THE UNIVERSITY OF MANITOBA

A VOLUNTARILY CONTROLLED VARIABLE RESISTANCE ABOVE-KNEE PROSTHESIS

BY

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ABSTRACT

Most above-knee amputees to date are using prostheses employing either constant mechanical friction or some type of programmed hydraulic damping, over which the wearer has no control and limited gait speeds.

The main aspect of this thesis is to design and test a new system in which voluntary control of a lower limb prosthesis is derived from the EMG signals of residual thigh muscles in the stump (originally involved in knee control), and after suitable conditioning, these signals operate solenoid bypass control valves in the shank. These valves form a closed hydraulic loop with a damping cylinder in the knee joint. Thus the amputee is able to voluntarily open and close the valves, and vary the resistance to fluid flow around the hydraulic cylinder from 'free swing' to full 'lock'.

The main advantages of this system are a variable and more aesthetic gait, stability over uneven terrain, and because the 'lock' is only to flexion, the amputee can rise on his prosthesis and so use a passive appendage as an active element of his skeleton.

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CHAPTER I

INTRODUCTION

The design and development of an improved prosthesis for the above-knee amputee constitutes an unusual problem for the engineering designer. The prosthesis worn by an amputee is a specialized mechanism which becomes a part of all his daily activities, and must provide unfailing performance under many different conditions. Its design must therefore be based on both biological and engineering principles.

Historically, the improvement of prosthetic devices has depended on the skill and imagination of an experimentally minded fitter or an inventor working privately. The medical profession has helped to establish certain functional requirements, based on knowledge of the physiological behaviour of normal joints and musculature. Many remarkable devices have thus been developed, although only a few have been manufactured commercially in quantity.

The main objective in designing a lower limb prosthesis is, 'to give the knee joint of an above-knee prosthesis as many as possible of the weight bearing functions which the quadriceps and hamstrings give the natural knee joint'. The effect which the quadriceps group exerts on the knee joint in weight bearing is threefold: it can extend the joint (which is also the case in swing phase), lock it, or let it yield slowly. To provide the first function, external energy requirements are necessary, where as the second and third functions can be achieved via purely passive devices (a glossary of medical terminology is given in Appendix 1.)

The most problematic part of the previous statement is the qualification, 'as many as possible', which as always is dictated by the state of contemporary technology, implying the need for making design decisions regarding the benefit of possible functional features versus the burden they impose in terms of complexity, maintenance needs, weight increase, training requirements, cost, etc.

1.1. Mechanical Friction Prosthesis

In present use, approximately 85 percent of the above-knee prostheses employ mechanical friction, which is approximately constant for the control of swing phase, and in some cases a locking mechanism to maintain the knee in full extension during the weight bearing phase. These prostheses are mostly wooden, with some variations in knee mechanisms [1] (Fig. 1). Constant friction during swing phase, however, is undesirable. For example, the angular velocity of the swinging shank must be stopped before heel contact, requiring higher resistance immediately prior to heel contact; and deceleration of an upward swinging shank is also desirable. Therefore some inventors (such as Wheeler, Catranis, and Radcliffe [2]) developed the variable friction knee with locking mechanisms at full extension, for stability.

The advantages to this type of prosthesis are it is simple, reliable, low cost, and makes use of standardized equipment. Its disadvantages are poor swing control, difficulty in knee flexion at pushoff, and difficulty in walking up stairs and ramps, because the amputee cannot lock the leg in a position other than full extension and therefore cannot support weight on a partially flexed knee - he simply isn't strong enough. Also as a result of mechanical friction, the amputee is limited to one gait speed, which is a function of the mass and geometry of the

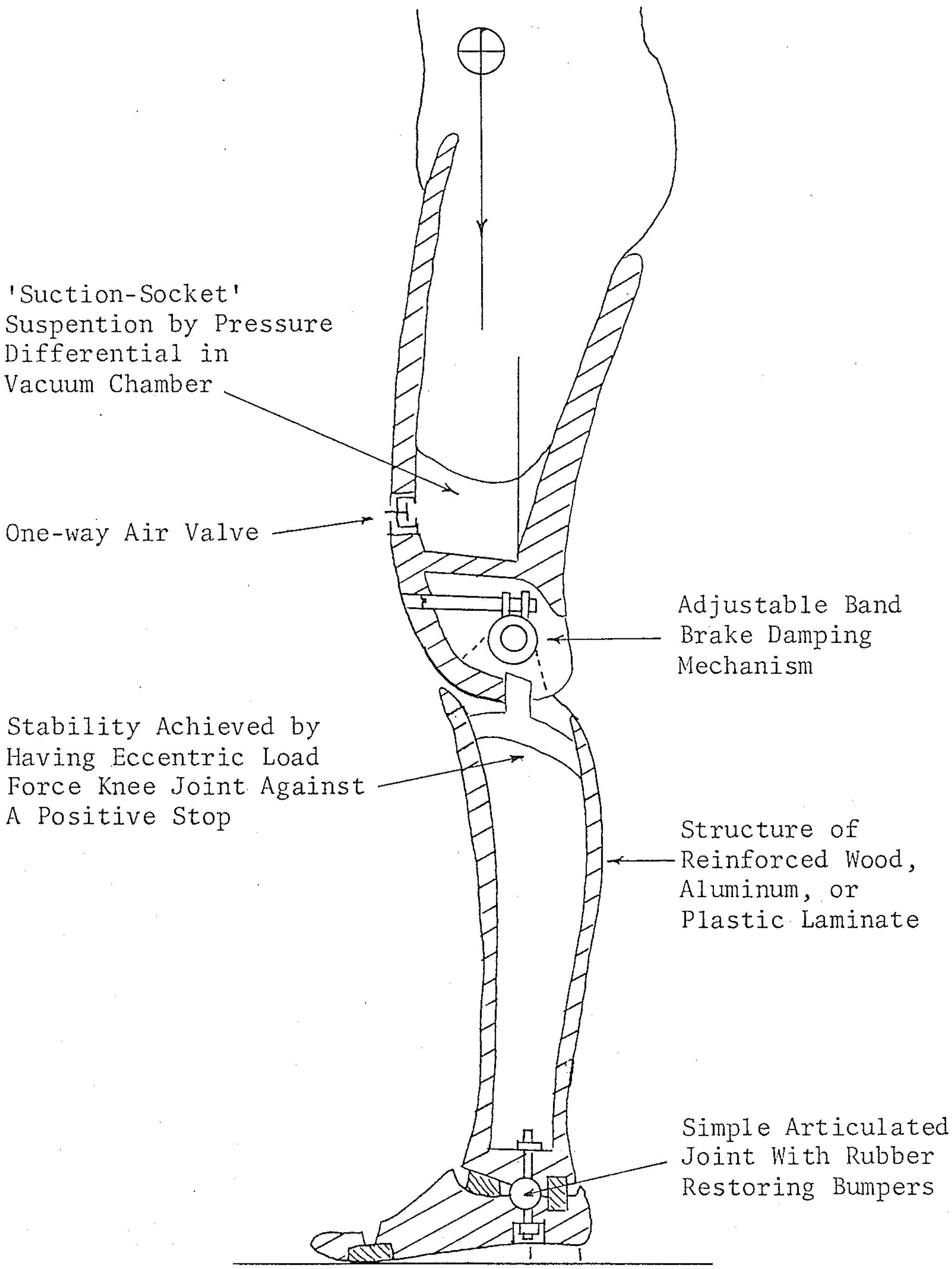


Fig. 1 Conventional Prosthesis for A/K Amputees

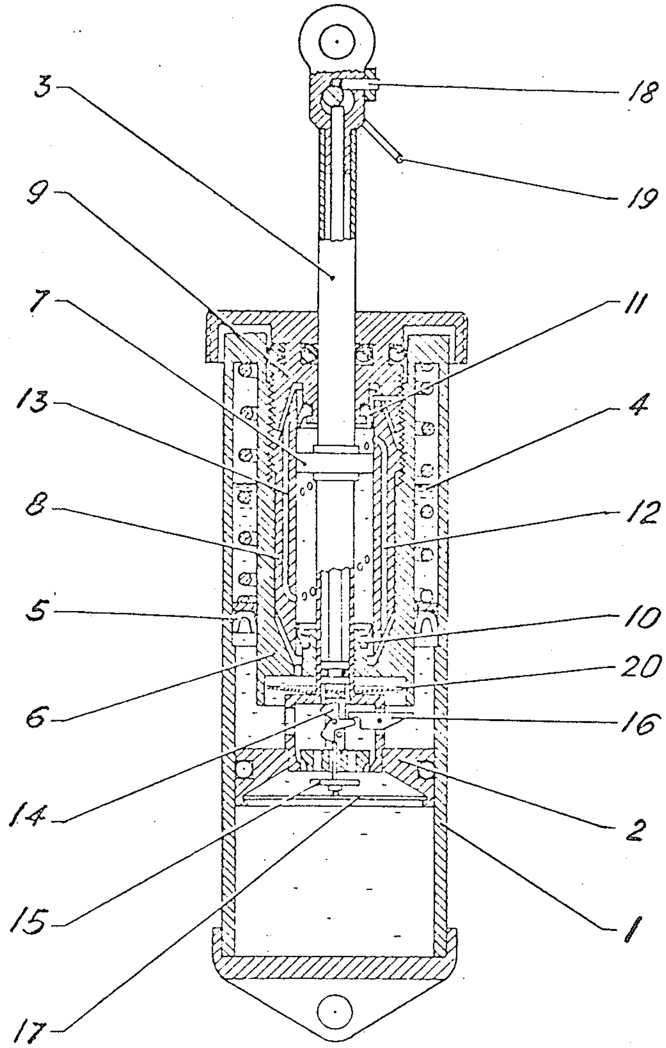
prosthesis. The effect of the knee locked in a fully extended position, instead of slightly bent, causes one to 'pole-vault' over the prosthesis which creates a high energy cost in walking.

1.2. Hydraulic and Pneumatic Controlled Prostheses

The inherent flexibility and basic simplicity of hydraulic systems, with the possibility of designing a yielding lock in place of a full lock, together with the recent advances in the design in hydraulic components and controls, has directed more recent investigations toward the development of hydraulic locking and control devices. But as in the mechanical knee devices, provision of an adequate control still remains a major unsolved problem. The standard example of such a hydraulic control prosthesis is the Henschke Mauch 'swing-n-stance phase' control knee [3](Fig. 2). Catranis [2] is another name associated with this type of prosthesis and has at least ten different models to his credit.

In these cases, the damping resistance is obtained from a hydraulic cylinder in the knee which is constructed in such a way so as to produce a damping resistance which is a function of knee angle and direction of angular velocity [4, 5, 6]. In the Henschke Mauch system, two levels of patterened resistance to knee flexion are incorporated into the design; each are independantly adjustable. The lower level of resistance provided resistance to achieve flexion characteristics, during swing phase, that are most suitable to the amputee's gait pattern. The higher level of resistance provides resistance to flexion of the knee at all other times.

The design of the system provides that the higher level of resistance acts at all times until the amputee generates a prolonged



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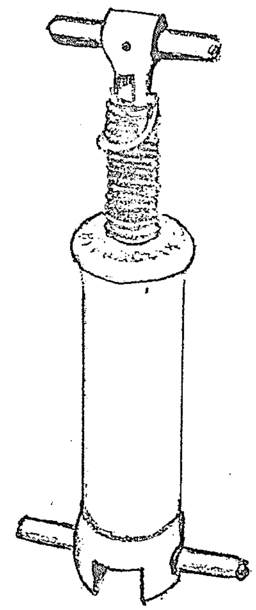
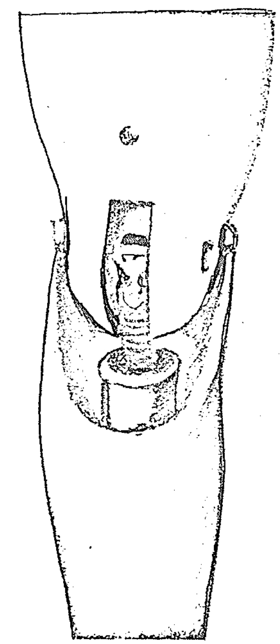


A.



B.

Fig. 2 Henschke-Mauch Hydraulic System
A. Cylinder Including Attachment Bolts
B. System in a Prosthetic Knee Joint
C. Exploded View Showing: (1) Cylinder, (2) Stance-Control Piston, (3) Piston Rod, (4) Hydraulic Fluid, (5) Accumulator Piston, (6) Dashpot, (7) Swing-Control Piston, (8) Control Bushing, (9) Swing Adjustment Screw, (10,11) Check Valves, (12,13) Fluid Channels, (14) Pendulum, (15) Valve, (16) Counterweight, (17) Spring, (18) Stance Adjustment Screw, (19) Selector Switch, and (20) Belleville Spring.

hyperextension moment above the knee. This hyperextension moment occurs naturally while walking as the amputee rolls over the ball of the prosthetic foot after mid-stance. Prolonged hyperextension moment, which can only occur when the knee is safely extended, results in disengagement of the high resistance range and permits the knee to flex properly to begin swing phase. As the knee reaches maximum flexion and the speed of rotation decreases to almost zero velocity, the higher level of resistance is reinstated (similar to automatic downshift of an automobile's automatic transmission as speed decreases). Thus, if during extension of the shank the toe is stubbed, the high resistance to flex is available to aid in stumble recovery.

Release of the high flexion resistance typically available in stance phase can be accomplished by an amputee who is standing and wishes to sit down easily. He simple extends his stump while maintaining the foot in contact with the floor, thus generating and maintaining a hyperextension moment at the knee for at least a tenth of a second. With the high flexion resistance released, stump flexion initiates knee flexion which he continues as he then sits down. Should the amputee walk in such a way that the knee is not fully extended and a hyperextension moment is not generated in stance, then the knee will not flex freely.

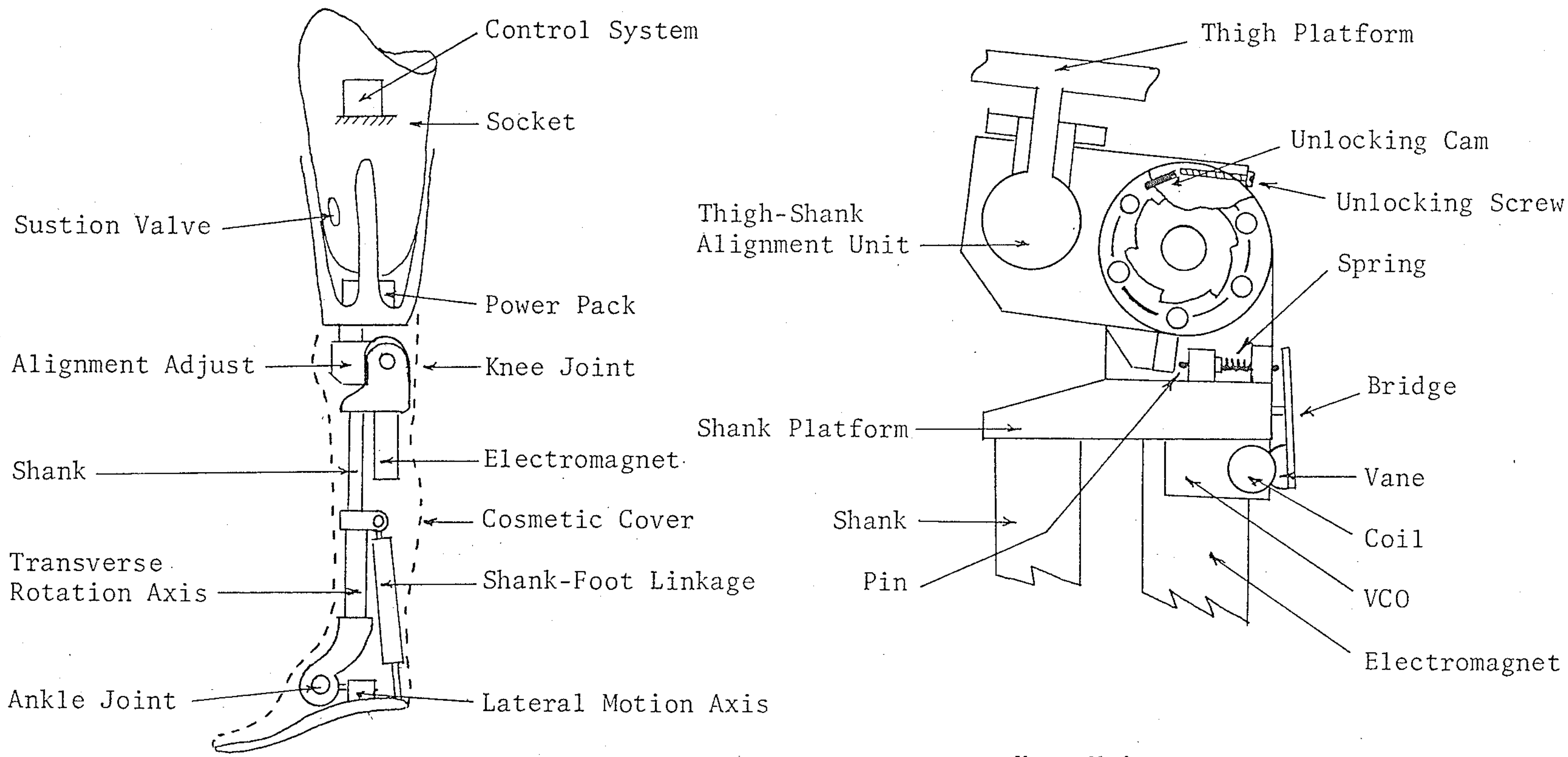
Using this model, the amputee has the choice of stance control or no stance control, of yielding lock or full lock, yet to employ these modes, must manually set a stirrup lever in the prosthesis to a predetermined position. Therefore, the mode is somewhat voluntary, but still manual rather than automatic.

At this point it may be worthwhile mentioning that it is generally agreed that hydraulic dampers, whose resistance varies with the angular velocity of the knee, have better characteristics than either the prostheses using dry mechanical friction or pneumatic dampers. Mechanical friction is essentially independent of speed, and a prosthetic knee equipped with enough mechanical friction to provide yielding support for the amputee at a given knee angle would tend to collapse increasingly violently as knee flexion progresses. With pneumatic devices, a speed dependant, self-stabilizing, yielding function can be provided by letting the compressed air escape through an orifice, however, they are not suitable for producing a firm locking function because of the compressibility of air. Moreover, this elastic compressibility will also reduce the value of the yielding function by producing an undesirable 'bouncy' support and by causing the leg to extend upon weight removal, leads to the risk of scuffing and stumbling.

Also, the resistance cannot be voluntarily controlled, an amputee is again limited to a predetermined range of gait speeds and still must 'pole-vault' over his prosthesis. Generally, this construction does not allow for locking the knee when it is flexed, and therefore is not much of an improvement when it comes to walking up or down stairs.

1.3. Myoelectrically Controlled Knee Locking Mechanism

A recent development (1970) in above knee (A/K) prostheses was done by G.W. Horn [7], in Italy, who developed an EMG controlled A/K prosthesis flexion lock (Fig. 3). Locking of the flexion of the knee joint in this prosthesis can be accomplished by means of a mechanical clutch released by a control pulse derived from the myoelectric activity of the amputee's stump. When locked, the knee joint cannot be flexed any more, but is still free to extend under the amputee's control. At



Knee Unit

Mechanical Layout

Fig. 3 Horn's Voluntary Lock Prosthetic Knee Unit full extension, the knee unlocks allowing the limb to swing freely in a normal manner.

Horn's prosthesis employs a rather unconventional control system. Pulse triggering is used, instead of pulse integration as in upper extremity prostheses. A logic devise inhibits the detection system once the knee lock has occurred. The whole system is automatically reset by a vane controlled oscillator as soon as the patient fully extends the artificial leg.

