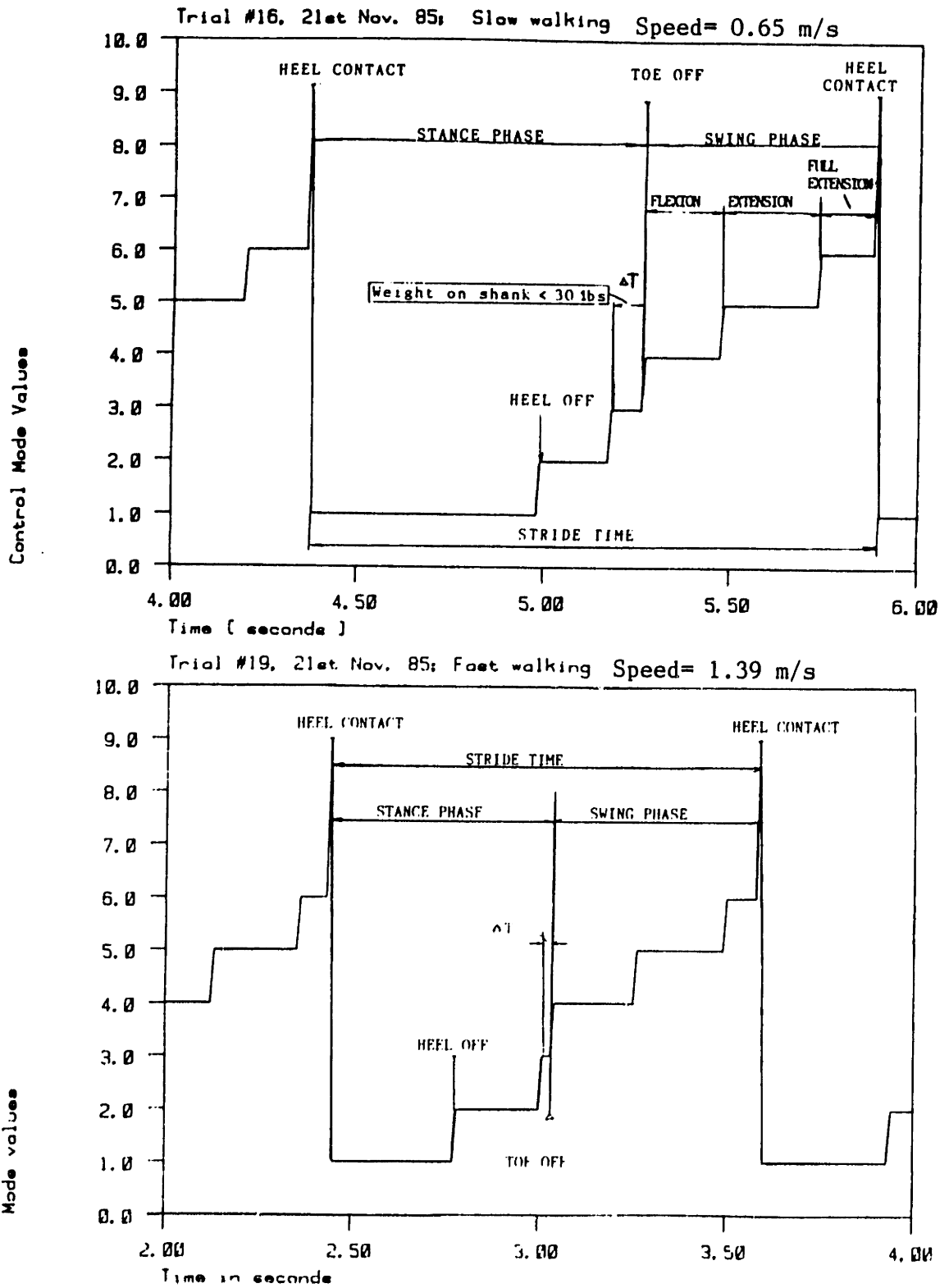


FIGURE V-7 Comparison of corresponding periods of two walking cycle

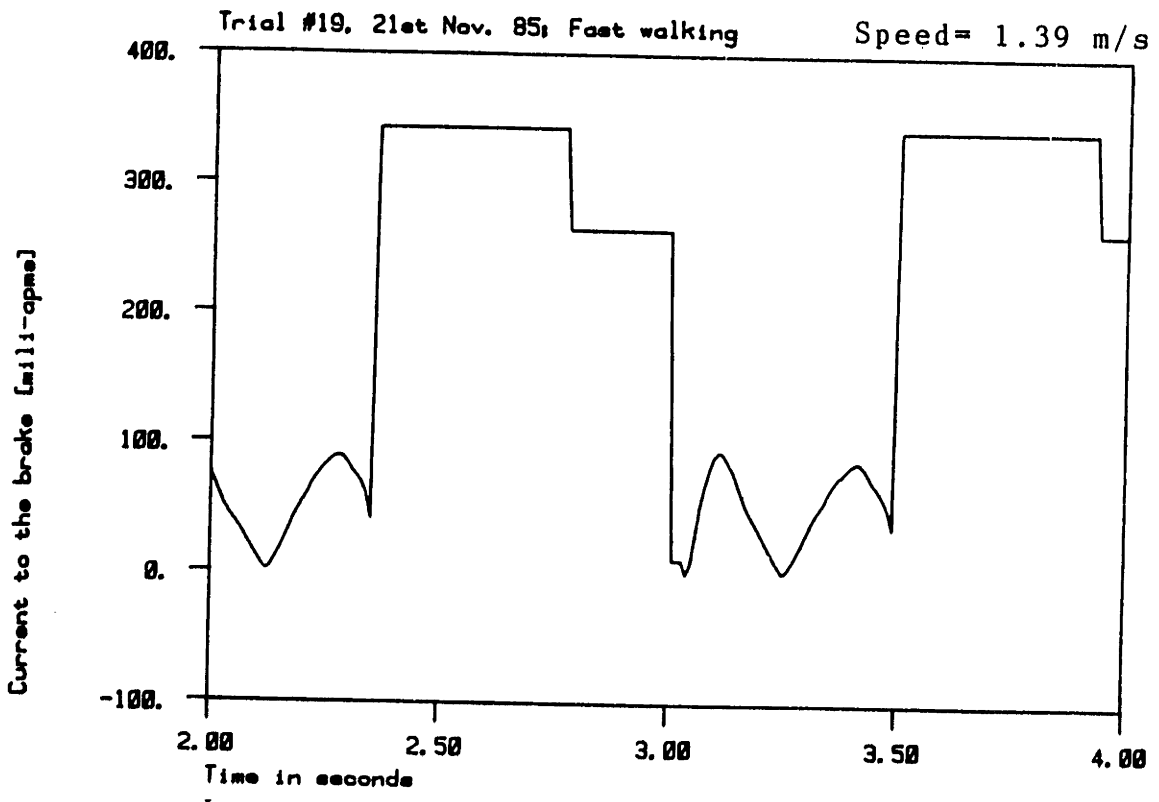
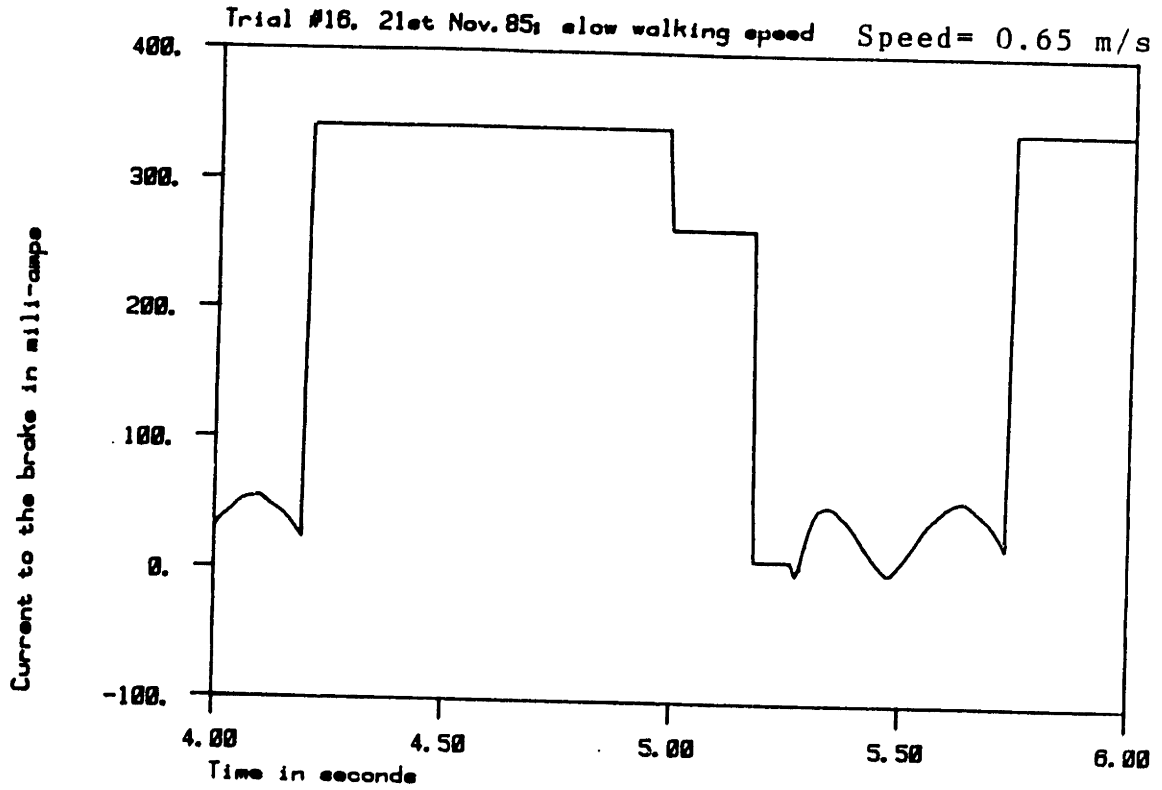