Two advantages of this system are, a wearer can bear weight on a slightly bent knee allowing for a more normal appearing gait, plus a wearer can manage to climb stairs on his prosthesis in a normal way by not being restricted to taking one step at a time. This design does not, however, allow the amputee to vary the resistance as a function of the knee angle nor to control this resistance in order to accommodate different gait speeds.

1.4. Voluntarily Controlled Variable Resistance Device

Research has begun on hydraulic dampers whose resistance can be voluntarily controlled by the wearer through the use of either myoelectric or various mechanical signals [6]. This appears to be the course that future efforts will follow. The following is a quotation out of a report by the Committee on Prosthetic Research and Development of the National Academy of Science [8], Washington, D.C.,

'Electromyogram and various mechanical signals have been suggested as sources for voluntary control of artificial knee joints. This approach is certainly the next phase in the development of improved knee-joints. Mauch Laboratories, UC-Berklye, and VAPC are interested in the development of such systems.'

The research pursued in this thesis follows this course. Voluntary control of the lower limb prosthesis is derived from EMG signals of residual thigh muscles in the stump, and after suitable conditioning, these signals operate solenoid bypass control valves in the shank. These valves form a closed hydraulic loop with a damping cylinder in the knee joint. Thus the amputee can voluntarily open or close the valves, and vary the resistance to fluid flow around the hydraulic cylinder from free swing to full lock. When actually tested on an amputee, this electro-hydraulic system was operated very well, very naturally, and in a very short time.

CHAPTER II

FUNCTIONAL CRITERIA [1]

2.1. Introduction

The functional criteria for the design of an improved artificial limb may be established in many ways. For example, the amputees themselves may be asked which improvements in function they consider most necessary as a result of their personal experience. Doctors and prosthetists may be questioned in a similar manner. Another means lies in a comprehensive fundamental study of the process of normal human locomotion. The purpose of such a study is to obtain, from a better understanding of how a normal person walks, data useful in establishing design criteria. What is needed, then, is a technique that will yield a complete analysis of the magnitudes, directions and rates of change of translations, rotations and forces in the lower extremity. A current study is now taking place in the Winnipeg Shriner's Hospital Locomotion Laboratory under the direction of Dr. D.A. Winter and Mr. D. Hobson. Most of the data used in the following chapters were taken from these studies.

Human locomotion, a complicated process largely taken for granted, involves the transformation of a series of controlled angular movements occurring simultaneously at the several articulations of the lower extremities into a curvilinear motion of the center of gravity of the body considered as a whole. The major function of the lower extremity during bipedal locomotion on a level surface is to propel the body forward while the total mechanical energy of the body is maintained at as nearly constant value as possible. Any change in the total energy level, whether input or output, must be accompanied by muscle work and, hence, fatigue. Changes in potential energy are minimized by holding the center of gravity in kinetic energy are reduced by propelling the body with minimum acceleration and deceleration.

Fig. 4 indicates the co-ordinate system to be used in describing a typical walking cycle. The function of the muscle groups acting at the joints of the lower extremity will be described in terms of their internal behaviour with respect to the leg. For instance, the net external load acting on the foot at a certain instant may tend to cause flexion at the knee joint. In order to maintain a stable angulation of the joint, muscle tension must oppose this action; that is, there must be an internal extension moment. All moments at the knee joint will be described as internal moments applied at the knee joint by either internal musculature or an internal prosthetic device.

A single step consists of two separate parts - the stance phase and the swing phase. The stance phase is characterized by the need for stability and by the fact that the amputee is readily able to determine the location of the prosthesis. During the stance phase, the movement of the prosthesis is under more or less direct control of the amputee. The swing phase is characterized by the need for smooth effortless motion and for some means to enable the amputee to sense the instantaneous position of the prosthesis. During the swing phase, the motion of the prosthesis is controlled more by the inherent characteristics of the device than by the action of the amputee.

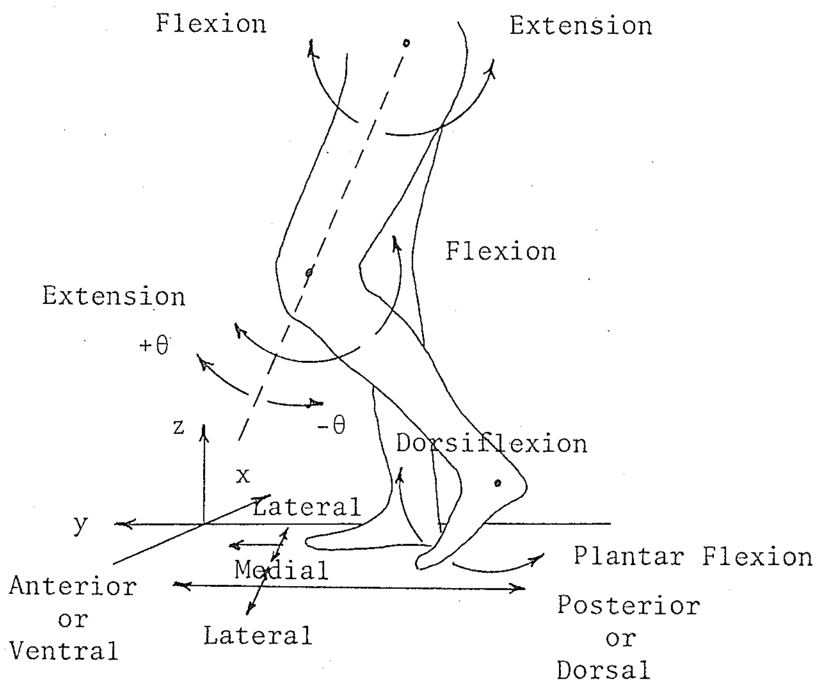


Fig. 4 Coordinate System for Description of Displacement Pattern

2,2, Stance Phase

2.2.1. *Heel Contact*

At the end of swing phase, just prior to heel contact, the entire lower extremity undergoes rapid deceleration preparatory to planting the heel for stance phase. Primary muscle activity during this period (Fig. 5) is concentrated in the hamstring muscle group. The hamstring group is attached to the pelvis posterior, to the hip joint, and to the posterior aspect of the tibia below the knee joint. Contraction of the hamstring group may cause hip extension, knee flexion, or both simultaneously. As the foot strikes the floor the heel is brought back forcibly into contact, causing an initial forward horizontal component of the floor reaction force. This ensures that the knee is secure against any tendency to buckle at the time of heel contact, since the external moment at that instant would be such as to extend the knee. The internal knee joint moment, due to hamstring action, is a positive moment tending to maintain knee flexion.

Force reaction at the hip joint during the deceleration of the lower extremity near the end of swing phase accounts, in part, for the increase in velocity of the remaining mass of the body. Both leg deceleration and body acceleration are due to the internal force system at the hip joint. Push-off from the opposite foot also causes an increase in velocity of the center of gravity of the body during this phase. After heel contact, this increase in velocity must be reduced a corresponding amount in order to keep the average velocity of progression constant. This deceleration is accompanied by a reversal in direction in the anterior posterior component of the horizontal shear force applied to the sole of the shoe by the floor. A horizontal shear force which acts on the foot in a posterior direction tends to cause

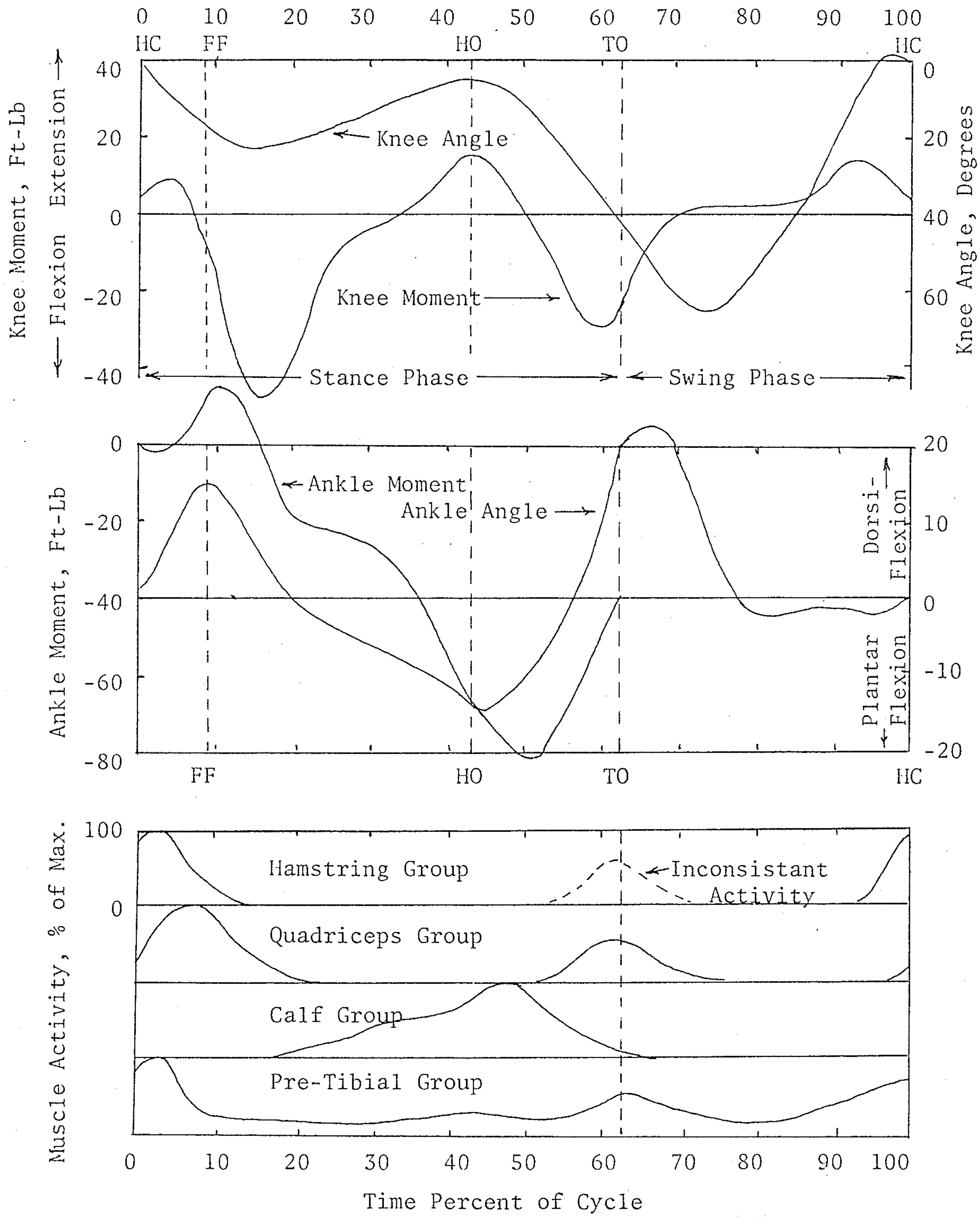


Fig. 5 Human Locomotion Data of Normals

knee flexion. However, a controlled amount of knee flexion is desirable just after heel contact. It reduces the energy required to walk by reducing the vertical rise of the body necessary to vault over the leg, since full knee extension then does not take place until the center of gravity of the body is anterior to the foot. It absorbs some of the shock at heel contact and, by providing means for deceleration in the horizonal plane, aids in providing a smooth reversal of the horizontal rotation of the pelvis as weight is transferred from one leg to the other. Therefore, at this time the quadriceps muscle group comes into action as an antagonist and provides an extension moment, controlling the amount of knee flexion.

Fig. 5 shows this action. Note the change in sign of the resisting muscle moment about the knee joint at about 6% of the walking cycle. At this point the muscle action is changing from predominantly hamstring to quadriceps action. The ankle also contributes to shock absorption in normal persons through the restraining actions of the pretibial group during plantar flexion at the ankle.

Ideally, the prosthesis, in order to serve as an efficient shock absorber, should provide controlled knee flexion, simulating the action of the quadriceps, as well as controlled ankle flexion, similar to the action of the pretibial group. A further requirement is that the motion of the two joints be co-ordinated so as to balance energy dissipation at the two joints as well as provide for normal aesthetic appearance in the gait pattern.

2.2.2. *Mid Stance*

During the second portion of stance phase, designated as mid-stance, the body is carried over the ankle pivot on a flexed but gradually extending knee joint. The center of gravity of the body accelerates downward, and the vertical load on the leg reaches a minimum value of approximately 60% of the body weight. The center of vertical pressure on the sole of the foot moves forward rapidly, and the fore-and-aft horizontal shear force changes from a rearward decelerating force to a forward accelerating force. This results in a change of the externally applied knee moment from flexion to extension and causes the knee joint to move into full extension. The phenomenon of restraining the knee in a flexed position at heel contact, relaxing to allow general extension during mid-stance, and locking again in full extension has been called the 'double-knee lock action' in normal locomotion, and is important in accomplishing a smooth, energy saving gain in normal persons.

Thus, ideally the prosthesis should allow a sequence of restrained initial flexion, yield, and lock in full extension during stance phase if a typically normal gait is to be achieved. This sequence must be properly phased with the ankle action.

2.2.3. *Push-Off*

The third period in stancephase is the so-called push-off. The normal leg accomplishes two essential functions during push-off. It maintains the smooth forward progression of the body as a whole and initiates the angular movements of the swing phase that follows. This is accomplished by active extension of the ankle by the calf group, which provides the energy source for forward propulsion and simultaneous flexion of the knee, primarily to accelerate the leg ahead into swing phase. The knee is flexed forward in such a manner as to reduce the offset distance between the knee and the inclined forward line of the floor reaction. This reduces the internal knee moment and allows more precise control of the knee movement through muscular action about the hip and knee joints. When the toe leaves the floor the knee has flexed 40-45° of the maximum 65° it will reach during swing phase. Knee flexion in swing phase is not primarily due to hamstring action in normal persons, as might be presupposed.

Complete prosthetic restoration of normal function in push-off is difficult, if not impossible. A proprioceptive sense of knee position by the amputee would be necessary, as well as an active source of energy in the ankle. Because there is no active source of ankle energy, initiation of knee flexion in amputees wearing a prosthesis must come from active hip flexion.

2.3. Swing Phase

2.3.1. *Deceleration of Upward Swinging Shank*

As soon as the toe is clear of the ground, an extending or decelerating moment of gradually increasing magnitude must come into action to stop the upward swing of the heel, increasing in magnitude as cadence is increased, for heel rise of the normal is approximately constant irrespective of cadence, increasing only from about 10.5 inches at 60 steps per minute (SPM) to 11.5 inches at 125 SPM, the full range of cadence in normal walking.

The knee flexes and continues to flex after toe-off. During rapid walking, this would result in excessive knee flexion were it not for the action of the quadriceps group in limiting the angle of knee flexion to approximately 65 degrees and then starting knee extension. Knee extension continues as a result of a combination of pendulum effects, due to the weight of the inclined shank foot, and muscle action. Little quadriceps action is required, since other factors are also important; for example, the primary flexor of the hip, iliopsoas, contributes by development of active hip flexion which accelerates the knee forward and upward.

2.3.2. *Mid Swing*

Immediately following maximum knee flexion, knee extension occurs as a result of the continuing extending moment that acted to cause shank deceleration. Ankle flexion continues into slight dorsiflexion. During this period of the swing phase, the only allowable resistance to extension of the knee, if any, would be that slight amount necessary to provide a sense of perception of the position of the shank.

To obtain toe clearance at swing through, lift obtained by ankle dorsiflexion of 5 to 10 degrees may be desirable at this time. This would be a total of 25 to 30 degrees of ankle flexion from the position of maximum plantar flexion of 20 degrees at toe-off.

2.3.3. *Terminal Deceleration*

As the knee approaches full extension, a flexing or decelerating moment of increasing magnitude must occur to stop the forward swing of the shank smoothly and to prevent jarring of the amputee when the shank reaches full extension prior to heel contact. The magnitude of this moment must also increase with increasing cadence. (This terminal deceleration of the normal leg is primarily the result of hamstring action.)

2.4. Some Design Problems

In designing a prosthetic replacement for the complex system of musculature acting at the knee during swing phase, the difference in physical constants between the normal and prosthetic shank-foot must be accounted for. This can be done by assuming that a shank-foot with the physical constants of a typical prosthetic device is forced through space in a normal manner by a system of forces and moments acting at the knee joint. These forces and moments will be smaller than those calculated for a normal knee joint, since the prosthetic shank-foot has considerably less mass. A natural shank-foot for a 165 lb. man weights about 4 lb. Another assumption made, essential in prosthetic manipulation, is that there is an intimate and functional fit between stump and socket and that the amputee will be able to control the motion of the socket in a comparatively normal manner.

To fit the needs of the individual amputee correctly, it has been found that resistance to dorsiflexion and plantar flexion in the ankle must be independently adjustable. Likewise inversion and eversion should be adjustable independant of dorsiflexion and plantar flexion. And similarly, the resistance to transverse rotation should be substantially independent of the adjustment for resistance to plantar and dorsiflexion and inversion and eversion.

Stability in the above knee prosthesis has long been achieved by placing the knee axis slightly posterior to a line joining the ankle axis and center of gravity with the amputee standing balanced with body weight on the prosthesis. This so-called 'alignment stability' causes resistance to initial flexing just before toe-off, results in a 'snapping' action when flexure does start, and, when excessive, also causes a 'snapping' action at full extension. One of the problems with which designers have been faced has been the incorporation of the advantages of this simple alignment stability in new devices, at the same time overcoming the disadvantages of combining the principle with the various devices under development. It is desirable, for instance, to provide for alignment stability of some degree so that, in case the knee lock or controls become inoperative, the amputee can still walk.

Special activities, such as stair and ramp ascent and descent, do not introduce new functional requirements, although they do affect the adjustment of the prosthesis to the amputee. In general it is desirable to maintain a high degree of voluntary control over the tendency of the knee joint to buckle in flexion. In many activities, such as descending ramps, the amputee must be able to flex the knee forward at precisely the right instant or be in danger of being pitched forward over the extended knee.

Finally, as in all prosthetic devices, the design must take into account the problems of simple maintenance, considering the facilities available in most contemporary limb shops and the fact that most users will themselves be untrained and unskilled in adjustment and maintenance.

2.5. Summary

The following is a summary of the functional requirements of a 'maximum function' prosthesis under normal conditions of level walking:

Stance Phase

a) Knee stability at heel contact and during weight bearing

b) Controlled plantar flexion at ankle to absorb impact at heel contact

c) Smooth knee flexion at push-off

d) Fairly stiff dorsiflexion action at ankle

e) Controlled emergency braking action for fall recovery

f) Provide for response to changes in cadence, especially as to control of knee flexion at the start and end of stance phase

g) Respond to the actuating signal without delay perceptible to the amputee

h) Use normal muscle patterns, where there is a choice, and not require reversal or changing of these patterns in controlling particular activities

Swing Phase

a) Smoothness of action, friction free forward swing of shank

b) Simulation of muscle functions about the knee, including deceleration of an upward and forward swinging shank, etc.

c) Automatic adjustment to varying walking speeds.

It must also be kept in mind, that constant attention should be focused on a minimum consumption of energy in normal walking.

CHAPTER III

DESIGN OF THE HYDRAULIC SYSTEM

3.1. Proposed Design

The proposed system for a voluntarily controlled variable resistance hydraulic knee unit for an above knee prosthesis is shown in Fig. 6. A hydraulic damping cylinder is placed across the knee joint, and across this cylinder are a set of solenoid bypass control valves. The solenoids on these valves are then connected to an EMG control unit, which takes raw electromyographic signals and converts them into a signal which controls the energizing of the correct combination of valves. In this manner, an amputee will be able to voluntarily control the resistance to flexion and extension of the prosthesis, from full lock to free swing, and thus actively control the passive resistance of the moving appendage.

The prosthesis used with this proposed hydraulic system is a standard endoskeletal wedge type adjust A/K prosthesis which was previously fitted with a DuPaco hydraulic damping cylinder. This prosthesis also has a SACH foot (solid ankle cushioned heel), which does not provide some of the desirable features looked for in the artificial ankle. For example, instead of controlled plantar flexion at heel contact, a cushioned heel is used to absorb the impact; the slight dorsiflexion for toe clearance during mid swing is lost; and functions such as inversion, eversion and transverse rotation are non-existant.

It should be noted though that this appliance does, however, include inherent alignment stability because of the fact that the knee axis is slightly posterior to the line joining the center of gravity with the ankle.

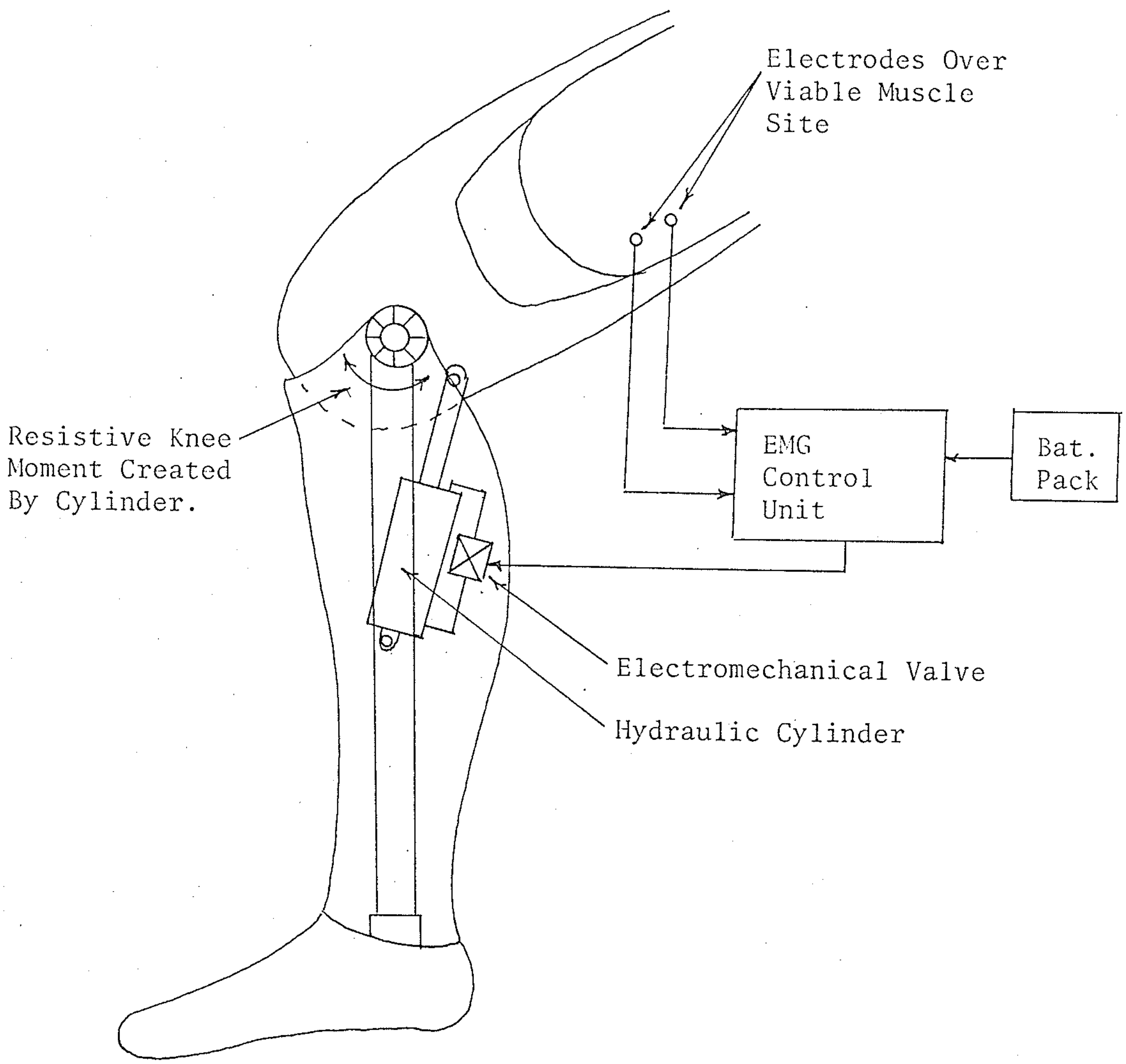


Fig. 6 Schematic of Proposed Electrohydraulic Knee Unit

In this design, a hydraulic system was chosen over the others because of its damping characteristics with respect to angular velocity, and because of the other advantages of such a system as was discussed in Section 1.2. It was hoped that the disadvantages also discussed, would be reduced by the new design considerably, and thereby provide a better appliance to an amputee.

Proportional control was considered but not used because there was simply nothing available commercially, that could satisfy all the constraints that would be imposed on such a controller when used in an A/K prosthesis. The following is then a design of a discrete system which would give adequate control during each walking cycle during gait.

3.2. The Model

Because of the preset ankle unit used in the design, the entire functional criteria focuses itself on the knee unit. In order for the characteristics of the hydraulic system in the knee to be determined, a model was set up. This model was used to convert known locomotion data of normals, such as angle of knee versus percent walking cycle and moment about knee versus percent walking cycle during normal ambulation, to useful design data for the hydraulic system. The result of such a conversion yields such facts as pressure ratings of each component, rates of flow each component is to possess, resistance to flow versus percent walking cycle and, therefore, the number of valves necessary and each of their resistances. This resistance to flow, creating the necessary flow rates and forces and thus moments, can be looked upon as a simulation for the quadriceps and hamstring action at the knee during walking.

The model (Fig. 7) represents the cylinder placement in the knee

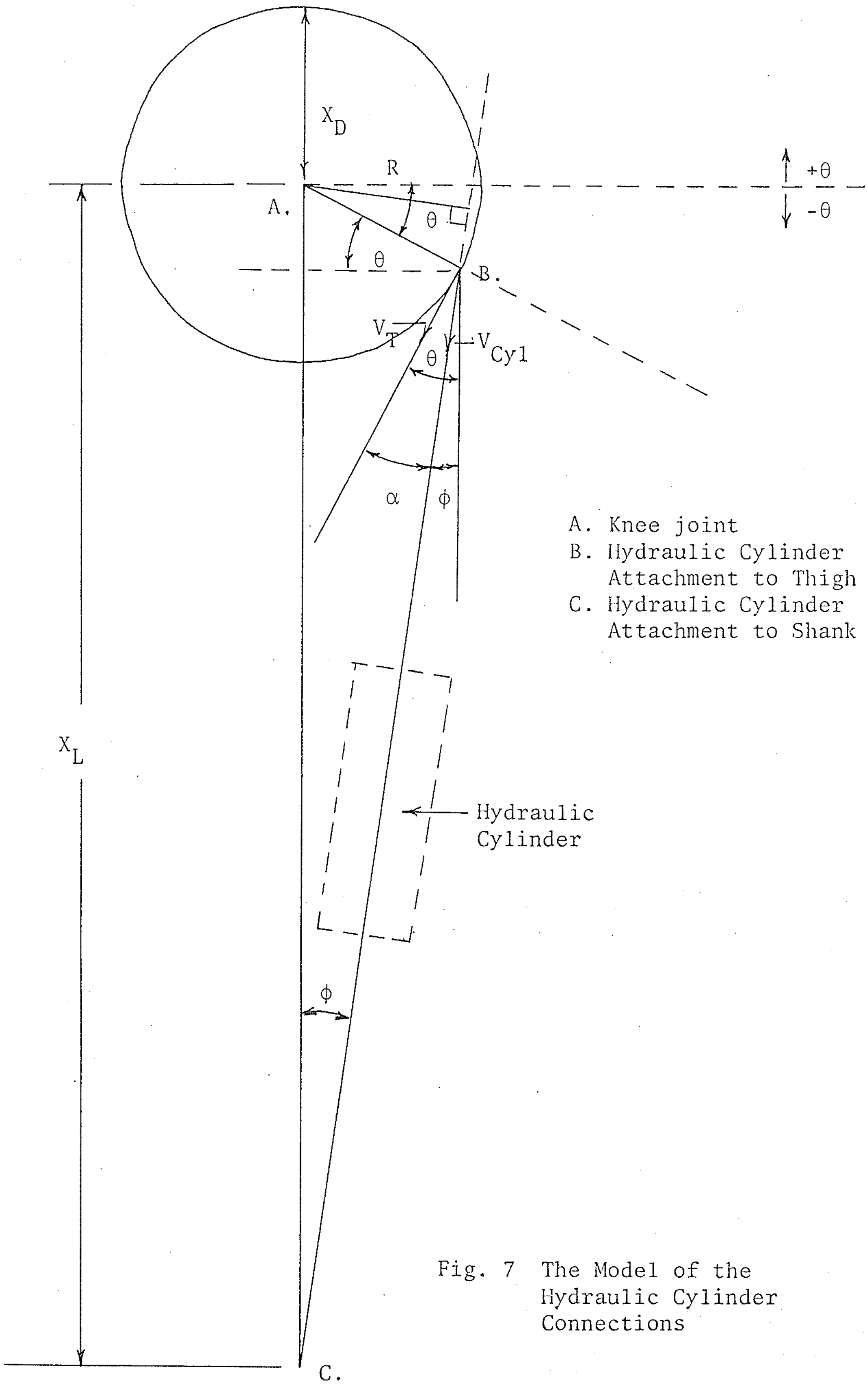


Fig. 7 The Model of the Hydraulic Cylinder Connections

unit, with A representing the centre of rotation of the knee, B representing the upper attachment of the cylinder with the thigh portion of the prosthesis, and C the lower attachment to the shank portion. The line AC, labelled X_L, on the diagram, is fixed in length and position in the model. The line AB, labelled X_D in the diagram, is fixed in length, but not position. It is the back and forth rotation of this line segment which simulates the angular change between the shank and thigh in the leg of a normal person while walking. The line BC is not fixed in position or length and simulates the damping cylinder stroke. The line segment labelled R is also variable in length and position and is defined as being the moment arm of any force developed in BC to create a moment about A.

3.3. The Design

The model was then first used to check if the cylinder placement in the knee could be improved. X_L, X_D, and the reference angle existing in the prosthesis were 7.5 inches, 1.125 inches and 0.0 degrees respectively. (The reference angle is defined as the angle X_D makes with the horizontal, with the knee fully extended). The criteria used was to calculate the force that the hydraulic cylinder would have to exert in order to cause a moment about the knee as would exist in normal walking, and then try to minimize the maximum force.

A computer program representing this model was then set up (Appendix 2). Forty points were taken off the graph of knee angle versus percent walking cycle of a normal (Fig. 8) and used in the computer program to increment the angle theta (θ) which is analagous to changes in knee angle in the model. Since θ is now known, the prosthetic

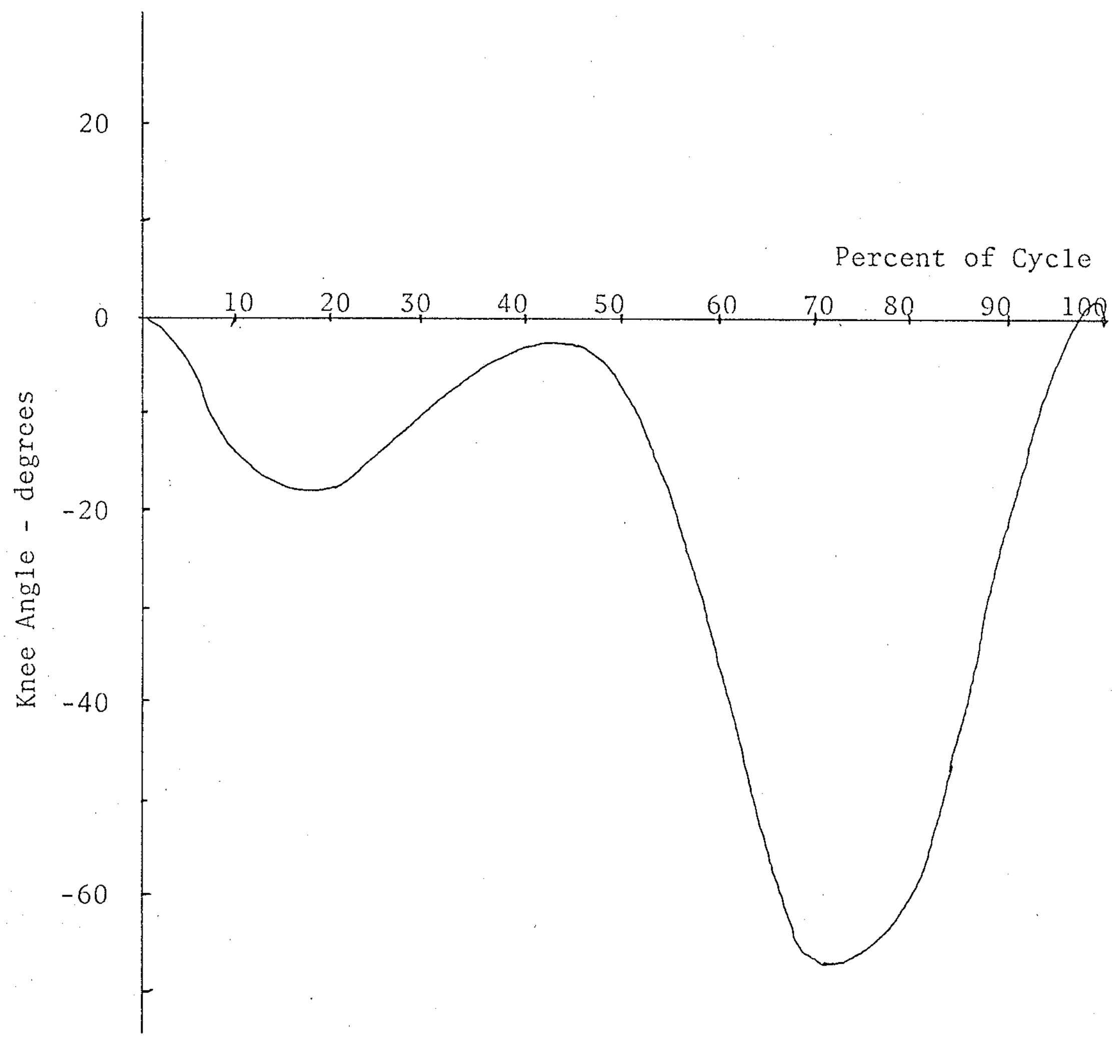


Fig. 8 Graph of Knee Angle vs. Percent Walking Cycle

knee unit (triangle ABC) was moved through one complete walking cycle and the moment arm of the cylinder, R, was calculated for each increment:

$$R = X_L \sin(\phi) \ ,$$

$$\phi = \arctan \left[X_D \cos(\theta)/(X_L - X_D \sin(\theta))\right]$$

The moment versus percent walking cycle of a normal (Fig. 9) was then read into the program (also forty points), and a force versus percent walking cycle of the cylinder was calculated.

$$F = \text{MOMENT} \ / \ R$$

To check the cylinder placement, three runs were performed on the computer. The first run consisted of holding $X_L = 7.5$ inches and $X_D = 1.125$ inches constant, and varying the reference angle, θr, with the result shown in TABLE 1. Next, $X_L = 7.5$ inches and $\theta r = 0.0$ degrees were held constant and X_D was varied with the result shown in TABLE 2, and finally X_L was varied with X_D and θr constant, with the result shown in TABLE 3.

It can be seen that increasing X_L, decreases the maximum force slightly. Yet, because of the physical dimension of the knee, X_L is constrained to be at least 7.5 inches.

Increasing X_D, which can almost be looked upon as a moment arm, will decrease the maximum force considerably, but because of the geometry of the prosthesis, is limited to 1.125 inches. The optimum reference angle is about 30 degrees, suggesting that the existing 0.0 degree reference angle is not an ideal placement. It should be noted, however, that if an energy storing device - such as a compression spring - were to be used in this device, to decelerate an upward swinging shank and

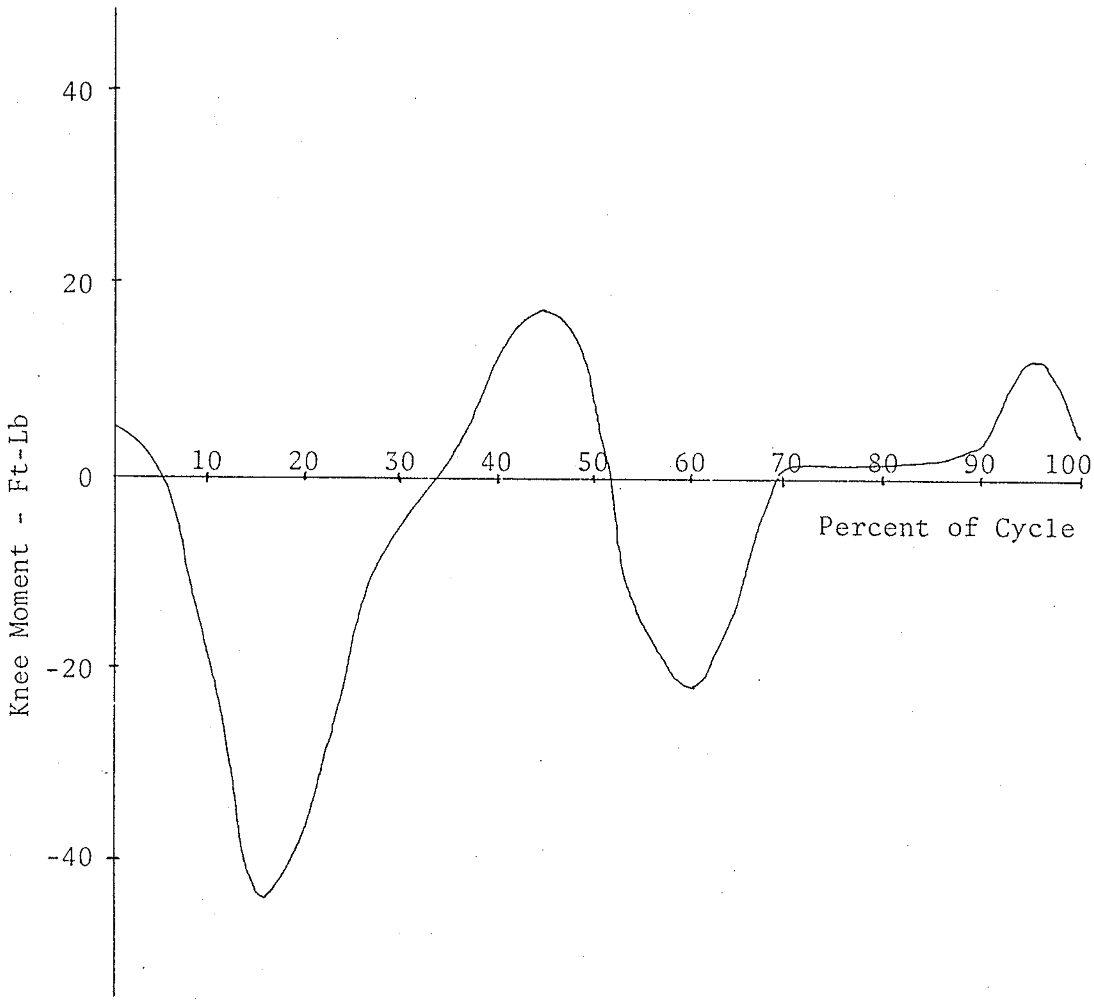


Fig. 9 Graph of Knee Moment vs. Percent Walking Cycle

TABLE I

Effect of Varying θ_r with $X_L = 7.5$ $X_D = 1.125$

θ_r	50°	45°	40°	35°	30°	25°	20°	15°	10°	5°	0°
F_{MAX}	506.2	490.1	479.0	472.3	469.5	470.3	474.6	482.4	494.0	509.6	529.8

TABLE II

Effect of Varying X_D with $X_L = 7.5$ $\theta_r = 0.0$

X_D	1.0	1.125	1.25	1.375
F_{MAX}	591.7	529.8	480.3	439.9

TABLE III

Effect of Varying X_L with $X_D = 1.125$ $\theta_r = 0.0$

X_L	6.0	7.0	7.5	8.0	9.0
F_{MAX}	538.7	532.3	529.8	527.6	524.1

then accelerate a downward swinging shank, the 0.0 degree reference angle would have to be chosen over the optimum 30 degree angle. The reason for this is obvious. From a standing to a sitting position, the knee flexes -90.0 degrees. If the θ_r were 30 degrees, the final flexed angle would be - 60.0 degrees, the cylinder and spring would be under maximum compression, and would possess a moment arm - resulting in a 'kicking forward motion' of the knee. One could not sit with a flexed knee and therefore sitting would appear abnormal. If, however, the θ_r were 0.0 degrees, the cylinder and spring would have no moment arm after 90 degrees of flexion and thereby exhibit no 'kick'. As a type of safety margin, the knee could be flexed slightly more than - 90.0 degrees, where upon a stopper is met. The spring would now, in effect, be acting to keep the knee very slightly flexed for a more normal seated appearance. Thus it is seen, that if an energy storing device is to be used, the cylinder is in the best possible placement.