FIG V-8 Comparison of brake current at two different walking speed (TORQUE Control)

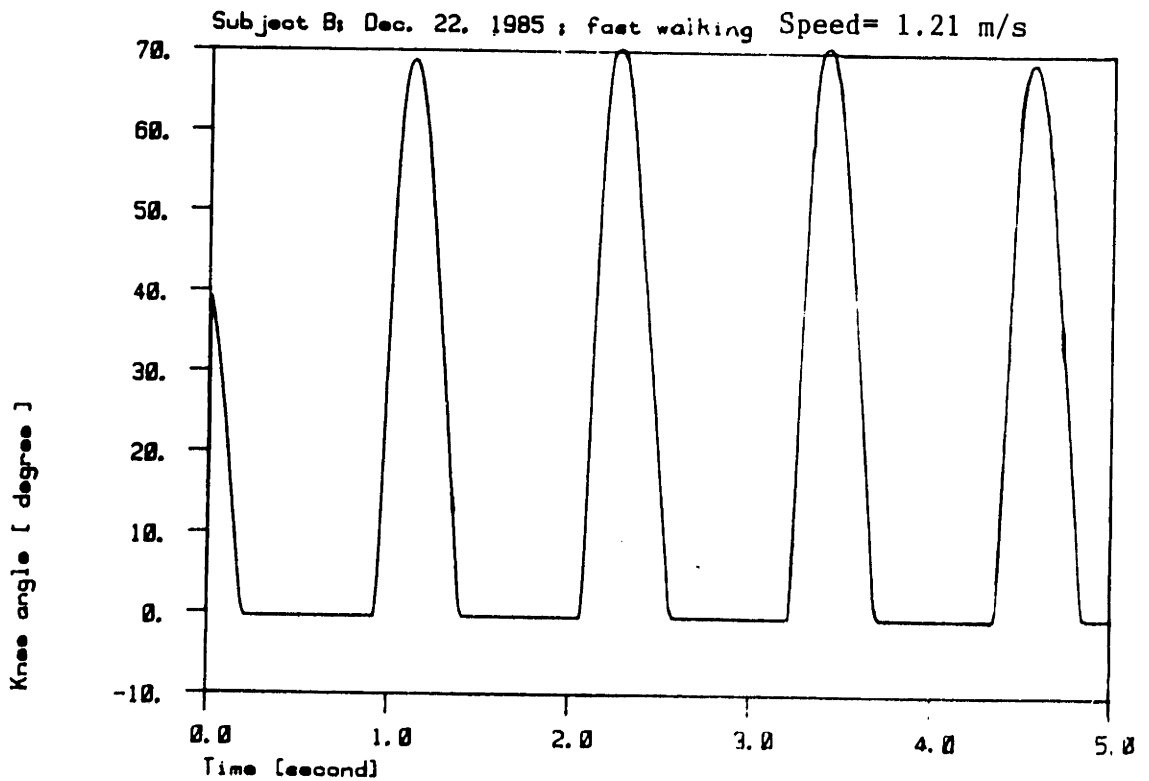
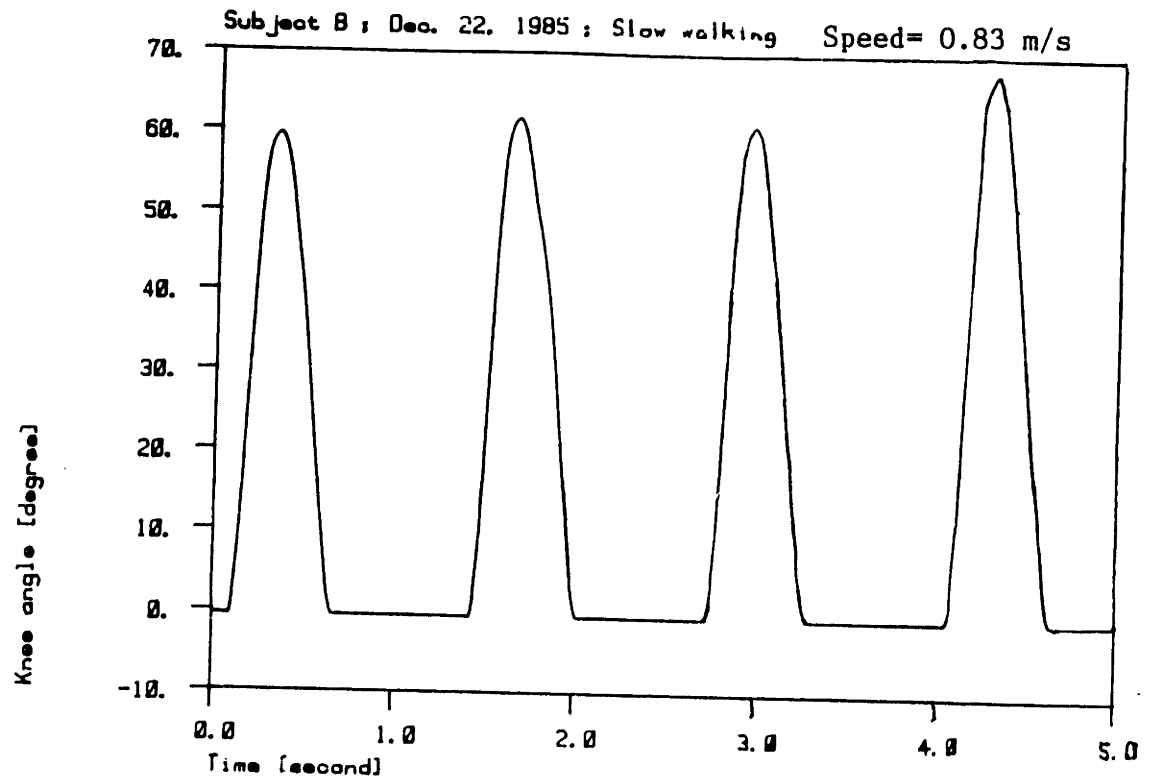


FIGURE V-9 Comparison of knee angles at two different walking speeds. (Torque Control Scheme)

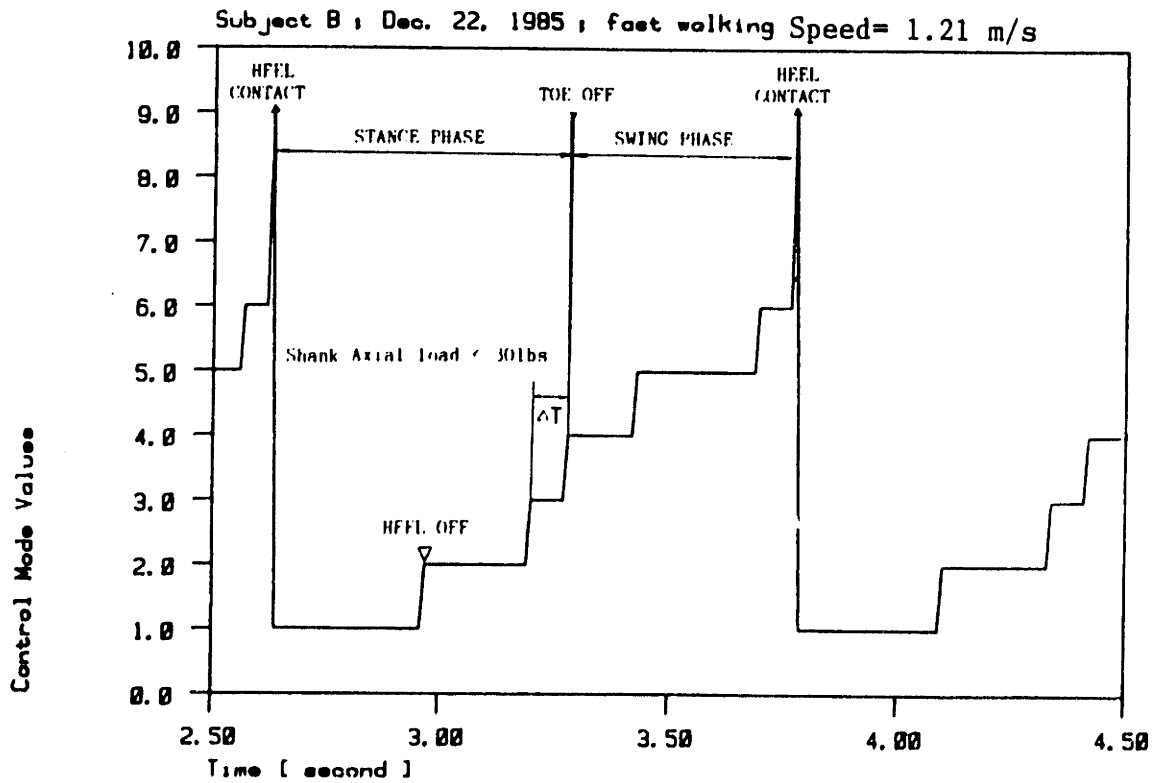
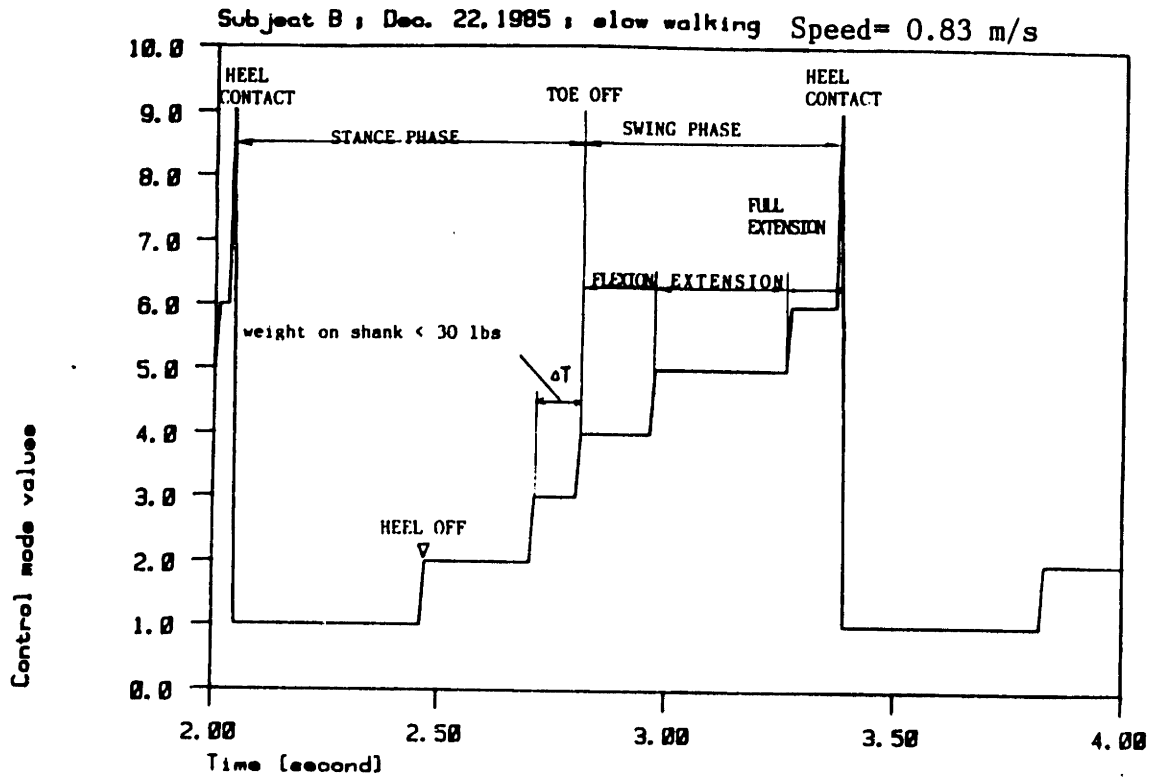