Various velocities, flows, pressures, and resistances to flow were calculated next, using the same computer program, in order to establish the resistive characteristics of the bypass valves. Other data were also extracted from these results, as will be seen later, which completely describes the cylinder.

Flow is defined as a velocity through a cross-sectional area, which therefore necessitates the calculation of the piston velocity versus percent walking cycle. This was done as follows. Using the computer, the graph of knee angle versus percent walking cycle, was differentiated, which yielded the angular velocity of the knee versus percent walking cycle, (Fig. 10), labelled V_T on the diagram of the model (Fig. 7). Knowing the angular velocity of the knee, the piston

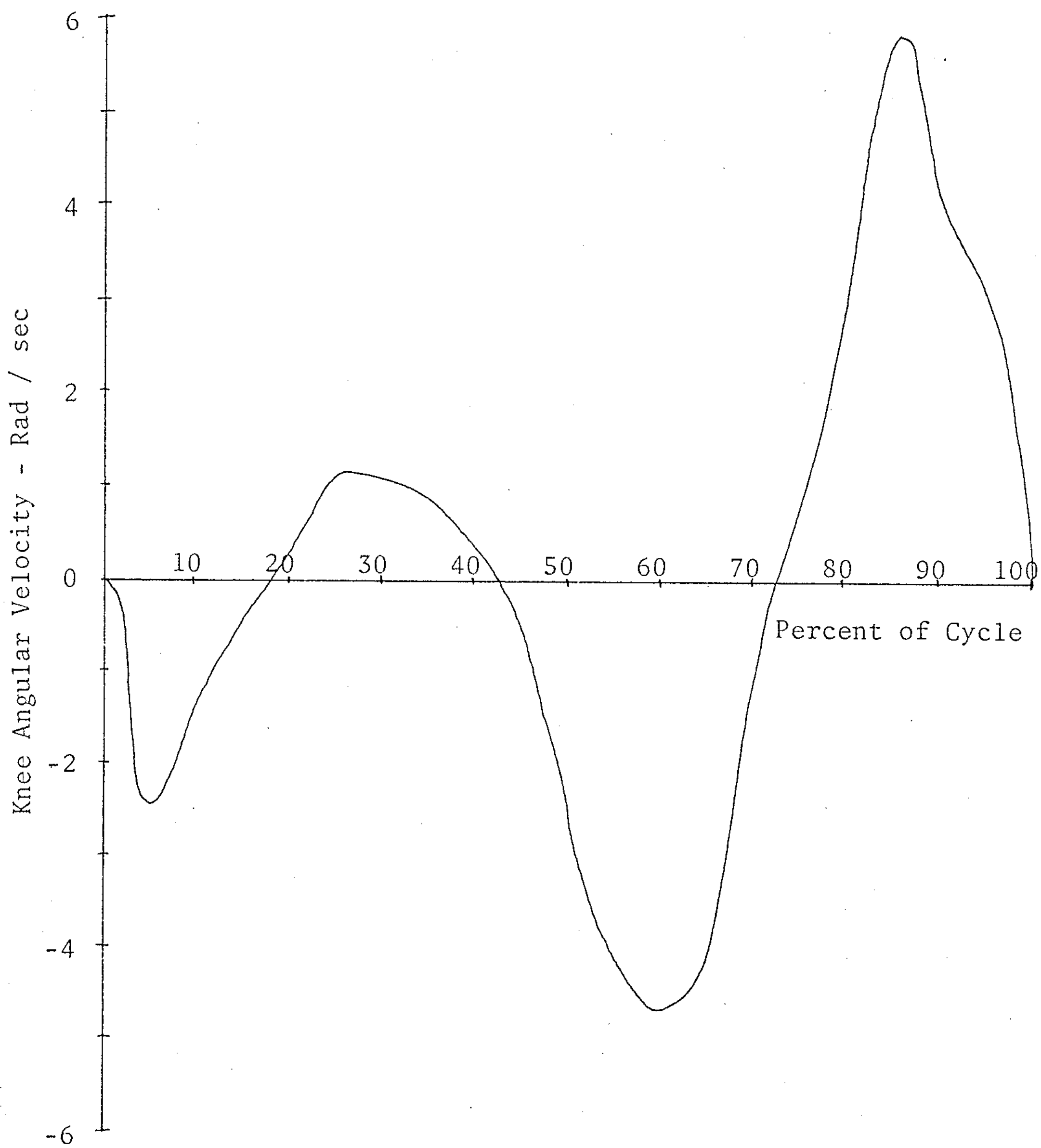


Fig. 10 Graph of Knee Angular Velocity vs. Percent Walking Cycle. (Positive excursions depict extension while negative excursions depict flexion.)

velocity, V_p, was calculated using the following formulas,

$$V_P = X_D \cdot V_T \cdot \cos(\alpha)$$

where $\alpha = \theta - \phi = \theta - \arctan(X_D \cos(\theta)/(X_L - X_D \sin(\theta)))$.

Now, in order to extract a flow measurement from the data, a cross sectional area of 1 sq. in. was chosen for the cylinder, because it seemed to be the appropriate size for the physical space available in the knee unit. This choice would make flow values correspond to those of piston velocity and also make pressure values correspond to those of force.

As was mentioned earlier, several bits of information can already be used from the above calculations. First, the pressure ratings of the components have now been established. Since the maximum force for the given geometry is 529.8 lb. and the cross sectional area of the piston was chosen as 1 sq. in., the normal operating pressure of the cylinder must be 529.8 psi. At this point, an electrical analogy is made. The cylinder, in which the pressures are produced, can be looked upon as a pressure or voltage source, and because the bypass valves are connected in parallel across the cylinder ports, they must also have the same voltage or pressure rating. In addition, because there exists a difference in physical constants between a normal and prosthetic leg, a component pressure rating of 500 psi was chosen as adequate. Here, also, a burst pressure calculation can be made. If an amputee suddenly was to find himself in a position where his full weight of 170 lb. was on his prosthesis, which was flexed to 60 degrees, a moment of approximately 1530 in-lb. would exist at the knee ($M \simeq 9'' \times 170$ lb.). The moment arm of the cylinder, R, at this point would be 0.643 inches which implies that the cylinder would have to produce a force of about 2400 lb.

to balance this moment. Therefore, a burst pressure of at least, say 5000 lb., is required.

Secondly, the bore and stroke of the cylinder can also be found. A cross sectional piston head area of 1 sq. in. corresponds almost exactly to a 1.125 inch diameter or bore. At hyperextension, the cylinder length is 7.622 inches while the length when sitting (or $\theta = 90°$) is 6.375 inches. Thus the stroke is 1.247 inches and 1.25 inches was chosen. In summary, the cylinder is completely specified:

bore = 1.125 in.,

stroke = 1.250 in.,

two port, double acting,

500 psi. operating pressure rating.

The next step taken was to decide how many valves are needed to bypass the cylinder ports, and what would their resistances have to be. This was accomplished by producing a graph of the desired resistance to flow versus percent walking cycle, and then attempting a curve fit with a combination of valves with known resistances to flow. The calculation of this resistance graph, however, is not as simple as dividing the pressures by the flow rates versus percent walking cycle (voltage/current) because the resistances of the commercially available valves were quite nonlinear, and thus a curve fit was impossible.

A valve's resistance to flow, however, is characterized by its C_v factor (analagous to electrical conductance) which is defined as the quantity of 60°F. water which will flow through the valve with a 1 psi pressure drop across it (given in gallons/min.). The approach to determine the valves needed for this system was then to calculate and plot the C_v factor required (which is nonlinear) versus percent walking cycle and try to fit a valve system to it. To do this, first a C_v factor of 1.0 was chosen. Referring to the graph of flow rate versus pressure drop for a C_v factor of 1.0, given in the manufacturer's specification sheets and reproduced on Fig. 11, the flow rates corresponding to the pressure versus percent walking cycle were fed into the computer. Now, knowing the desired flow rate for that pressure, a new C_v factor was calculated, which gives the correct flow for pressure during one complete walking cycle. ie.

$$C_v = \frac{\text{flow required at a particular pressure}}{\text{flow with a } C_v \text{ of 1.0 at that pressure}}$$

Because the medium used is hydraulic fluid instead of water, a correction factor of 1.087 must be used in determining flow, however, it is not needed here because it will cancel itself in the ratio.

Fig. 12 shows the resultant C_v versus percent walking cycle curve. Notice, that the closer the valve curve fit is to this graph, the greater will be the restoration of normal function, because this graph translates physiological requirements into mechanical requirements. For example, just after heel contact, there exists a high conductance (low resistance) which allows for initial knee flexion up to 5 percent walking cycle, followed immediately by a sharp decrease in conductance which simulates the quadriceps controlling the amount of flexion.ie., creating a yielding lock.

The C_v curve fit was decided upon in the following manner. First flexion was considered. A valve of $C_v = .02$ was chosen to approximate the low conductance portions of the graph.ie..gives high resistance to flow at various times during one walking cycle where

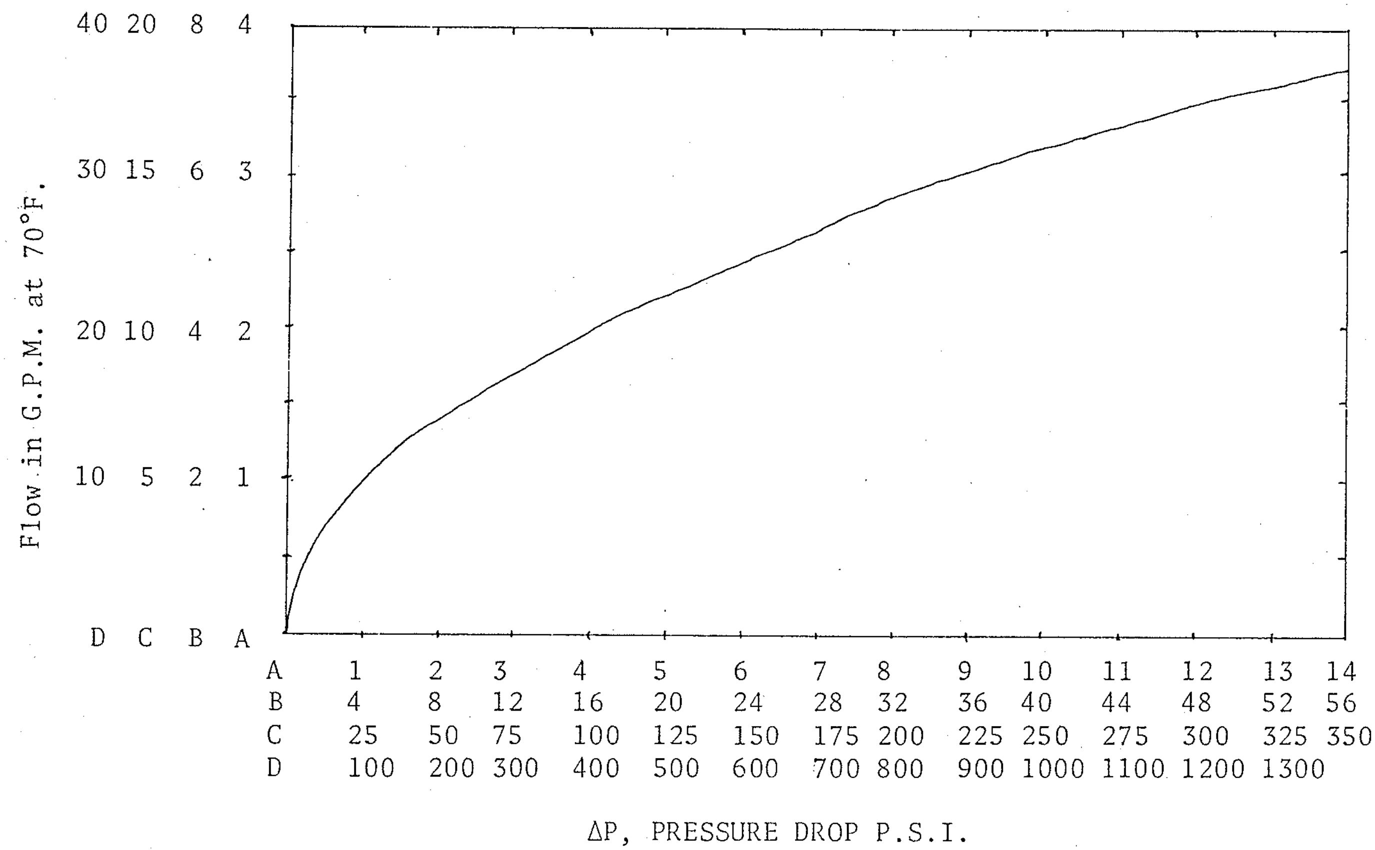


Fig. 11 Water Flow Chart For Valve With C_V Factor = 1

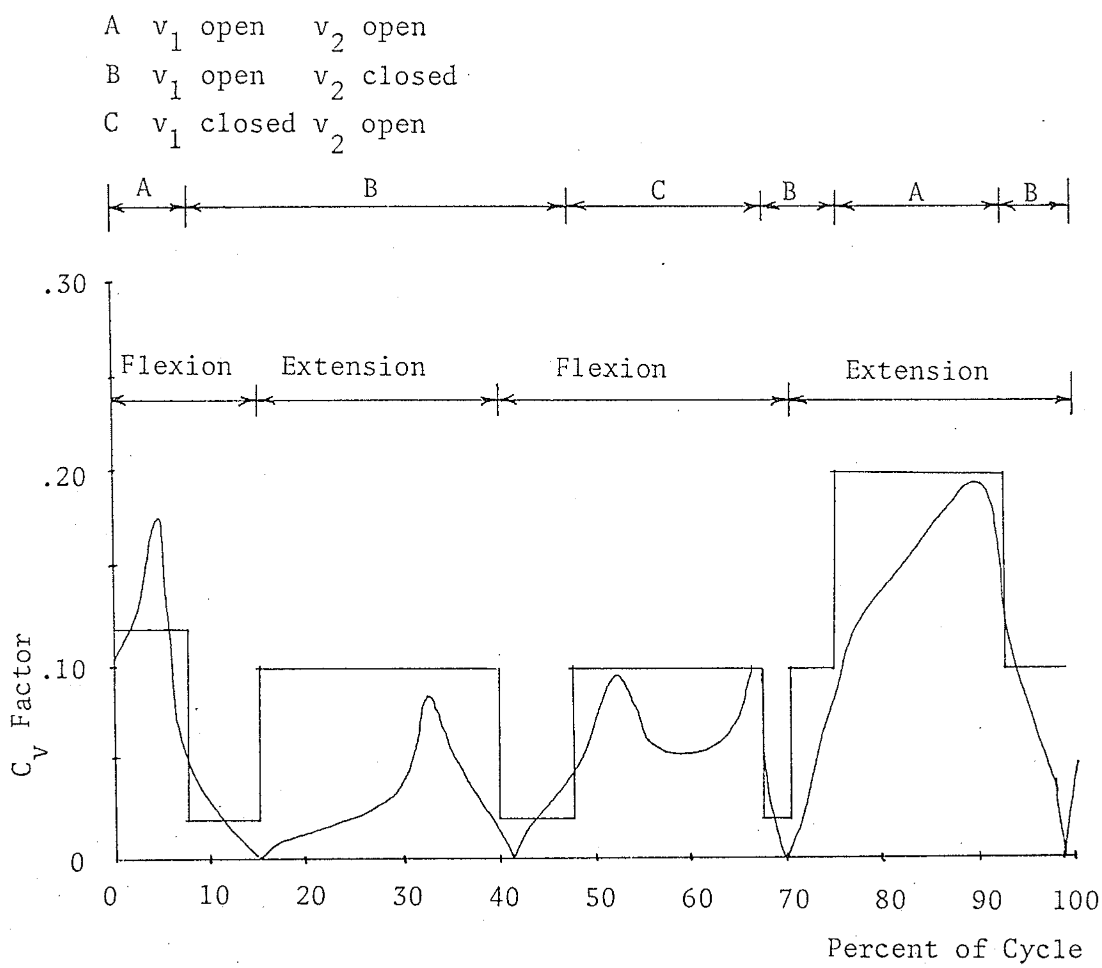


Fig. 12 C_v Factor vs. Percent Walking Cycle and The C_v Curve Fit braking is required. A second valve with a C_v factor equal to 0.1 was then chosen to fit the remaining parts of the curves in flexion. For the sake of convenience, let v_1 represent the valve of $C_v = .02$, and v_2 that of $C_v = .10$. Now, during the initial knee flexion associated with heel strike, 0.0% - 7.5%, by having v_1 and v_2 open, a smooth knee flexion is achieved preparing the leg for the weight bearing process on a partially flexed knee. To complete the initial flexion, 7.5% - 15.0%, v_1 is open and v_2 is closed simulating the yielding lock action of the quadriceps.

This is followed by a weight bearing extension during mid-stance, immediately followed again by a flexion phase at push-off. At this point in the cycle, 40.0% - 47.5%, v_1 is open while v_2 is closed, giving security during double limb support and a somewhat controlled push-off. The next state, 47.5% - 67.5%, consists of v_1 closed and v_2 open, to give a moderate degree of freedom to allow the knee to flex after push-off. Then, between 67.5% - 70%, v_1 is again open while v_2 is closed which brakes or decelerates the upward swinging shank in preparation for the swing phase. It should be noted here, that whenever v_1 acts alone it simulates the quadriceps in either a yielding or a braking action, therefore it could be considered the quadriceps in the mechanical servo-system.

Looking next to extension, the swing through phase is considered first. Here a free forward movement of the leg should occur, requiring a high C_v or low resistance mode. Since opening both v_1 and v_2 cannot accomplish this, an additional device must be incorporated into the system. Also, it should be noted here, that if v_1 and v_2 are closed, the knee is locked to both flexion and extension. A lock to flexion is a very desirable feature; whereas, one to extension is very undesirable. To alleviate both these problems, a check valve was thus inserted into the system, in parallel with the two other bypass valves. Because the check valve closes during flexion, the previous fit is so far unaltered, and because it is open to extension, recovery from a locked condition is now possible. For the best curve fit during swing phase, a check valve with a C_v of .08 was chosen. Hamstring activity shows up at the end of swing phase just prior to heel contact as a higher resistance (v_2 closes) which decelerates the entire lower extremity preparatory to planting the heel for stance phase.