FIGURE V-10 Comparison of corresponding periods of two walking cycle



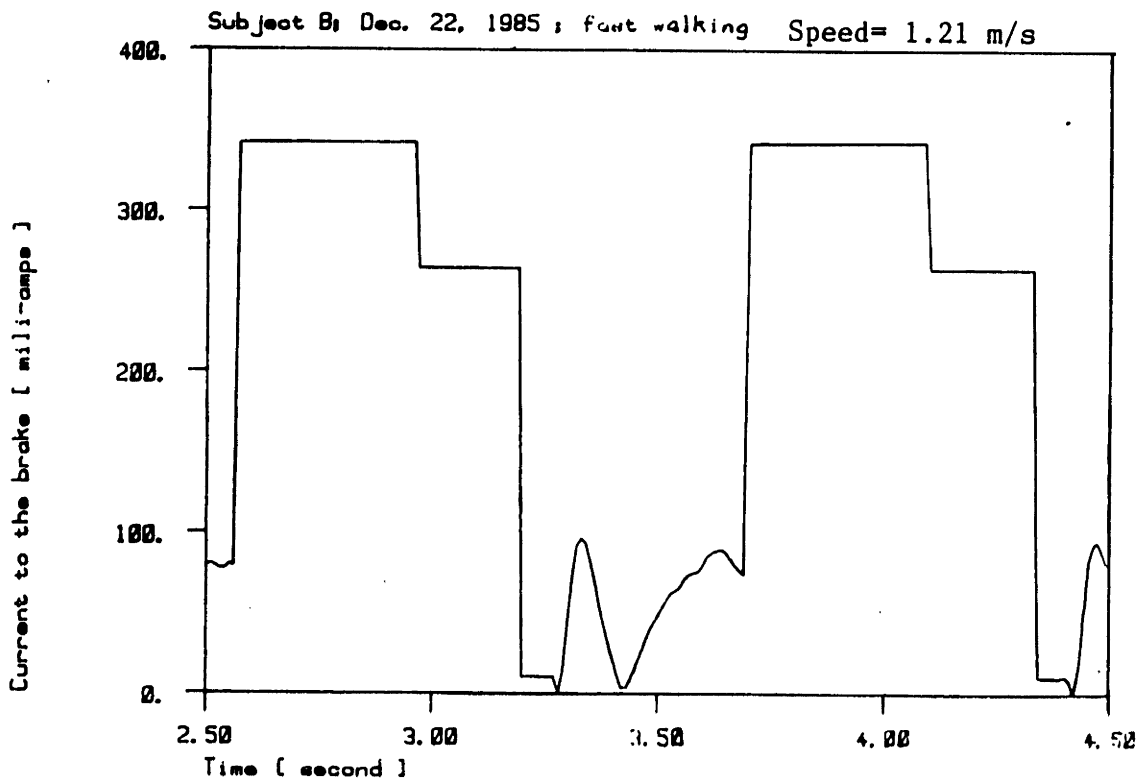
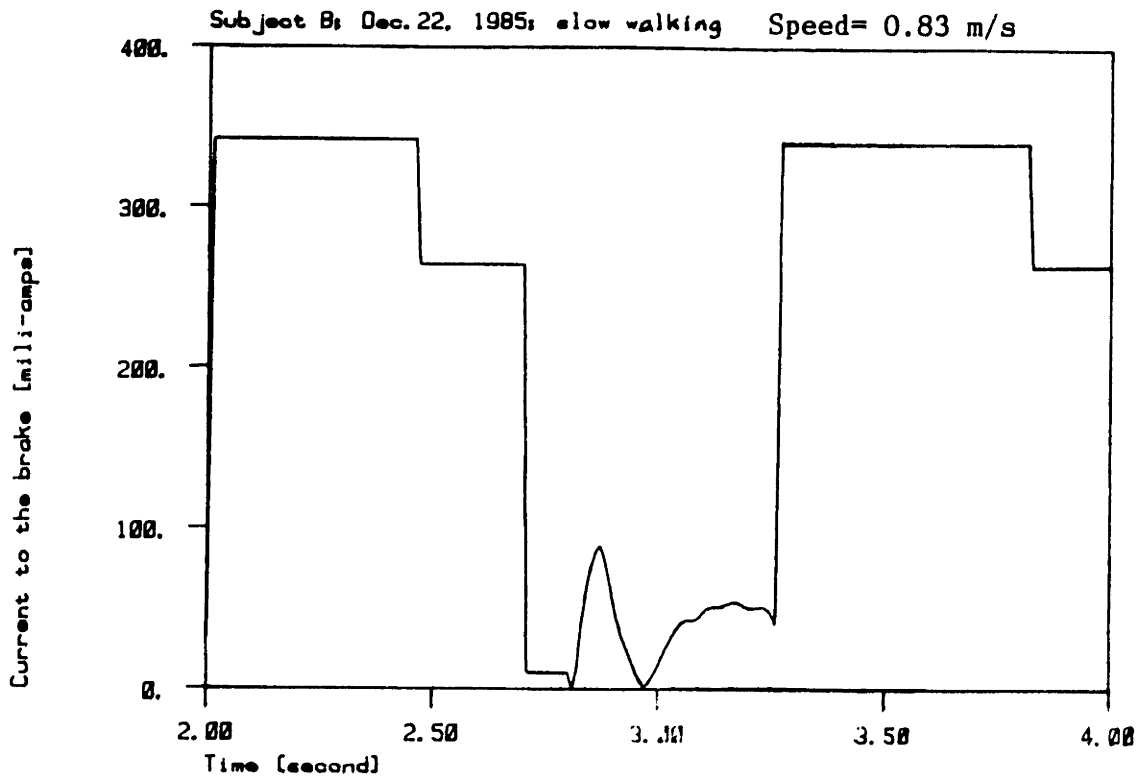


FIGURE V-11 Comparison of brake currents at two different walking speeds. (Torque Control)

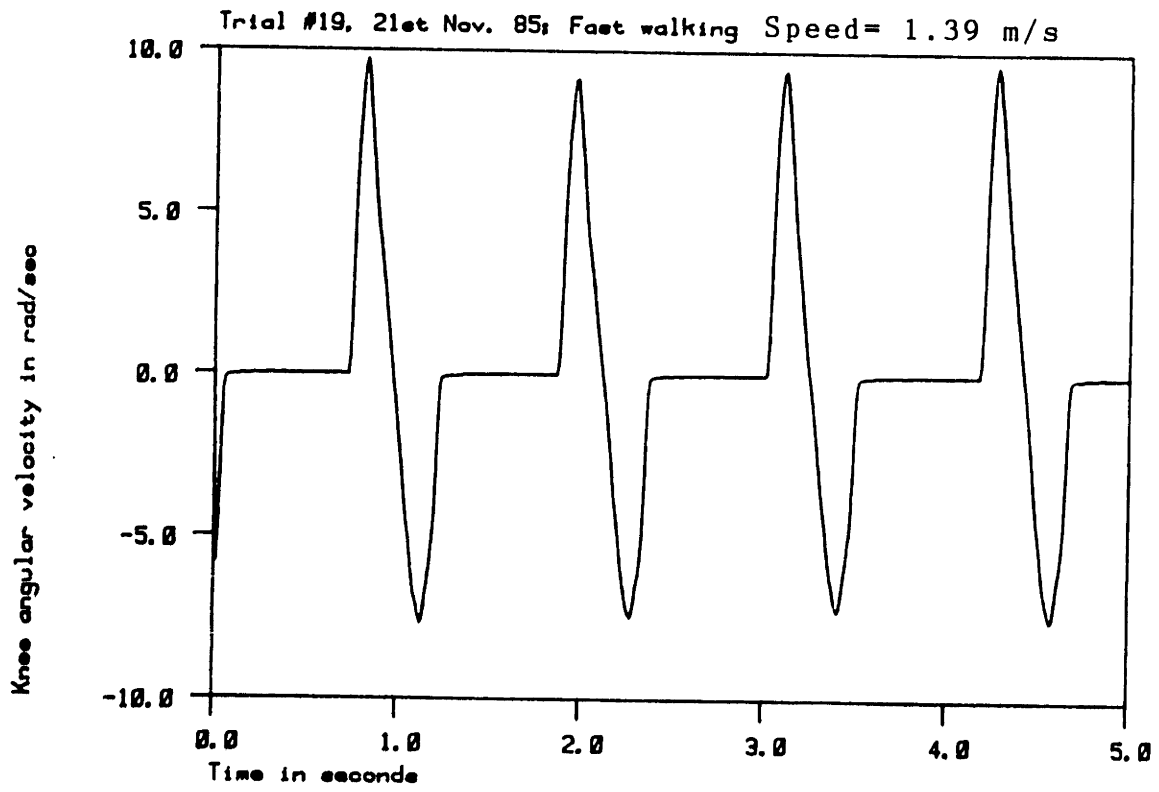
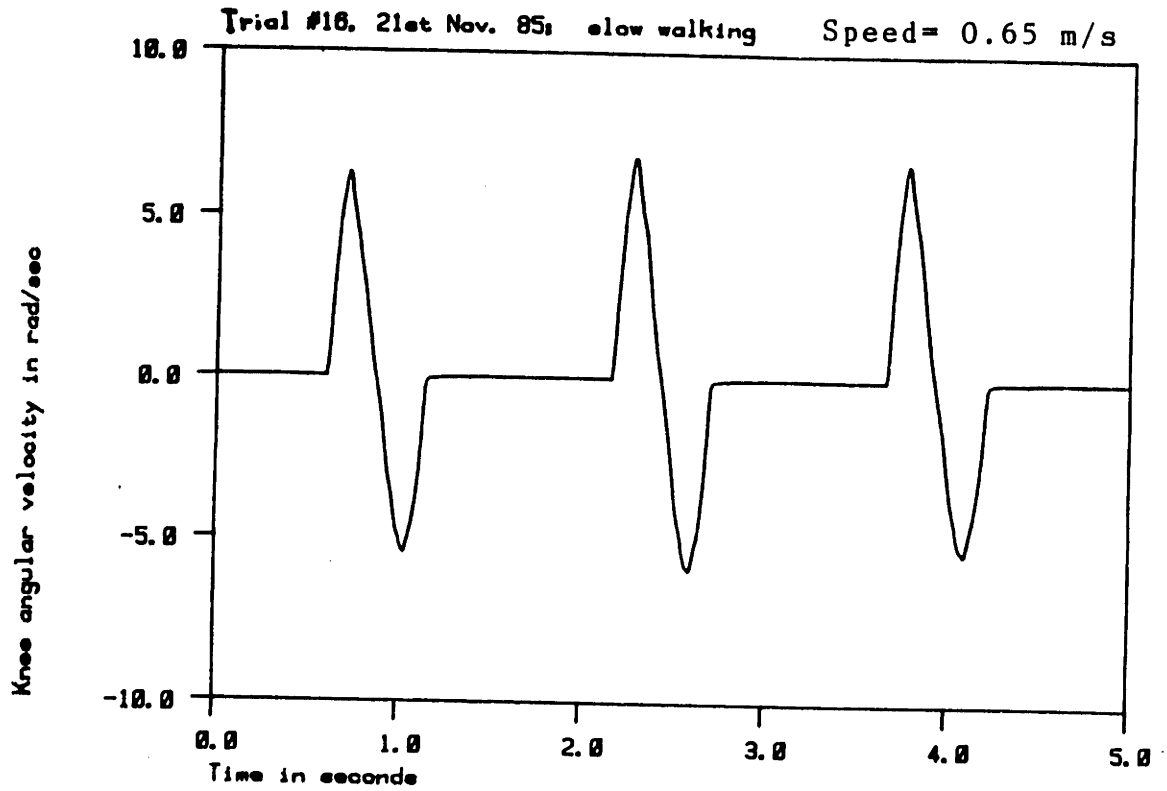


Fig.V-12 Comparison of knee angular velocity at two different speeds. (Torque Control Scheme)

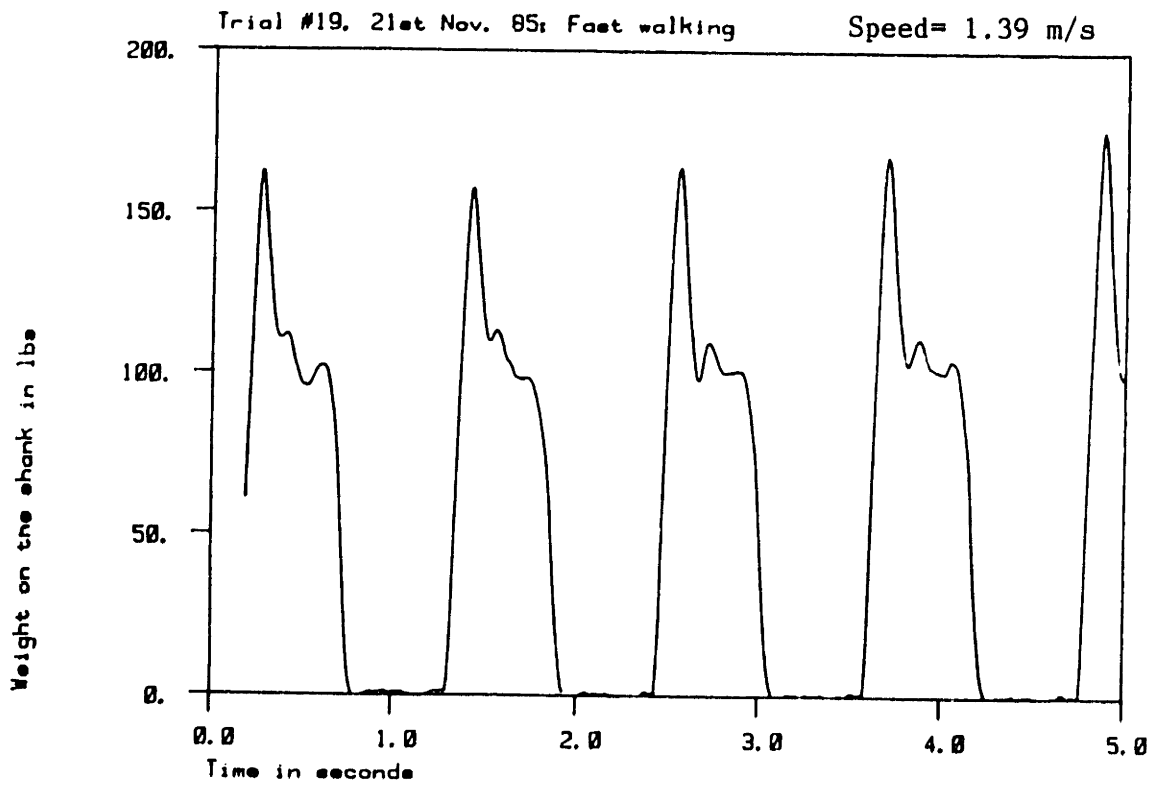
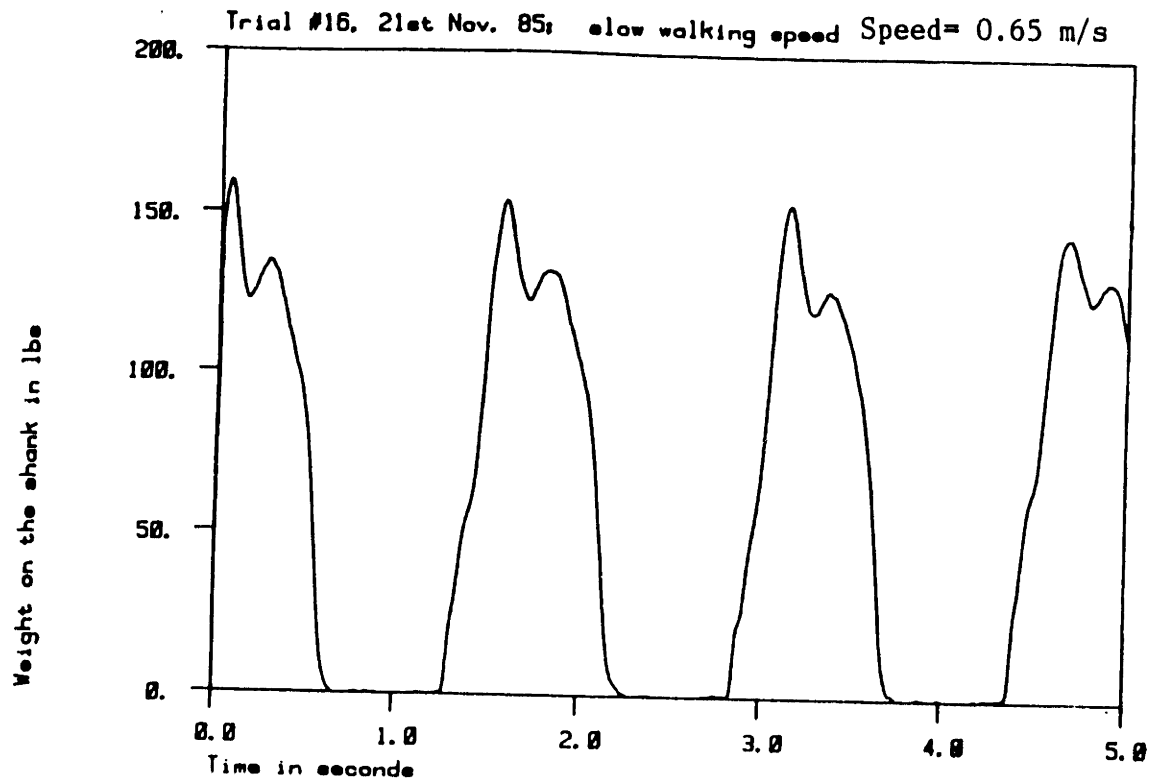