The last consideration is now given to extension during weight bearing. At this point, there is no longer any possibility of changing the hydraulic system as it now stands. Because of the inclusion of the check valve, resistance to extension here is quite low, much lower than necessary. It is thought, though, that it would be better that the resistance is too low rather than too high, and it is hoped that with correct training, the amputee will be able to control the inherent 'snapping' action with his stump, and so, present no great problem.

The valves have now also been completely specified along with the hydraulic system hook-up assembly. (Fig. 13). Among such specifications such as operating pressure ratings (500 psi), C_v factors, and burst pressure ratings (at least 5000 psi), a few other restraints were kept in mind. The system had to be small and light, response times in the order of a few milliseconds, and the valves operational in any position. Two other factors, considered of lesser importance, were quiet operation (no solenoid hammering), and self-lubricating.

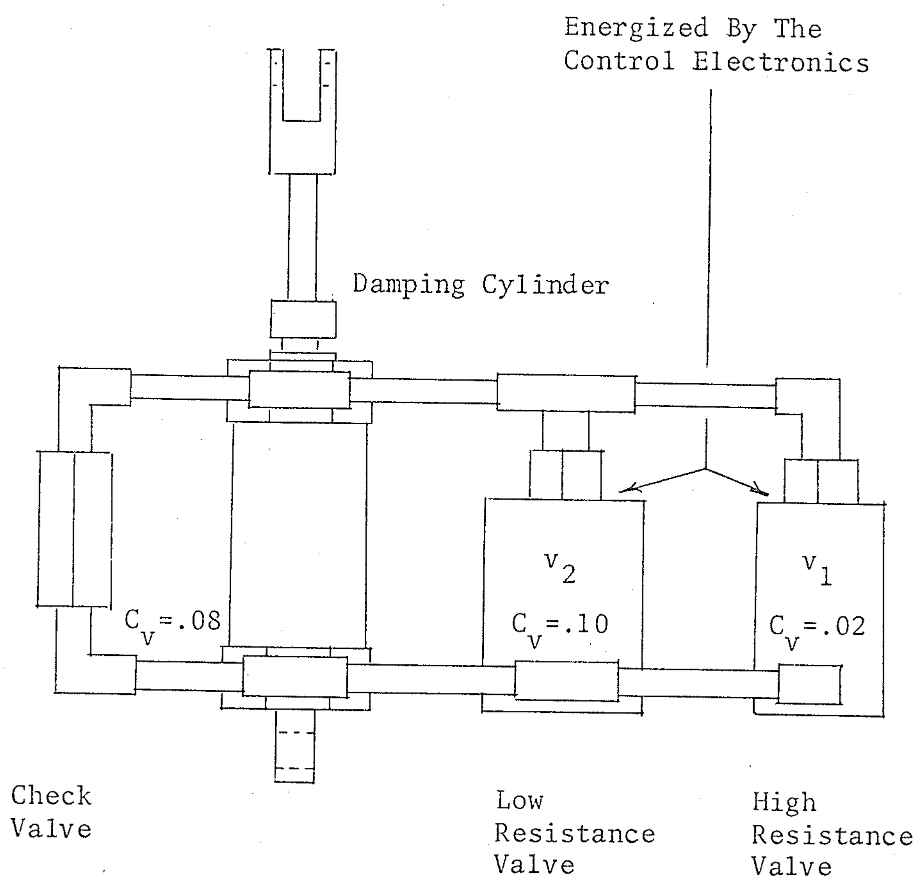


Fig. 13 Assembled Hydraulic System

3.4. Detailed Testing of the Hydraulic System

A mathematical evaluation was next performed to assess the flow behaviour of the designed system. These calculations comprised of a comparison of the flow characteristics of the cylinder, with those allowed by the valves if operated or switched in a manner described on Fig. 12. The cylinder flow data was already calculated by the computer in order to obtain conductance values. This new flow was found by multiplying the flow data fed into the computer from the $C_v=1$ curve (Fig. 11) by the C_v values as portrayed on Fig. 12. The comparison was done graphically and is shown in Fig. 14.

The flow, during the initial flexion, correlates very well, whereas the new flow during the mid-stance extension is excessively larger, as expected. As suggested previously, this excess flow could be made to match the proposed flow more accurately via amputee stump control of the prosthesis. The flows during the flexion phase, up to toe-off, also correlate quite well, but stray somewhat at heel rise, allowing for a slightly faster flow and upward swing. This upward swinging shank, though, is constantly under control in that the upward swing is never excessive, and if in the future sometime, an energy storing spring were to be introduced into the system, this slight discrepency might never be noticed. The curves then follow each other through the swing phase proving that the system is performing as it should, according to the flows.

The valves and cylinder were then purchased, and assembled according to Fig. 13. The following test performed on this hydraulic system, was to compare the actual forces in the cylinder with those the system should exhibit, as predicted by the computer in order to assess

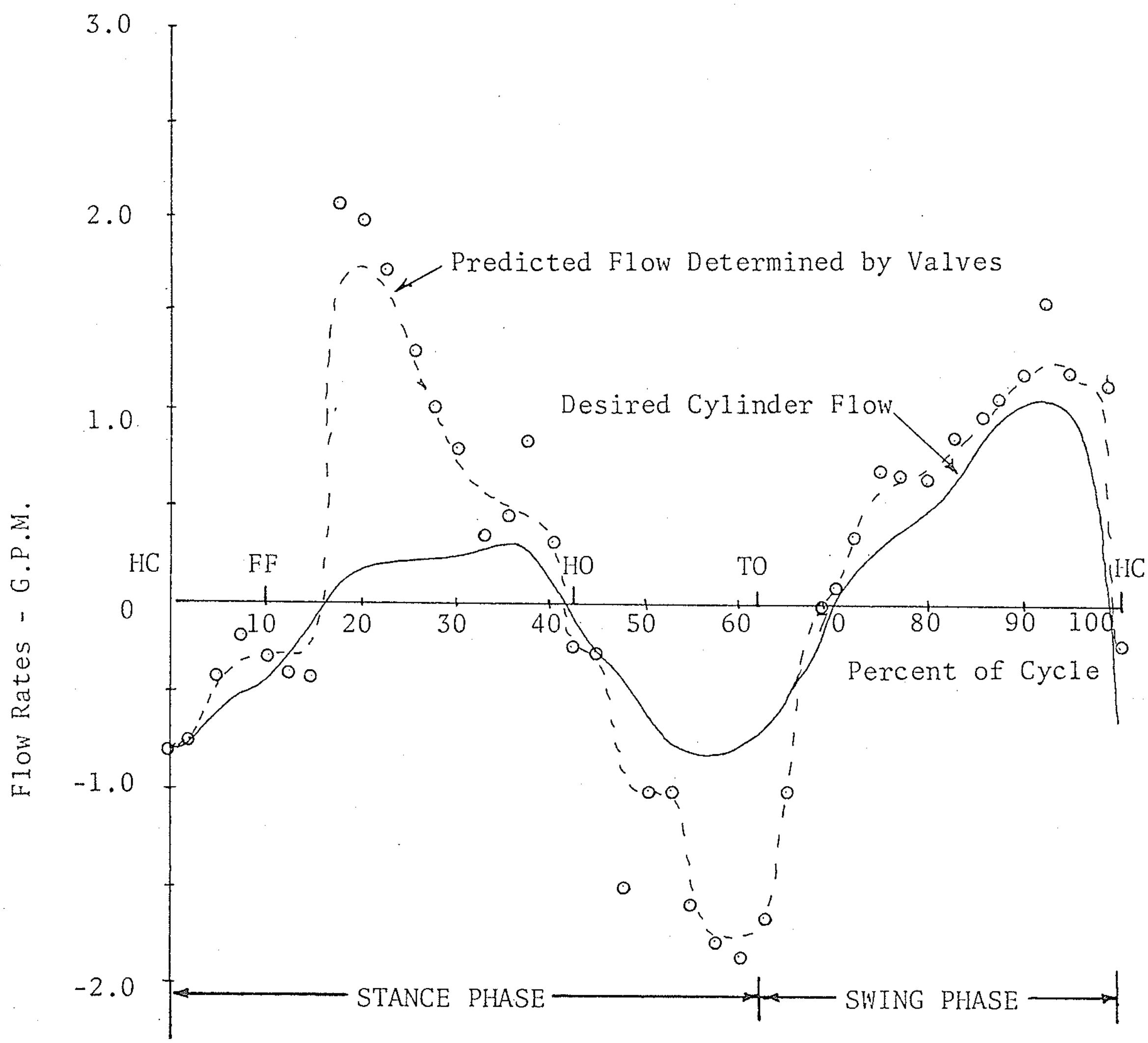
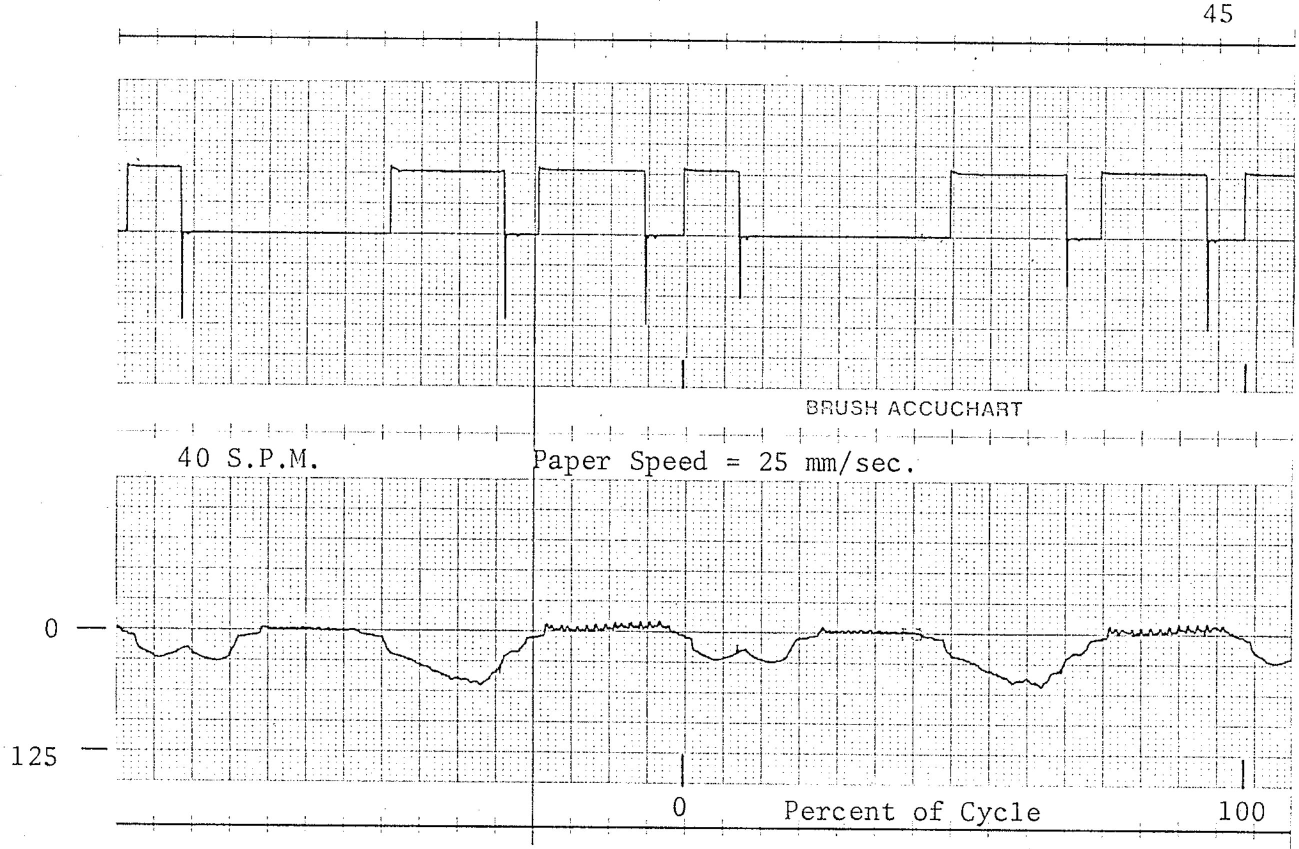


Fig. 14 A Flow Rate Comparison vs. Percent Walking Cycle

the force behavior of the system. This was done via a test jig, described in Appendix 3, which caused the piston in the cylinder to function as if it were in a prosthetic leg, going through a number of walking cycles. A variable speed constant torque D.C. motor, through a system of pulleys, turned a cam at speeds corresponding to different walking speeds. The cam was designed (with the aid of the computer) to actuate the damping cylinder, with the use of a lever riding it, on the same shaft as the driving cam, a second cam was attached which activated microswitches, at the correct times during one revolution (which constitutes one walking cycle), energizing the correct valves(s). Next, strain gages were mounted on the piston rod of the damping cylinder and then connected to a Tektronix Type Q Unit. The Type Q Unit then converted strain data into electrical signals which were calibrated to represent forces, and then monitored and recorded.

The first test consisted of running the system at 40,60,80,90 SPM and recording the forces at each speed. In order to compare basic curve shapes and magnitudes of different tests performed on the system, a reference time base was established by also recording the switching voltage of v_2 (which was a normally closed valve) simultaneously with the forces. V_2 was selected for this function because v_1 (which was a normally open valve) switched only once during the walking cycle, and it corresponded exactly with one of the v_2 switches. Zero time, or starting point, was heel contact, which could be identified as the rising edge of the narrowest square wave (and also as the first zero crossing of the force curve after the double square wave). The force curves are shown in Fig. 15.

Since locomotion data exists only for normal walking speeds of



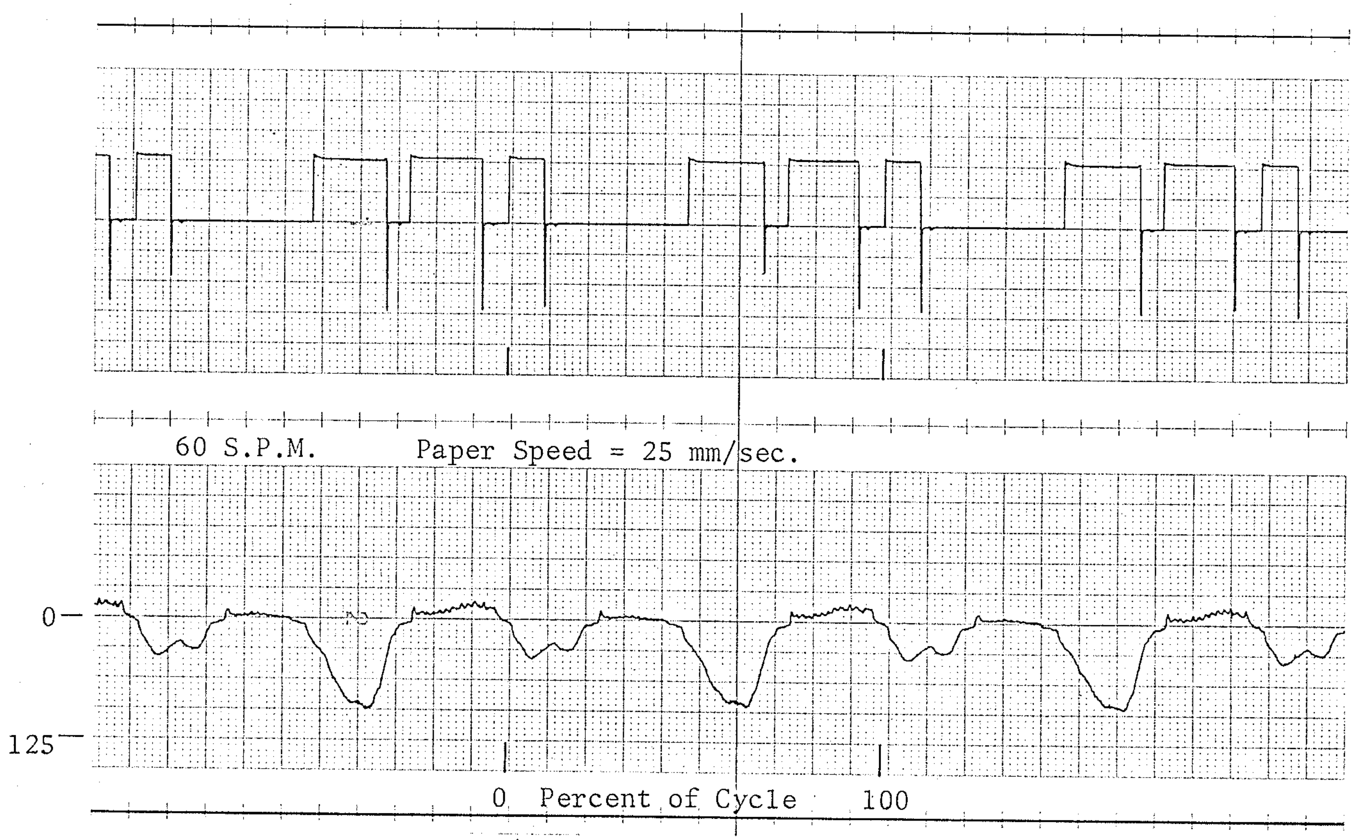
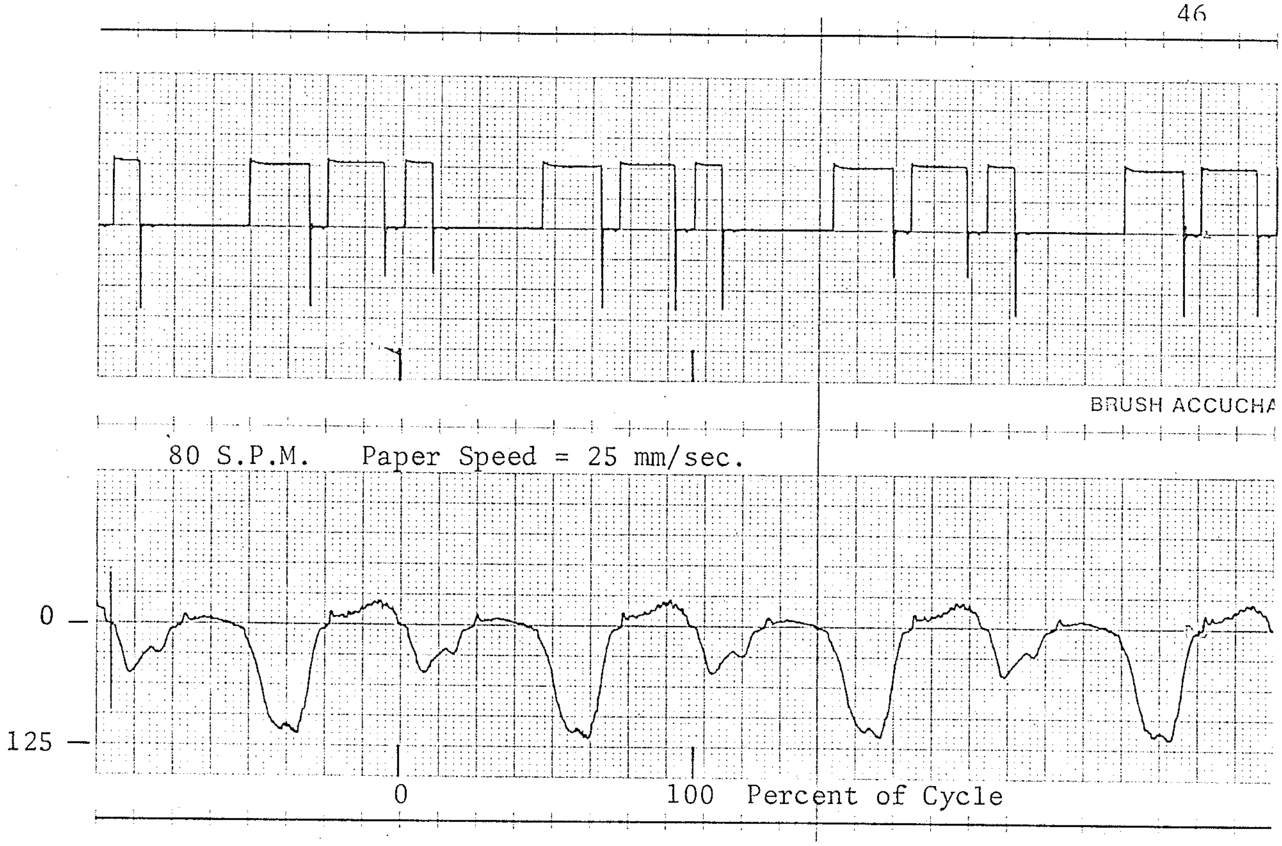