FIGURE V-13 Shank axial load

## DISCUSSION

The inconsistency of the trajectory control scheme could be a result of the unfavorable inertial properties of the prosthesis for the selected trajectory. In the region where the shank could not be sped up by gravity alone to follow the trajectory, knee torque was set to be zero. The amputee seems to be bothered by the fact that the knee joint was free of friction for a relatively long period.

It is not clear how we can select a trajectory to prevent the free-of-friction period from happening. A fourth order spline was tested and encountered more problem. The difference between the user's goal and controller target could cause system erratic behavior. The amputee varies his input to the system and "jerky" motion resulted.

The simple way used in this thesis to smooth the behavior is to limit the current to the particle brake. Doing this made it clear that the user preferred smooth control to accurate trajectory tracking.

## CONCLUSIONS

A method of predicting the intended walking speed with the signal from the sensors of the prosthesis itself was established. A control scheme that is self-contained and speed adaptive was subsequently developed. Tests by the

amputee subjects proved it is very beneficial to adjust the damping profile by the intended walking speed. Comparison of the reaction from the amputee subjects suggested that the smoothness of the action weighted heavier than the duplicating exactly the kinematics of the normal knee. The inertial properties of the leg plays an important role in the pattern of the gait kinematics. The selection of the target trajectory for the trajectory controller should be aimed toward requiring the least power being generated at the knee jointed. The trajectory control scheme should not be pursued when the inertial properties of the leg are unfavorable.

#### RECOMMENDATIONS

The control sheme should be further developed as to accomodate activities such as walking down the ramp or the stairs , sitting and standing up. The microcomputer implementation of this control sheme should be made so that the self-contained prosthesis can be finalv worn daily by the amputee.

APPENDIX  
LIST OF  
COMPUTER PROGRAM

```

C*****
C      OPEN-LOOP SPEED ADAPTIVE LEVEL WALKING CONTROL ALGORITHM
C      CHN IVAL  NOTATION  NOMENCLATURE
C      0      1      IFTSW      foot switch
C      1      2      ILOAD      loadcell
C      2      3      ISTOP      stop button by the subject
C      3      4      IVV        knee velocity
C      4      5      IPP        knee angle
C      5      6      IPH        photo cell signal
C      6      7
C      FUNCTIONAL MODE
C      1      heel contact -----> heel off
C      2      heel off -----> weight > 30 LBS
C      3      weight<30 LB -----> toe off
C      4      toe off -----> max. knee flexion
C      5      max. k. flex. -----> knee full extension
C      6      full ext. -----> hell contact
C      SUPPLEMENTARY MODE
C      7      DEFINITION
C      8      when foot leaves ground from heel
C      9      intermediate mode between MODE 5 and MODE 1
C      10     intermediate mode between MODE 6 and MODE 1
C      11     1 degree = 44.1913 units      1 unit=0.02263 degree
C      12     1 deg/sec = 3.61486 units      1 unit=0.27664 deg/sec
C*****
C      VIRTUAL JPOS(1001),JVEL(1001),JSIG(1001),JFTSW(1001)
C      VIRTUAL JLOAD(1001),JMODE(1001),JPD(1001),JVD(1001),JTIM(1001)
C      DIMENSION IOF(50)
C      COMMON /DATA/IVAL(7)
C      COMMON /FLAG/IDONE,IQUIT,IREC
C      COMMON IOFF(3),IZ,Z,ISW1,ISW2,ILCK1,ILCK2,LDIV,U(S),
C      * PMKE(200),ITM(200),ITN(200),ITS,IT4,ITS,IT6,JTS,JT(9,200),
C      * MODE,IP(3),IV(3),IL(3),IPP,IVV,IFTSW,ISIG,ILOAD,ISTOP,IPH,
C      * PD,VD,SIG,SIGP,SIGV,IPMAX
C      * A1,B1,C1,D1,A2,B2,C2,D2,GAL1,GAL2,CP1,CP2,GV1,GV2,IN,IM
C*****
C      Initialization *****
C      DATA ILCK1/3800/,ILCK2/3400/,ISW1/2800/,ISW2/2100/,JT/1800*0/
C      DATA IZ/2048/,Z/2047*0/,JTS,IT3,IT4,ITS,IT6/0.0,0.0,0.0/
C      DATA NCHAN,IEIRST/7,1/,IOFF/60,2015,4065/,NPNTS/1000/
C      DATA TLDIV/30.0/,DIST/7.27/,TM/0.25/,TN/0.4/
C      DATA U/0.0226288,0.2766393,0.087193,1.4,1.4/
C
C      TYPE,' WHAT IS THE SAMPLING FREQUENCY ?'
C      TYPE,' 1 = 100 Hz ; 2 = 200 Hz'
C      ACCEPT*,ISAM
C      DO 2 I=1,ISAM*25
C      ITM(I)=ISAM*25
C      PMKE(I)=55.0
C      ITN(I)=ISAM*20+I
C      DO 4 I=ISAM*25+1,200
C      ITM(I)=ISAM*25
C      PMKE(I)=55.0
C      ITN(I)=ISAM*45
C      set initial parameters
C      GAL=1.0
C*****
C      CP1=1.0
C      GV1=10.0
C      GA2=2.0
C      CP2=5.0
C      GV2=0.5
C      C set d/a register
C      CALL CLRDAC
C      C set brake current to zero
C      CALL IDAC(IZ)
C ***** start offset measurement *****
C      10     TYPE 912
C      912     FORMAT(' DO YOU WANT TO MEASURE THE OFFSET ? ',6)
C      ACCEPT 910,ANSW
C      FORMAT(A1)
C      IF (ANSW.NE.'Y'.AND.ANSW.NE.'Y')GOTO 15
C
C      PAUSE ' SET LEG TO ZERO ANGLE THEN <CR>'
C      IOFF(1)=IADC(6)
C      IOFF(1)=IOFF(1)+20
C      IOFF(2)=IADC(4)
C      TYPE 914, IOFF(1),IOFF(2)
C      914     FORMAT(1X,'KNEE ANGLE OFFSET = ',I5,/,
C      * 1X,'KNEE VELOCITY OFFSET = ',I5,/)
C      PAUSE ' PLEASE LIFT THE PROSTHESIS FROM GROUND <CR>'
C      DO 12 I=1,50
C      IOF(I)=IADC(2)
C      IFF=0
C      DO 13 I=1,50
C      IFF=IFF+4065-IOF(I)
C      IOFF(3)=4065-IFF/50
C      C lock knee to measure load cell offset
C      PAUSE ' HIT <CR> TO LOCK THE KNEE'
C      CALL IDAC(ILCK1)
C      PAUSE ' PLEASE PUT ALL WEIGHT ON PROSTHESIS <CR>'
C      LOAD=IADC(2)
C      FLOAD=FLOAT(IOFF(3)-LOAD)*U(3)
C      TYPE 916, IOFF(3),FLOAD
C      916     FORMAT(1X,' LOAD CELL OFFSET = ',I5,/,
C      * 1X,' TOTAL BODY WEIGHT = ',F10.5,2X,' lbs')
C      TYPE,' MEASURE OFFSET AGAIN ?'
C      ACCEPT 910,ANSW
C      IF (ANSW.NE.'Y'.AND.ANSW.NE.'Y') GOTO 15
C      PAUSE ' HIT <CR> TO UNLOCK THE KNEE. CAREFUL!!!'
C      CALL IDAC(IZ)
C      GOTO 10
C*****
C      parameter adjustment *****
C      15     TYPE 920,GAL,GP1
C      TYPE*,GV1
C      TYPE,' DO YOU WANT TO CHANGE THE FLEXION GAIN ?'
C      ACCEPT 910,ANSW
C      IF (ANSW.NE.'Y'.AND.ANSW.NE.'Y') GOTO 20
C      TYPE,' ENTER THE NEW GAINS : GAL,GP1'
C      ACCEPT*,GAL,GP1,GV1
C      20     TYPE 920,GA2,GP2
C      TYPE,' DO YOU WANT TO CHANGE THE EXTENSION GAIN ?'
C      ACCEPT 910,ANSW
C      IF (ANSW.NE.'Y'.AND.ANSW.NE.'Y') GOTO 30
C      FORMAT(1X,'GAIN= ',G12.6,/1X,' GP = ',G12.6)
C      TYPE,' ENTER THE NEW EXTENSION GAINS: GA2,GP2'
C      ACCEPT*,GA2,GP2