Fig. 15 Force in Pounds vs. Percent Walking Cycle



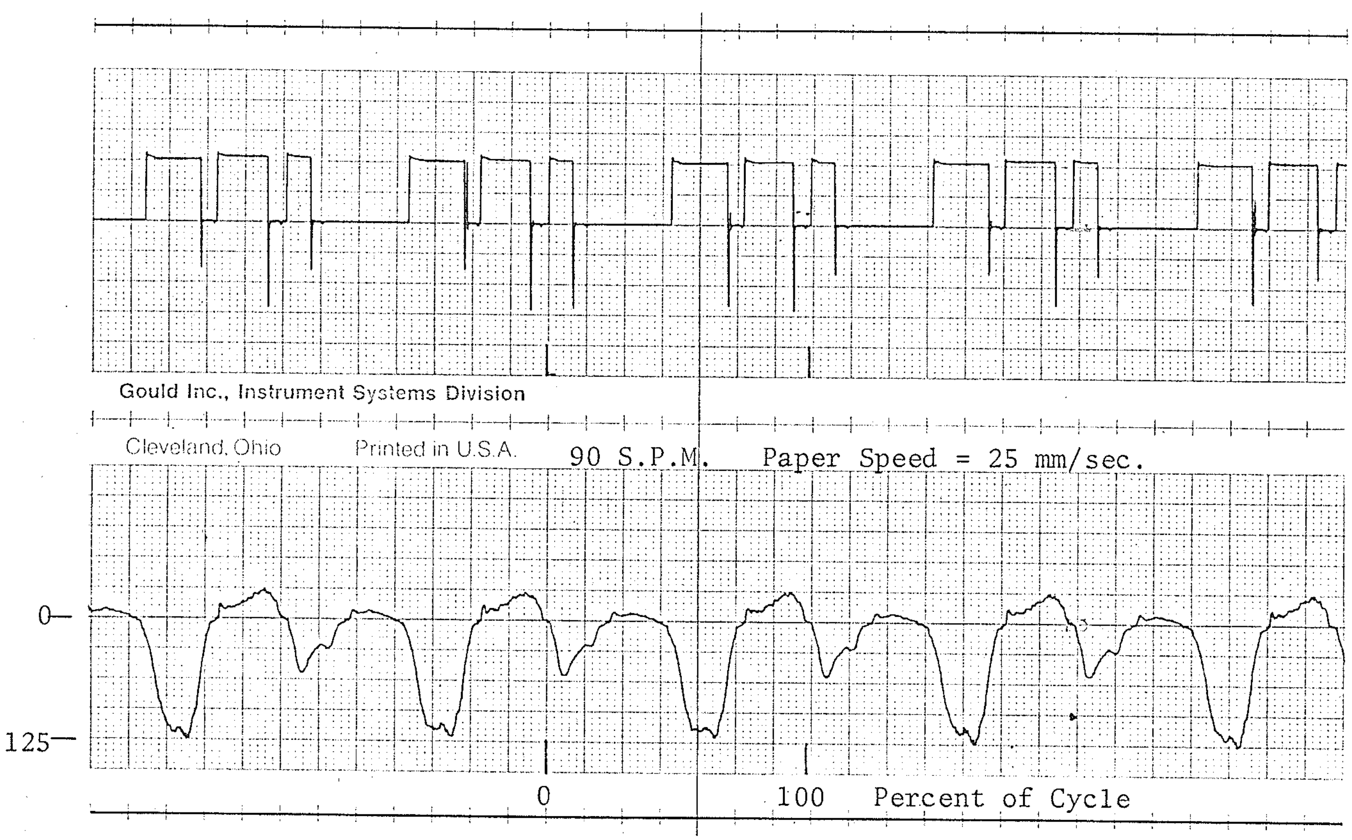


Fig. 15 cont'd. Force in Pounds vs. Percent Walking Cycle about 90 SPM, the computor could only predict forces for that speed. Actual forces were then compared graphically to predicted forces for one complete walking cycle at 90 SPM (Fig. 16). The outstanding difference immediately observed was that the predicted peak of some 650 lb. just after heel contact was missing in the actual curve, although the rest of the curve matched quite well. It was initially thought that this inaccuracy was caused by the fact that apparently, there was just not enough resistance to flow at heel contact to give such a high supporting force. Another factor considered, was that when the cylinder motion reversed, a dead zone or hysteresis was noticed. This was due to a loose connection between the driving lever and cylinder. Therefore, just when a large supporting force should be noticed (for a relatively very short period of time), the system is passing through a dead zone.

On this diagram (Fig. 16), the physiological force versus percent walking cycle is also drawn. This is the force, earlier calculated by the computer, that the cylinder would have to produce in order to create the correct moment about the knee (as seen in normals). Since this physiological moment is applied by purely active muscular activity, and the hydraulic system used to simulate this moment is purely passive, it is observed on this diagram that this simulation is quite impossible. This is so because the hydraulic system needs a nonzero flow in one direction to produce a desired force in the opposite direction. Just after heel contact, the flow is in a negative direction, producing a resisting force in a negative direction, up to and including the peak. Then, a continuing force in the negative direction must be supplied by a flow in the positive direction, which is the impossibility of this passive network. It can therefore only resist or brake, not extend,

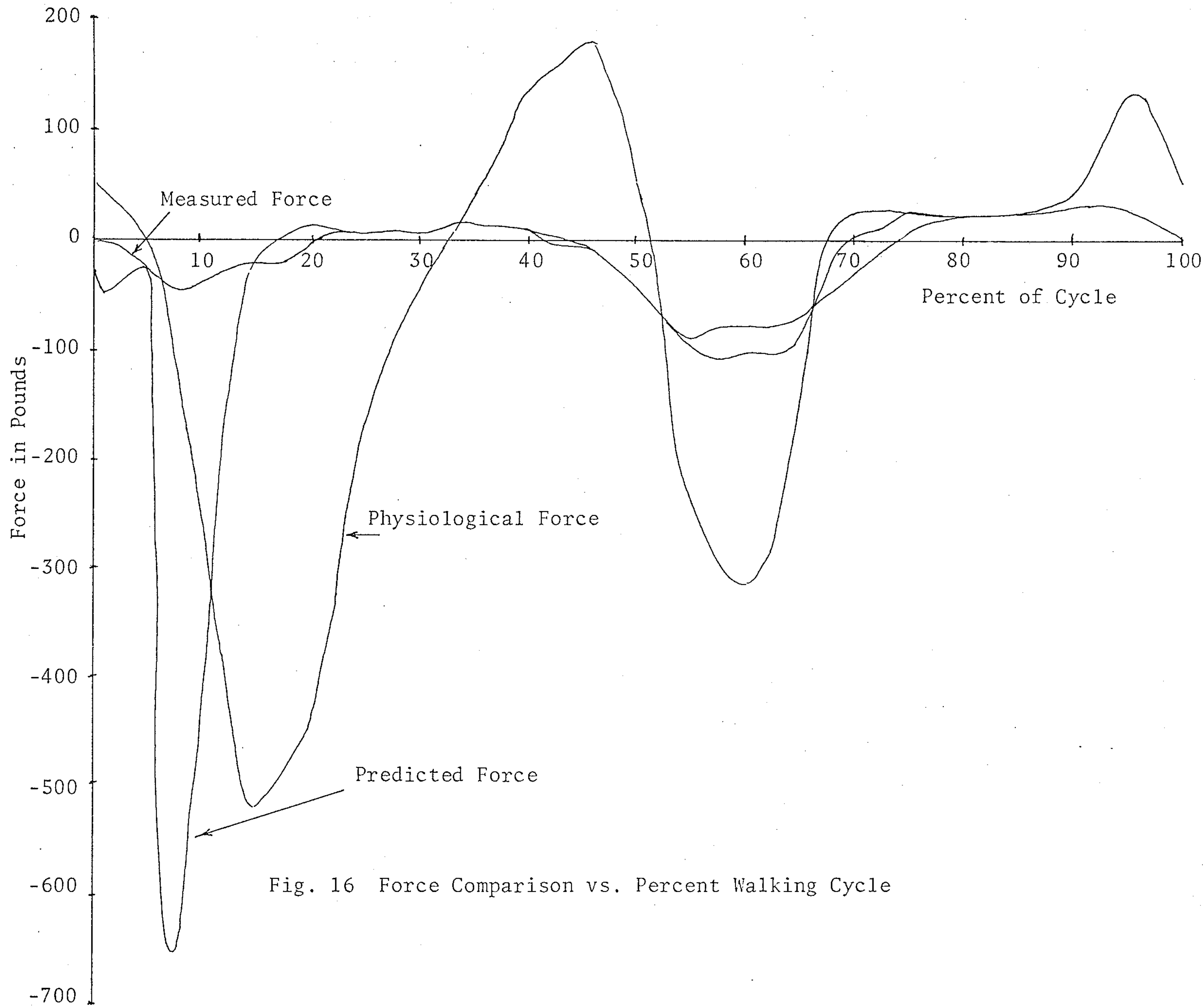


Fig. 16 Force Comparison vs. Percent Walking Cycle

although control extension to a degree.

The hydraulic system can only supply a partial force curve fit, and still needs an active input, ie., stump or hip flexion or extension. Therefore, the objective kept in mind in the design of the hydraulic system, was to replace as much function where possible in order to decrease to a minimum the active input via stump activity. This would, of course, automatically reduce physical control energy.

Returning to the system correction, the fit between the driving lever and cylinder was improved, and valve switching at heel contact was eliminated. Force curves were again recorded at 40,60,80,90 SPM (Fig. 17). On these curves, the zero reference would be the first zero crossing of the force curve after the double square wave. Comparing these results to previous ones, a doubling of the force at heel contact was observed, whereas the rest of the curve, naturally, remained unchanged. However, when a new predicted force curve was calculated, it was discovered that it had also doubled. Therefore, when these two curves were plotted against each other (Fig. 18), it was seen there was still a considerable discrepency to be explained. The actual force curve, although, now contained a far greater contribution to support and knee control during initial flexion, which reduced the necessity of stump activity.

To further investigate the discrepency, two new possibilities for error were considered. The first was a change in the angular velocity of the driving shaft. The constant speed motor, audibly worked harder at two distinct parts of the walking cycle; both corresponding to flexion (one at heel contact and the other at push-off). It was thought that at these two points, if the angular velocity was showing down, the piston velocity and, therefore, flow was reduced, producing a

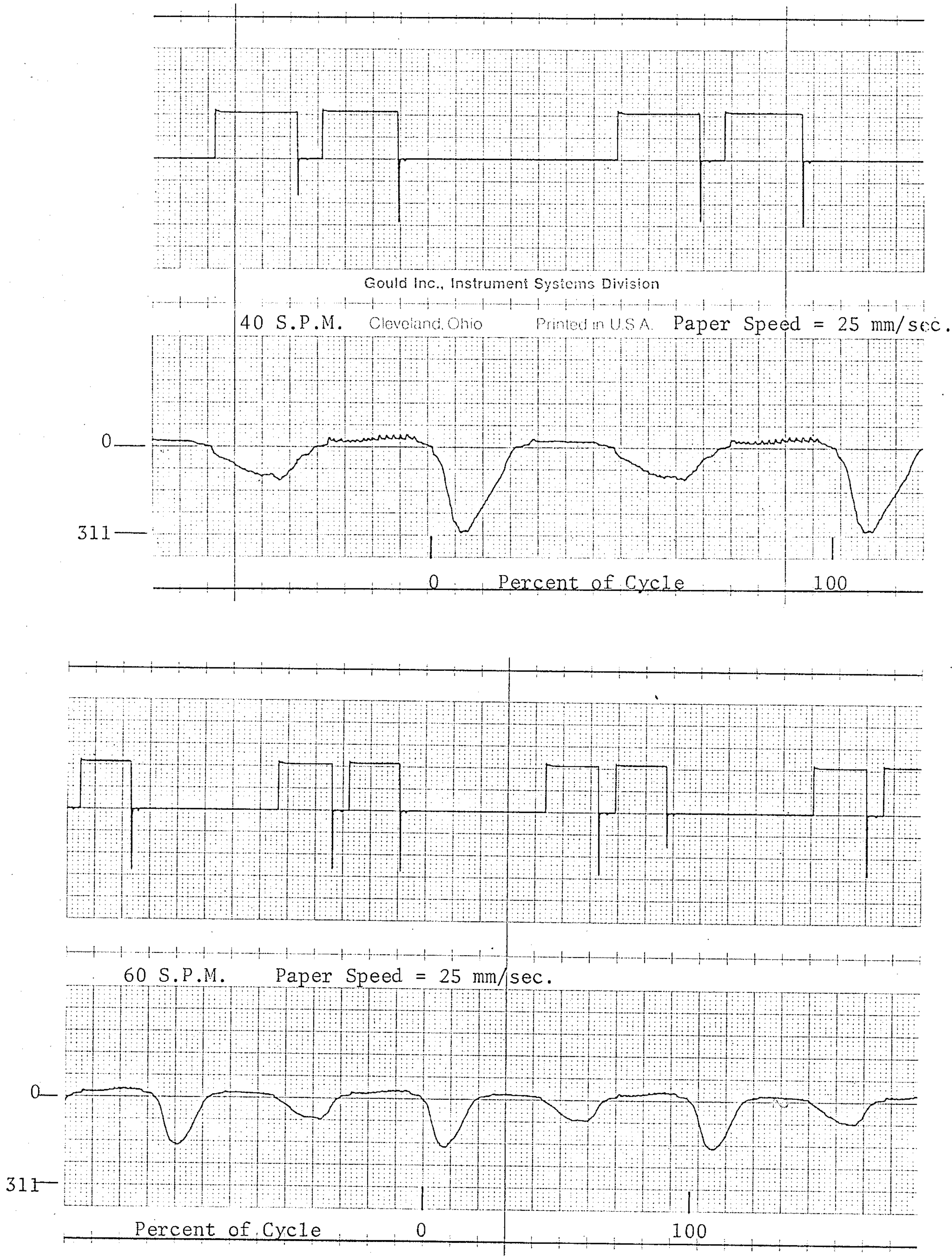
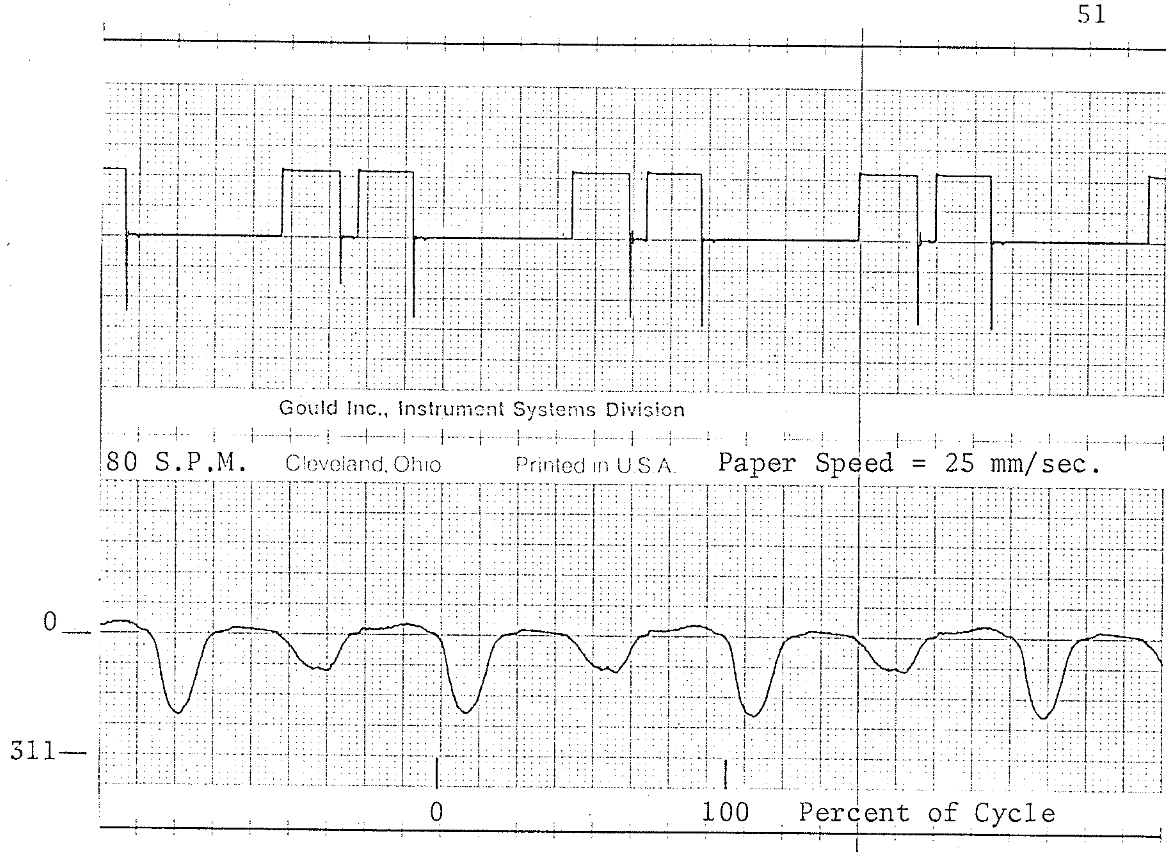