```

```

C CC25 TYPE*, ' THE LIMITS FOR THE BRAKE CURRENT ARE : '
C TYPE*, U(4), U(5)
C TYPE*, ' DO YOU WANT TO CHANGE THEM ? '
C ACCEPT 910, ANSW
C IF (ANSW.NE., 'Y'.AND.ANSW.NE., 'Y') GOTO 30
C TYPE*, ' ENTER THE NEW LIMITS FOR THE CURRENT : '
C ACCEPT*, U(4), U(5)
C
30 TYPE 924, ISW2
924 FORMAT(1X, 'ISW2= ', I5/
      1X, ' DO YOU WANT TO CHANGE IT ? ', $)
      ACCEPT 910, ANSW
      IF (ANSW.NE., 'Y'.AND.ANSW.NE., 'Y') GOTO 32
      TYPE*, ' ENTER NEW ISW2 : '
      ACCEPT*, SW2
      ISW2=IFIX(SW2)
32 TYPE*, ' DO YOU WANT TO CHANGE THE DISTANCE IN BETWEEN : '
      ACCEPT 910, ANSW
      IF (ANSW.NE., 'Y'.AND.ANSW.NE., 'Y') GOTO 35
      TYPE*, ' ENTER THE DISTANCE : '
      ACCEPT*, DIST
      TYPE 930, TLDLV
930 FORMAT(1X, 'THRESHOLD LOAD LEVEL IS : ', F10.6/
      1X, ' DO YOU WANT TO CHANGE IT ? ', $)
      ACCEPT 910, ANSW
      IF (ANSW.NE., 'Y'.AND.ANSW.NE., 'Y') GOTO 40
      TYPE*, ' ENTER THE LOAD THRESHOLD : '
      ACCEPT*, TLDLV
      LDLV=IFIX(TLDLV/U(3))
      IRATE=4
      ITICK=10
      TYPE 940
940 FORMAT(1X, ' THE CURRENT CLOCK RATE IS 100 Hz ; CHANGE IT ? ', $)
      ACCEPT 910, ANSW
      IF (ANSW.NE., 'Y'.AND.ANSW.NE., 'Y') GOTO 45
      TYPE*, ' ENTER THE CLOCK RATE AND NUMBER OF TICKS : '
      ACCEPT*, RATE, TICK
      IRATE=IFIX(RATE)
      ITICK=IFIX(TICK)
45 PAUSE '<CR> TO ENTER CONTROL LOOP'
C set initial control parameter
      IPF=0
      IPW=0
      ILOAD=0
      IL(1)=0
      IP(1)=0
      IV(1)=0
      ISTOP=2000
      IT3=0
      IT4=0
      IT5=0
      IT6=0
      PD=0.0
      VD=0.0
      K=0
      IDONE=1
      IQUIT=0
      IREC=0
      CALL SETUP (IRATE, ITICK, NCHAN, IFIRST)
      CONTINUE
      IF (IDONE.EQ.1) GOTO 100
      IF ((IQUIT.EQ.1).OR.(ISTOP.LT.1800)) GOTO 140
      CALL MODTCR
      CALL CONTR0
C collect data
      IF ((IREC.LT.1).OR.(K.GT.1000)) GOTO 130
      K=K+1
      JPOS(K)=IPP
      JVEL(K)=IYW
      JLOAD(K)=ILOAD
      JFTSW(K)=IFTSW
      JMODE(K)=MODE
      JSIG(K)=ISIG
      JTIM(K)=IPH
      IF (PD.GT.100.0) GOTO 130
      IF ((VD.GT.800.).OR.(VD.LT.-800.0)) GOTO 130
      JPD(K)=IFIX(PD/U(1))
      JVD(K)=IFIX(VD/U(2))
      JPD(K)=0
      JVD(K)=0
      IDONE=1
      GOTO 100
C ***** control loop ends *****
C set appropriate brake current
140 IF (ISTOP.LT.1800) GOTO 142
      CALL IDAC(ISW2)
      GOTO 145
142 CALL SETR(-1, .)
      CALL IDAC(ILCKI)
C ***** output data *****
145 JOLD=2200
      JNEW=JOLD
      IDT=0
      IREC=0
      DO 150 I=1, K
      JOLD=JNEW
      JNEW=JTIM(I)
      IF ((JOLD-JNEW).LT.1500) GOTO 150
      IDT=I-IDT
      IREC=IREC+1
      CONTINUE
150
C
      IF (IREC.LT.2) GOTO 154
      DT=FLOAT(IDT)*0.01
      SPEED=DIST/DT
      TYPE 949, DT, SPEED
      FORMAT(1X, ' THE TIME YOU TOOK WAS : ', F10.4./
      ' THE AVERAGE SPEED WAS : ', F10.4)
      GOTO 156
154 TYPE*, ' THE SPEED INFORMATION WAS NOT RECORDED ! '
156 TYPE 948, JTS
948 FORMAT(' THE VALUE OF JTS IS : ', I5)
      TYPE 950
950 FORMAT(' CHECK THE PREDICTION FUNCTION ? ', $)
      ACCEPT 910, ANSW
      IF (ANSW.NE., 'Y'.AND.ANSW.NE., 'Y') GOTO 160