Fig. 17 Force in Pounds vs. Percent Walking Cycle



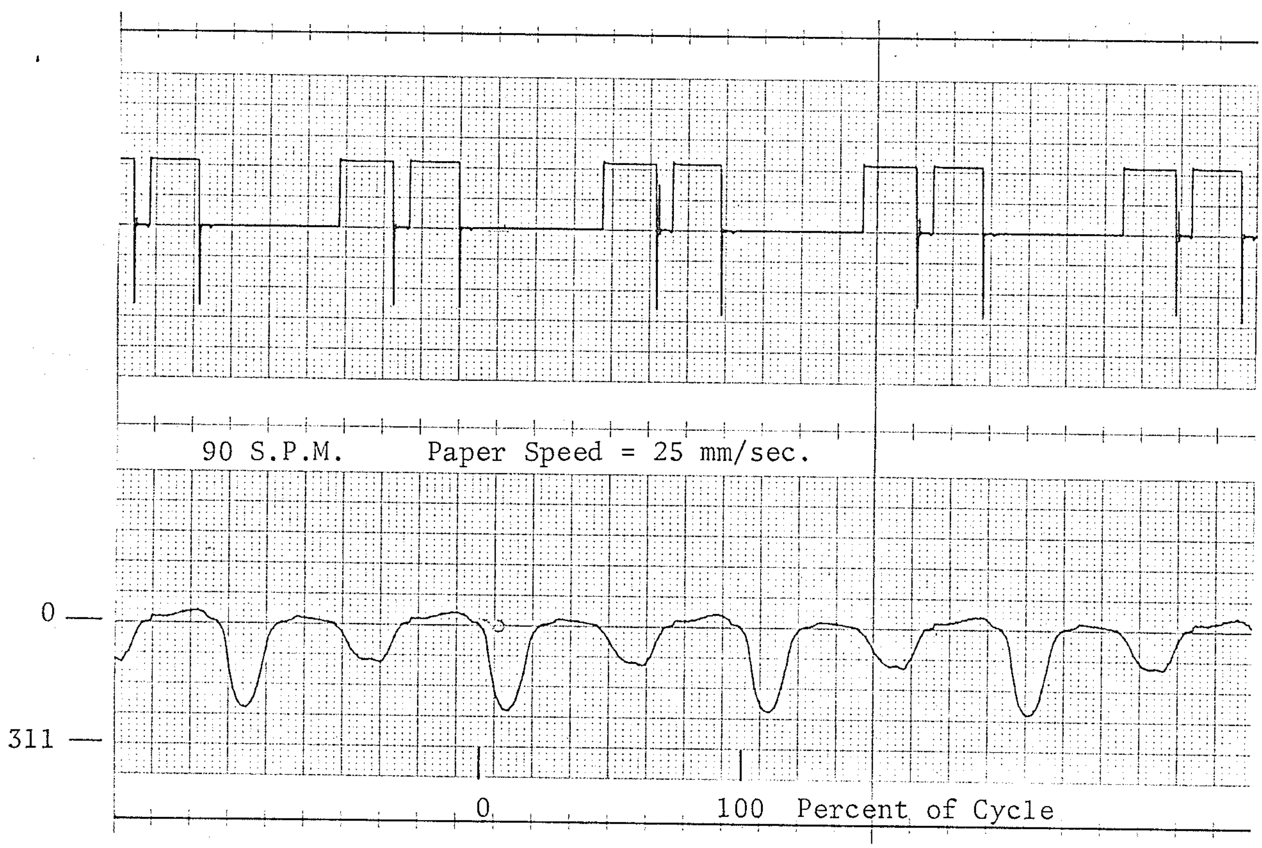


Fig. 17 cont'd Force in Pounds vs. Percent Walking Cycle

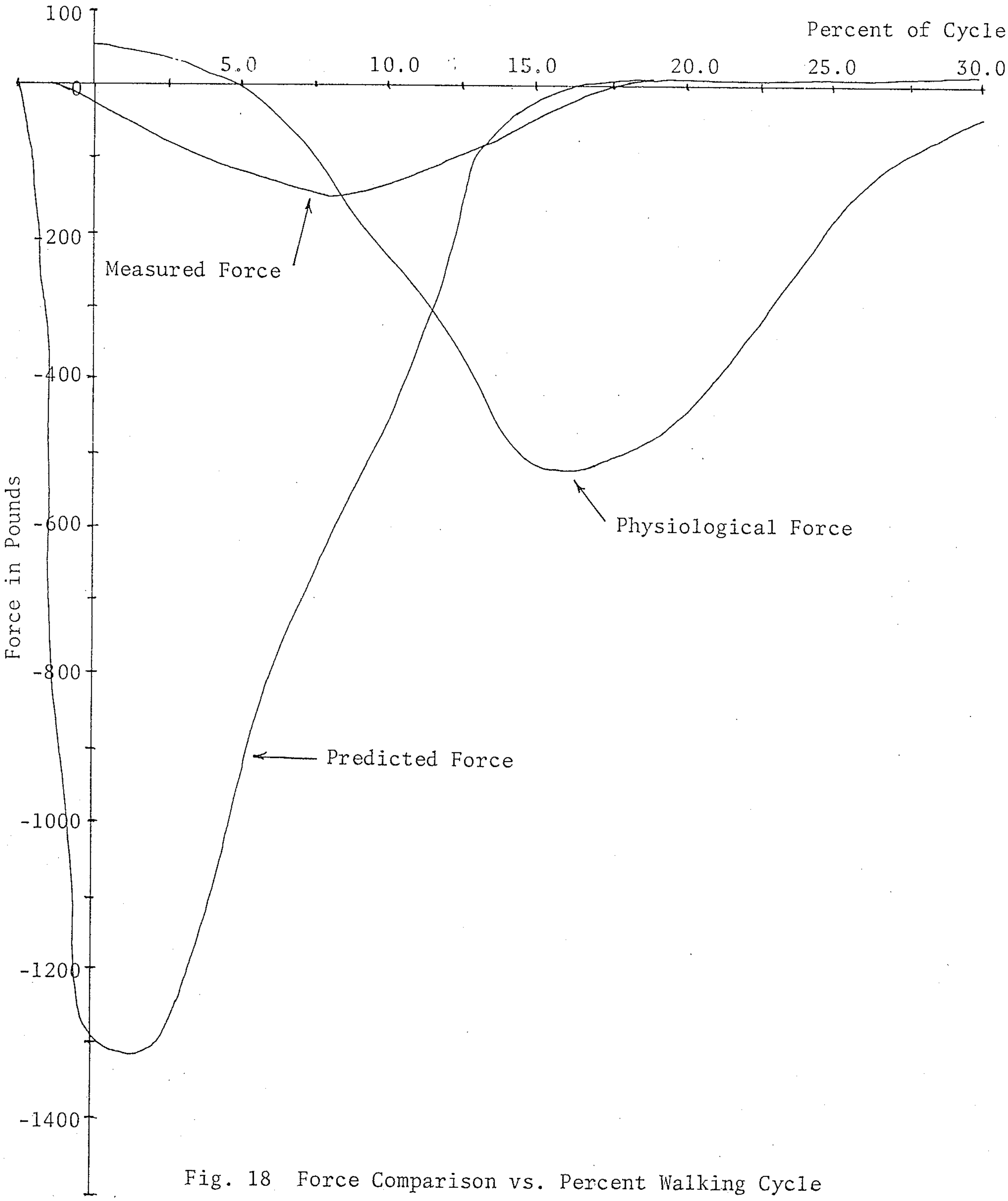


Fig. 18 Force Comparison vs. Percent Walking Cycle

force that was lower than desired. A tachometer was connected to the drive shaft, which was turned at 90 SPM, and a recording was made (Fig. 19). It was then noticed that there were indeed two segments in the cycle with a marked decrease in shaft speed. The greatest decrease occured at push-off, but because the resistance to flow was quite low at this period of the walking cycle, the forces still correlated well. The other decrease in angular velocity was at heel contact, and because resistance to flow here was high, it was suspected that there was a distinct force error due to the reduced flow. Because, however, the piston velocity and, therefore, flow varied directly with the variation of rotation (as derived in Appendix 4), and because the reduced rotation was not very large, it was concluded that this was not the major factor for the discrepency. (The existance of this reduced RPM was also suggested by the slight delay between practical and theoretic force curves).

The second possibility considered at this time, was the fact that immersed in the hydraulic fluid were tiny air bubbles which were being compressed by the high internal forces and thereby reduced the total external force contributed by the system. In an attempt to eliminate this problem, a standpipe or accumulator was incorporated into the system, which was then bled sufficiently of air bubbles, and so eliminate their effects. The system was then retested with the results shown in Fig. 20.

The force at heel contact again doubled to approximately 300 lb. The resultant force was now getting very close to the physiological force curve, even though there was still an extremely large error between actual and expected values (Fig. 21). At this point in testing

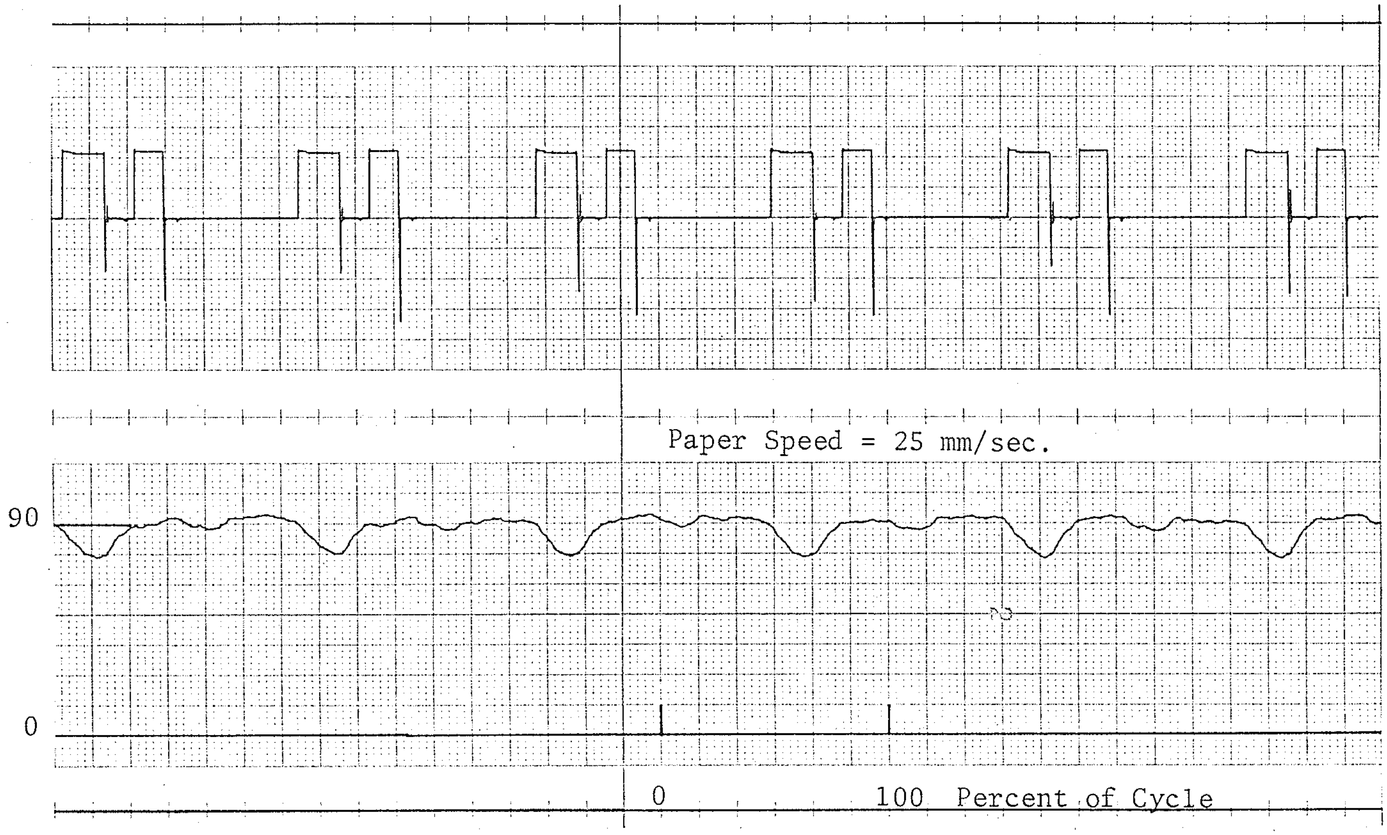


Fig. 19 Graph Showing Changes in Angular Velocity During One Cycle at 90 RPM. (Tachometer Output)