```



```

955 TYPE 955
    FORMAT(' IT3 ',' IT4 ',' ITM ',' ITS ',' ITN ',' PMAX ')
    DO 158 J=1,JTS
    PMAX=FLOAT(JT(7,J))*U(1)
    TYPE 956,JT(1,J),JT(2,J),JT(3,J),JT(4,J),JT(5,J),JT(6,J),PMAX
    FORMAT(516,F10.4)
    CONTINUE
957 TYPE 957
    FORMAT(' DO YOU WANT TO RESET JTS ? ',#)
    ACCEPT 910,ANSW
    IF (ANSW.NE.'Y'.AND.ANSW.NE.'Y')GOTO 159
    JTS=0
159 TYPE*, DO YOU WANT TO CORRECT SOME BAD MEMORY ?
    TYPE*, 0 = NO ; 1 = CORRECT INDIVIDUAL ONE
    ACCEPT*,ICHO
    IF (ICHO.EQ.0)GOTO 160
    TYPE*, ENTER THE VALUE OF IT3
    ACCEPT*,IT3
    TYPE*, ENTER THE VALUE OF IT4,ITS :
    ACCEPT*,ITM(IT3),ITN(IT3)
    TYPE 958
    FORMAT(' HAVE A LOOK OF THE DATA TAKEN ? ',#)
    ACCEPT 910,ANSW
    IF (ANSW.NE.'Y'.AND.ANSW.NE.'Y')GOTO 162
    DO 161 J=1,K
    FORMAT(1016)
    WRITE (5,960)J,JPOS(J),JVEL(J),JSIG(J),JFTSW(J),JLOAD(J),JMODE(J),
    * JPD(J),JVD(J),JTIM(J)
162 TYPE*, DO YOU WANT TO SAVE THE DATA FILE ?
    ACCEPT 910,ANSW
    IF (ANSW.NE.'Y'.AND.ANSW.NE.'Y')GOTO 180
    TYPE*, ENTER THE OUTPUT FILE NAME:
    CALL ASSIGN(1,'',NEW)
    DEFINE FILE 1 (1001:10.U.I.REC)
    DO 170 J=1,NPTS
    WRITE (1'J)J,JPOS(J),JVEL(J),JSIG(J),JFTSW(J),JLOAD(J),JMODE(J),
    * JPD(J),JVD(J),JTIM(J)
170 CONTINUE
    CALL CLOSE(1)
    CALL CLOSE(2)
    TYPE*, DO YOU WANT TO TAKE ANOTHER SAMPLE ?
    ACCEPT 910,ANSW
    IF (ANSW.EQ.'Y'.OR.ANSW.EQ.'Y')GOTO 10
    STOP
C*****
    SUBROUTINE MODTGR
C
C This subroutine is to update all the variables and mode values.
C Also a moving average filter is used for sampled data.
C*****
    COMMON /DATA/IVAL(7)
    COMMON IOFF(3),IZ,Z,ISWL,ISWZ,ILCK1,ILCK2,LDLV,U(5).

```

```

    PMAX(200),ITM(200),ITN(200),ITS,IT4,ITS,IT6,JTS,JT(9,200),
    * MODE,IP(3),IV(3),IL(3),IPP,IVV,IFTSM,ISIG,ILOAD,ISTOP,IPH,
    * PD,VD,SIG,SIGP,SIGV,IPMAX,
    * A1,B1,C1,D1,A2,B2,C2,D2,GAL1,GAL2,GP1,GP2,GV1,GV2,TN,TM
    IP(1)=IP(2)
    IP(2)=IP(3)
    IP(3)=IVAL(6)-IOFF(1)
    IPP=(IP(1)+IP(2)+IP(3))/3
    IV(1)=IV(2)
    IV(2)=IV(3)
    IV(3)=IVAL(4)-IOFF(2)
    IW=(IV(1)+IV(2)+IV(3))/3
    IL(1)=IL(2)
    IL(2)=IL(3)
    IL(3)=IOFF(3)-IVAL(2)
    ILOAD=(IL(1)+IL(2)+IL(3))/3
    IFTSM=IVAL(1)
    ISTOP=IVAL(3)
    IPH=IVAL(7)
    C
    GOTO (1,2,3,4,5,6,7,8,9) MODE
    IF (IFTSM.GT.1000)MODE=2
    IF (IFTSM.GT.1800.AND.ILOAD.LT.LDLV) MODE=7
    RETURN
    IF (ILOAD.LT.LDLV) MODE=3
    IF (IFTSM.LT.1000) MODE=1
    RETURN
    IT3=IT3+1
    IF (IFTSM.GT.1000) GOTO 31
    MODE=1
    IT3=0
    RETURN
    IF (ILOAD.LT.LDLV) GOTO 33
    MODE=2
    IT3=0
    RETURN
    IF (IFTSM.LT.1800) RETURN
    MODE=4
    TM=FLOAT(ITM(IT3))+GP1
    U(4)=GAL/(TM*TM)
    RETURN
    IT4=IT4+1
    IF ((IVV.GT.15).OR.(IT4.LT.15))RETURN
    MODE=5
    IPMAX=IPP
    TN=FLOAT(ITN(IT3))+GP2
    U(5)=GAL*(TN*TN)
    RETURN
    ITS=ITS+1
    IF (IPP.LT.20)MODE=6
    IF (IFTSM.LT.1800) MODE=8
    RETURN
    IT6=IT6+1
    IF (IFTSM.LT.1800)MODE=9
    RETURN
    IF (IFTSM.LT.1800)MODE=1
    RETURN
    JTS=JTS+1
    JT(1,JTS)=ITS

```

```

JT(2,JTS)=IT4
JT(3,JTS)=ITS
JT(4,JTS)=IT6
JT(5,JTS)=ITM(IT3)
JT(6,JTS)=ITN(IT3)
JT(7,JTS)=IPMAX

C update the timing prediction
82 IF(IT3.GT.25.CR.IT5.LI.15)GOTO 83
ITM(IT3)=IT4
ITN(IT3)=ITS
PMGF(IT3)=FLOAT(IPMAX)*U(1)
83 IT3=0
IT4=0
IT5=0
IT6=0
PD=0.0
VD=0.0
MODE=1
RETURN
9 IF(JTS.GT.199)GOTO 92
JTS=JTS+1
JT(1,JTS)=IT3
JT(2,JTS)=IT4
JT(3,JTS)=ITS
JT(4,JTS)=IT6
JT(5,JTS)=ITM(IT3)
JT(6,JTS)=ITN(IT3)
JT(7,JTS)=IPMAX

C update the timing prediction
92 IF(IT3.GT.25.CR.IT5.LI.15)GOTO 93
ITM(IT3)=IT4
ITN(IT3)=ITS*IT6/2
PMGF(IT3)=FLOAT(IPMAX)*U(1)
93 IT3=0
IT4=0
IT5=0
IT6=0
PD=0.0
VD=0.0
MODE=1
RETURN
100 END

```

```

C.....
SUBROUTINE CONTRO
C.....
COMMON IOFF(3),IZ,Z,ISW1,ISW2,ILCK1,ILCK2,LDLV,U(5),
* PMKF(200),ITM(200),ITN(200),IT3,IT4,IT5,IT6,JTS,JT(9,200),
* MODE,IP(3),IV(3),IL(3),IPP,IVV,IFTSW,ISIG,ILOAD,ISTOP,IPH,
* PD,VD,SIG,SIGP,SIGV,IPMAX,
* A1,B1,C1,D1,A2,B2,C2,D2,GAL1,GAL2,GP1,GP2,GV1,GV2,IN,IN,
GOTO(1,2,3,4,5,6,7,8,9) MODE
ISIG=ILCK1
CALL IDAC(ISIG)
RETURN
1 ISIG=ILCK2
CALL IDAC(ISIG)
RETURN
2 ISIG=ISW2
CALL IDAC(ISIG)
RETURN
3

```

```

RETURN
T4=FLOAT(IT4)
T44=T4*T4
SIG=U(4)*FLOAT(IVV)*T44/(T44+GV1)
IF(SIG.GT.4.0)SIG=4.0
ISIG=IZ+IFIX(Z*SIG/5.0)
CALL IDAC(ISIG)
RETURN
5 SIG=-U(5)*FLOAT(IVV)
IF(SIG.GT.4.0)SIG=4.0
ISIG=IZ+IFIX(Z*SIG/5.0)
CALL IDAC(ISIG)
RETURN
6 ISIG=ILCK1
CALL IDAC(ISIG)
RETURN
7 ISIG=ISW2
CALL IDAC(ISIG)
RETURN
8 ISIG=ILCK1
CALL IDAC(ISIG)
RETURN
9 ISIG=ILCK1
CALL IDAC(ISIG)
RETURN
100 END

```

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