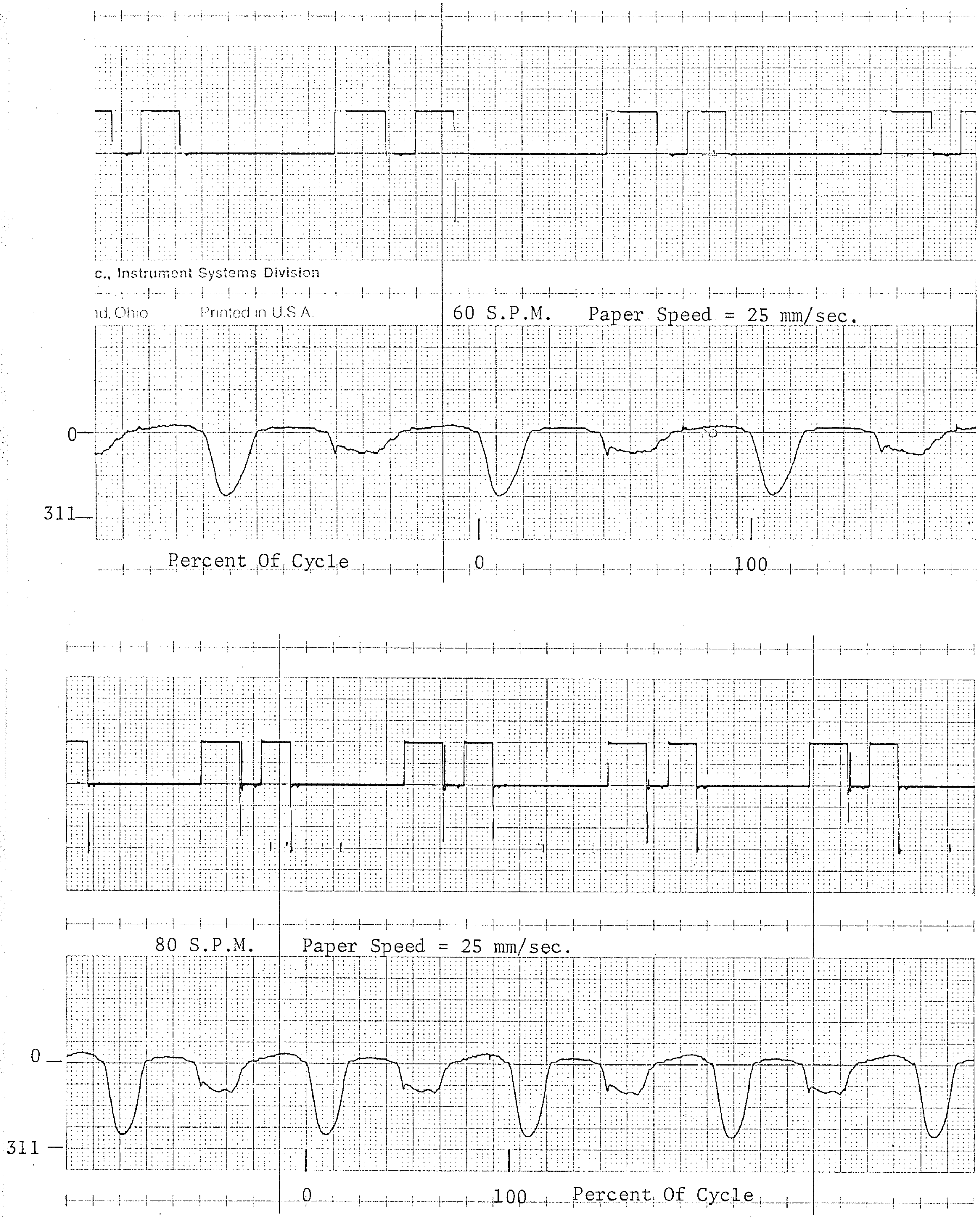


Fig. 20 Force in Pounds vs. Percent Walking Cycle

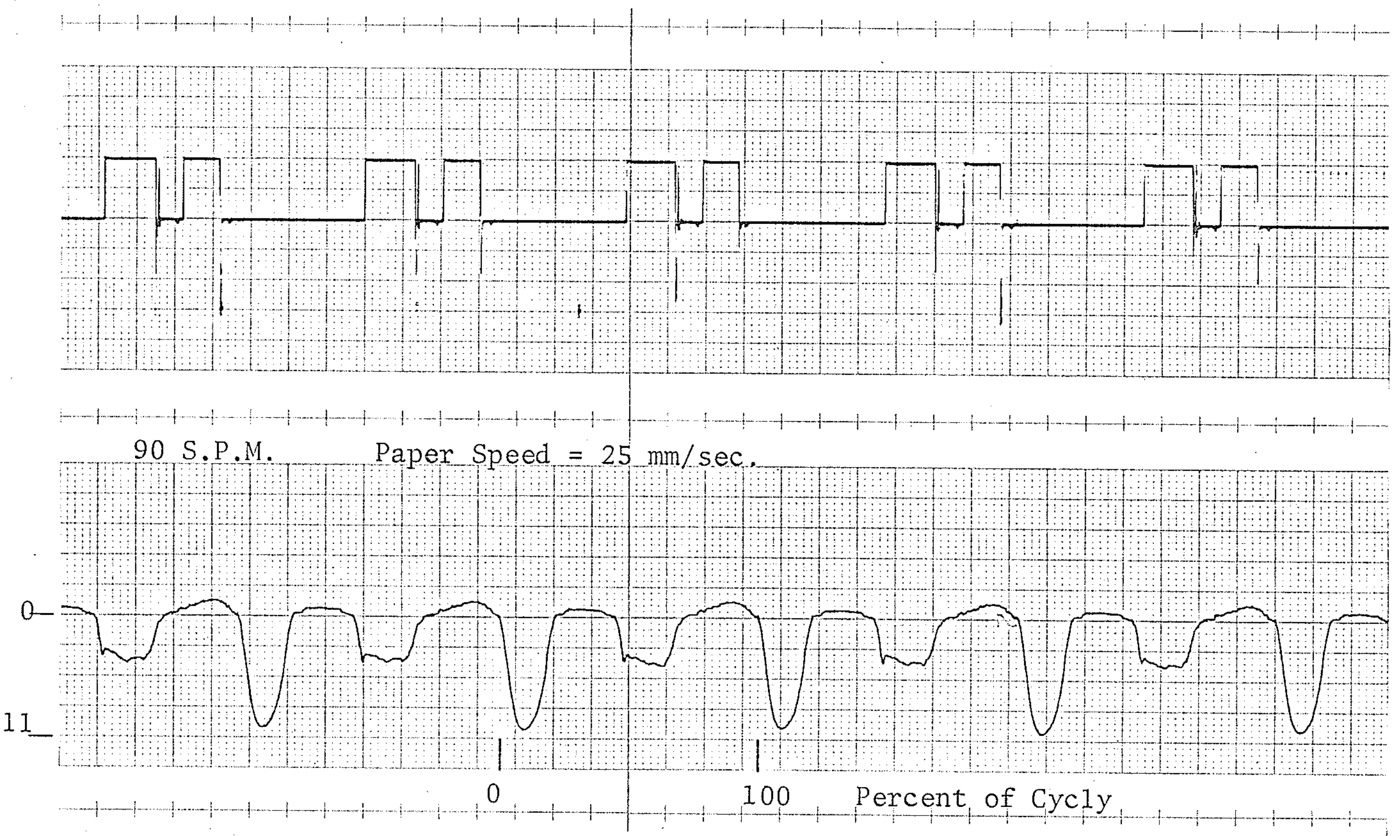


Fig. 20 cont'd Force in Pounds vs. Percent Walking Cycle

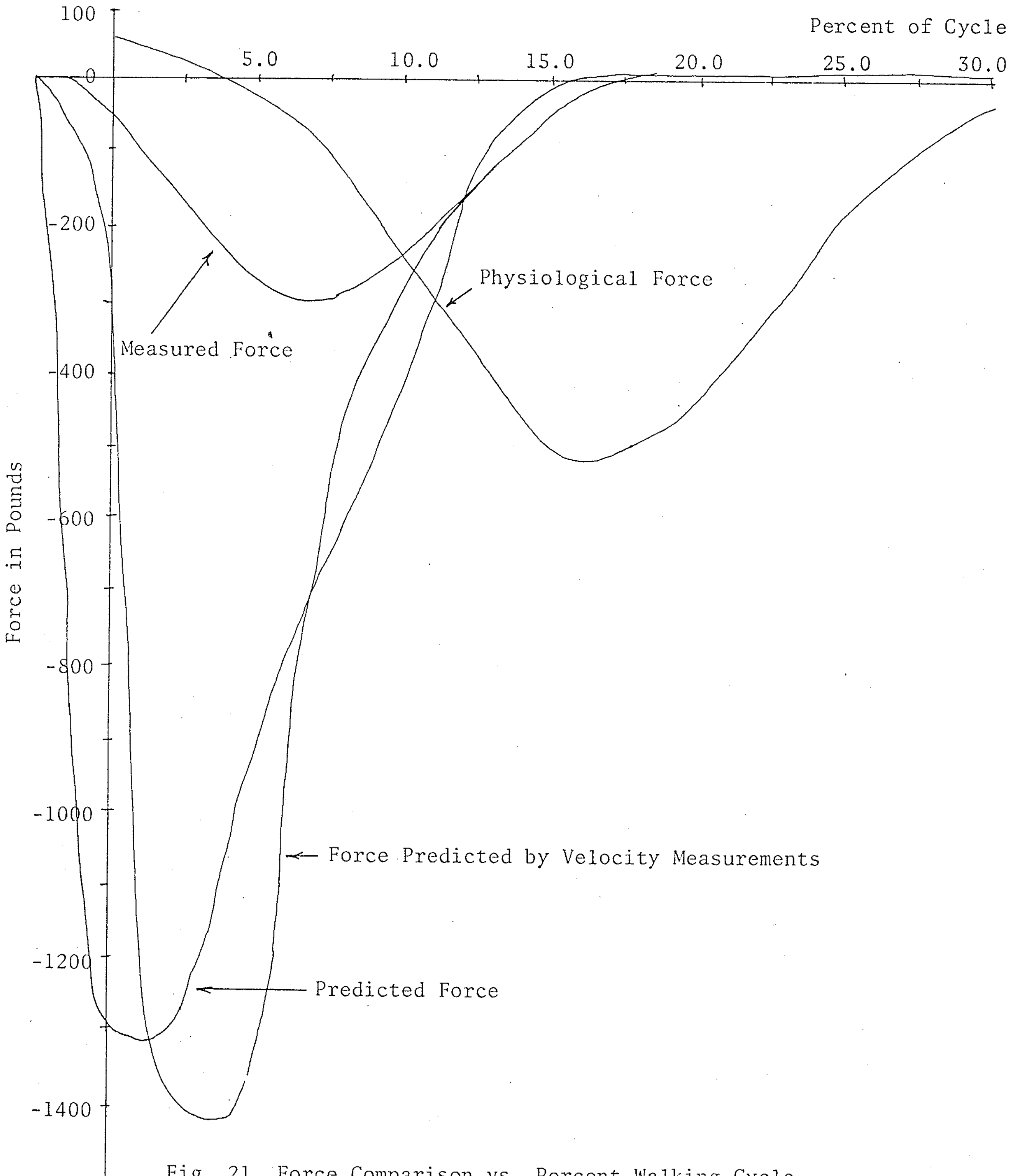


Fig. 21 Force Comparison vs. Percent Walking Cycle

it was noticed, that there was excessive bending of the motion arm or driving lever about the cylinder. On further testing, it was discovered, that by locking the system (closing both bypass valves), and producing a 165.24 lb force at the end of the lever, there was a displacement of 0.4 inches from the unloaded position, whereas cylinder depression was unnoticable. Therefore it was postulated, that the major contribution to the force curve discrepancy was due to the driving lever absorbing the peak force for bending purposes.

As an additional test, a linear potentiometer was attached to the end of the driving arm, which was operated at 90 SPM, and displacements of the lever were recorded (Fig. 22). From this recording, slope or velocity measurements were taken and recorded. These readings were next converted into a flow, through a valve with a C_v of one, and then a force corresponding to that flow using the force-flow diagram (Fig. 11). The resultant force readings, which now represent the force that the cylinder would exhibit if it were moving at the velocity of the end of the lever, were graphed on the previous force curves (Fig. 21). This new curve, when compared with the predicted force curve, matched almost identically, which then proved the postulate, that the bending phenomenon of the lever caused the errors in the force curves. Since there is no lever arm in the prosthesis it was further postulated then, that the hydraulic cylinder would develope the theoretic force curve, if moved in the predescribed manner.

The very small differences in the curves could then be explained by the decrease in angular velocity (which could not be corrected), the non-accurate curve fit on the C_v graph, and the fact that it was quite possible that in the presence of the high pressures developed in the

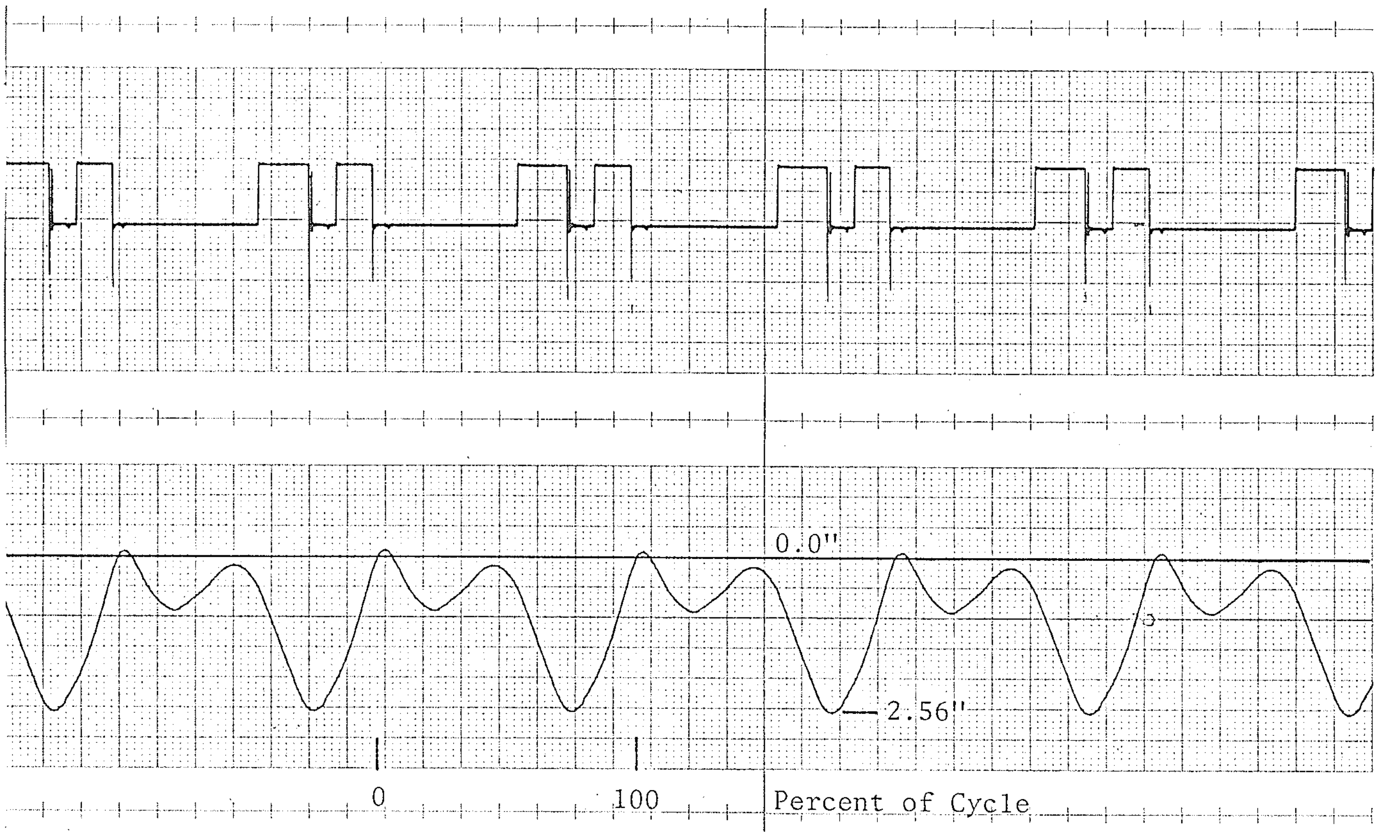


Fig. 22 Graph of Displacements of the Driving Lever End vs. Percent Walking Cycle Used To Determine Linear Velocities Created by Cam. (Output of Linear Potentiometer)

system, that the tubing could be conceivably stretching or bulging.

Up to now, the hydraulic system has demanded a 'four-state' control system ie. four possible levels of resistance available during one walking cycle. Three discrete levels of muscle contraction are therefore also implied (one state corresponding to rest). It was in anticipation of the possibility that this might be a too difficult task to expect of residual thigh muscles in the stump, that another test was performed to probe the possibility of a 'three state' control system. The difference between the modes where both valves, v_1 and v_2, are open and where v_1 is closed and v_2 is open, is very slight as can be seen on Fig. 12. Eliminating the mode where v_1 is closed and v_2 is open, and substituting the mode with both valves open would convert the four state control system to a three state. The system was again tested with this modification (Fig. 23).

Comparing these results to those of Fig. 20, there was a reduced force noticeable during the second flexion mode (smaller negative 'bump') which is more predominant at the lower speeds. At the higher speeds, however, this error is less than 15% and therefore the three state control system was considered acceptable as an alternative to the four state control system, if necessary.

Other tests performed (but not recorded) were to discover if there existed any significant changes to force wave shapes and magnitudes if valve switchings were slightly premature or delayed. Again, no noticable differences were present and it was decided that the system was ready to be used. The hydraulic system was then fitted into the prosthesis in preparation for amputee testing.

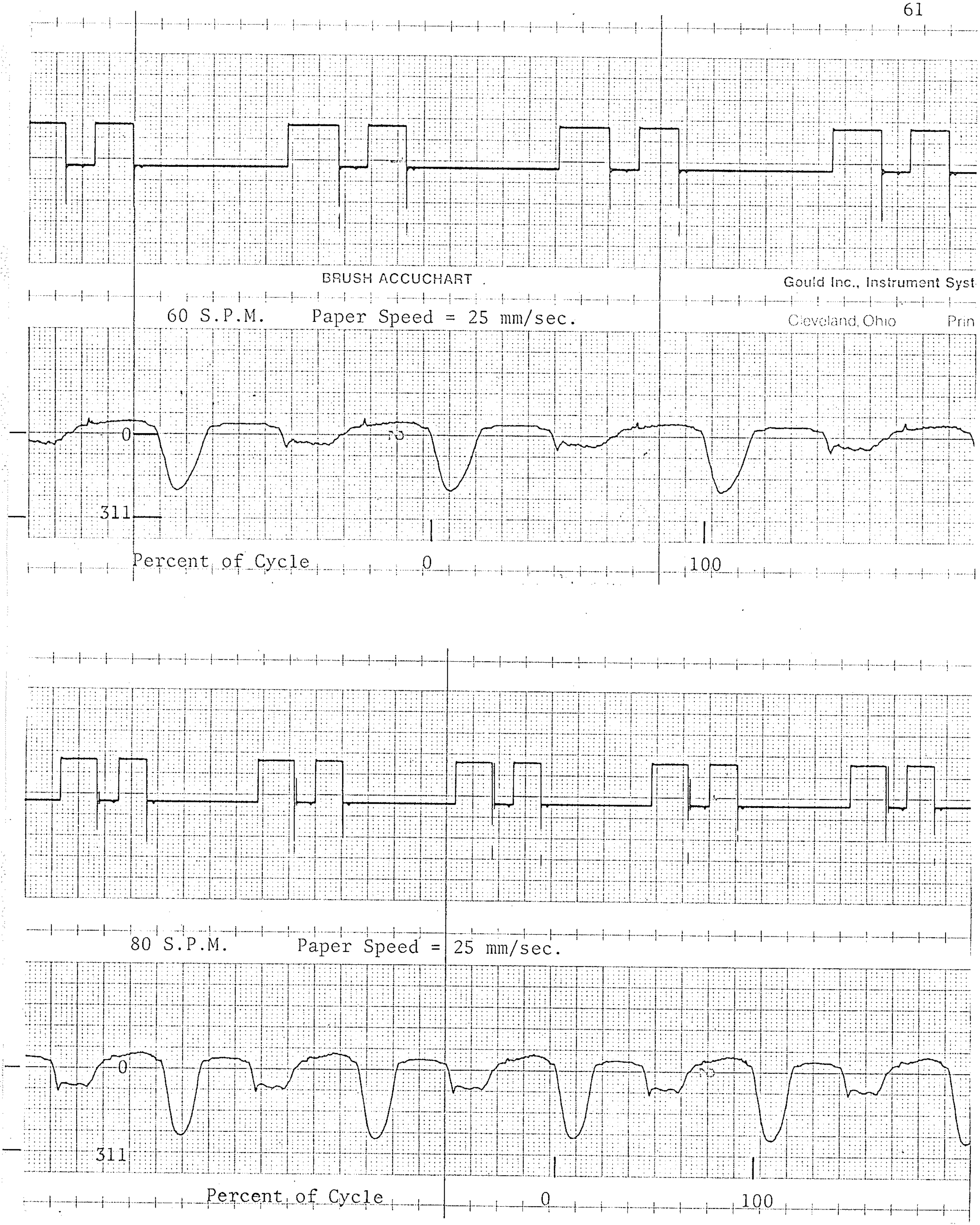


Fig. 23 Force in Pounds vs. Percent Walking Cycle

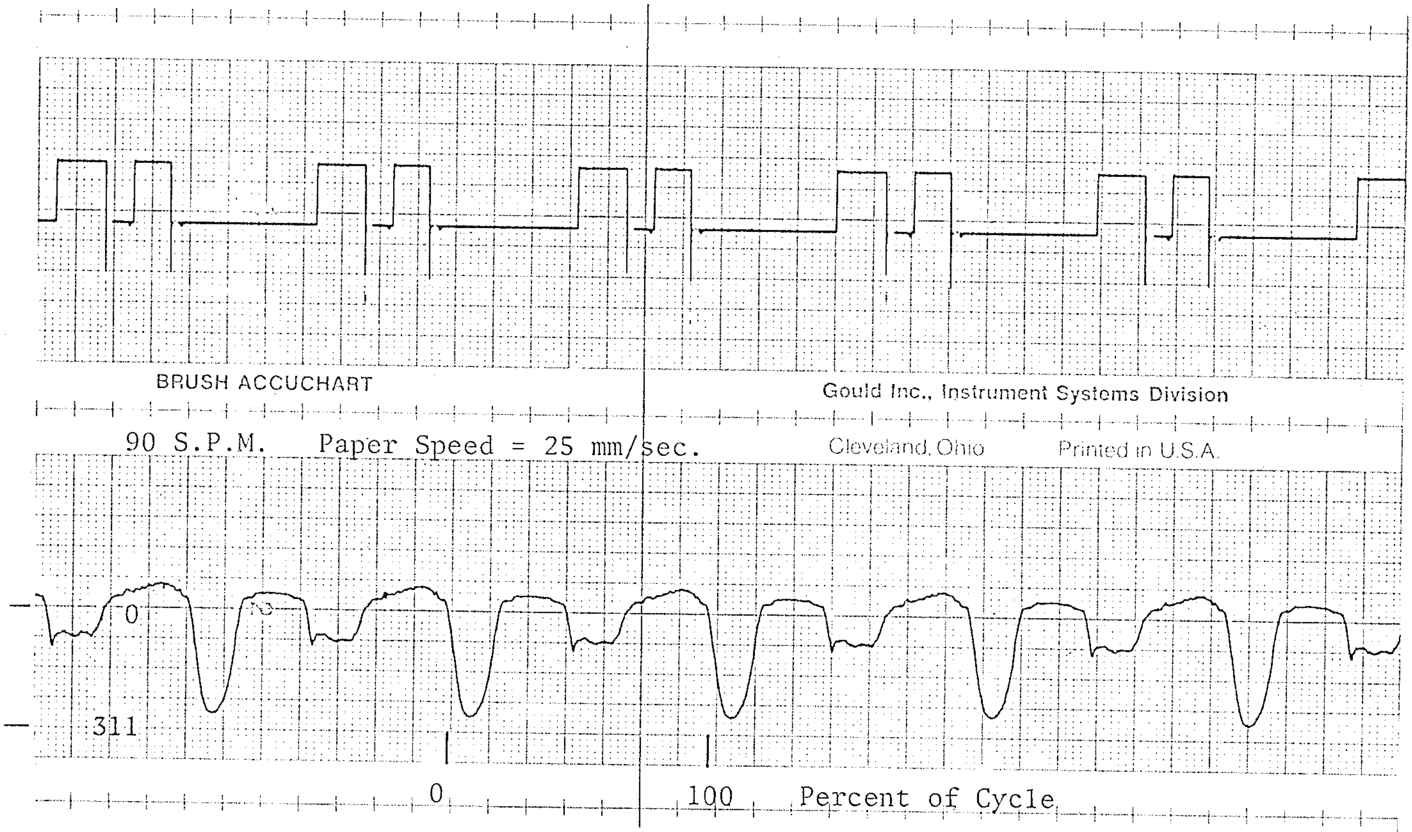


Fig. 23 cont'd Force in Pounds vs. Percent Walking Cycle

CHAPTER IV

DESIGN OF THE EMG CONTROL ELECTRONICS

4.1. Introduction

The EMG control electronics consists of a circuit which uses electromyographic (EMG) signals as an input, and depending on the level or strength of contraction, establishes the desired resistance to flow. Because EMG activity is characterized by AC bursts in the order of a millivolt (peak), what is needed is a high gain amplifier and envelope detector, which convert the electromyographic signal into a DC level, and have a set of three comparators with different reference voltages to sense this level. The outputs of the comparators activate a logic circuit which in turn energizes the correct valve combination.

Included in the design of this four state control system is a provision for an easy conversion to a three state control system, in case the amputee experiences too much difficulty in controlling so many levels.

Before proceeding with the discussion of the design, the logic of operation of the four states must be considered. Keeping the functional criteria of one walking cycle in mind, and observing the myoelectric activity in Fig. 5, it should be noticed that during stance phase there is much quadriceps and hamstring activity to attain stability during weight bearing, whereas during swing phase, there exists minimal muscular activity. Locking would normally occur in a stumbling situation, aiding recovery, and be associated with a high degree of muscular activity. These facts then led to the following as a logical sequence for the different states. The first or rest state would consist of both valves open, for a free and easy swing, and the fourth state would consist of both valves closed, for a lock situation, while states two and three would consist of only one or the other valve open with increasing resistance respectively.

4.2. Amplifier-Envelope Detector

A myoelectric amplifier plus envelope detector, developed by the Winnipeg Shriner's Hospital, was modified (Fig. 24). The modifications included adding a complete front end buffer stage, changing the variable gain potentiometer to give the desired DC voltage level range at the output, and replacing the resistor in the envelope detector with a 10K potentiometer to give the amputee some control over the time constant. This circuit accomplishes the task of yielding a DC output voltage level depending on the strength of the differential myoelectric voltage signal at the input.

The input or buffer stage to this circuit has a unity gain, and provides a high input impedance for the differential muscle signal - common mode 59.4M, differential mode 10M. Since wet electrodes were used corresponding to a skin resistance of 1-10KΩ, these buffer stages provided good isolation. The outputs were then fed into an adder, also of unity gain (for more details see Appendix 5). This stage also includes an input impedance balancing potentiometer, which, when adjusted correctly, establishes a -94db common mode rejection (CMR) at 60 cycles. To stabilize this CMR, the buffer stages are AC coupled to the adder with two matched capacitors, to isolate DC drifts at the output of the buffers.

The adder was AC coupled to the first of a two stage amplifier. The maximum output voltage level was then arbitrarily chosen to be

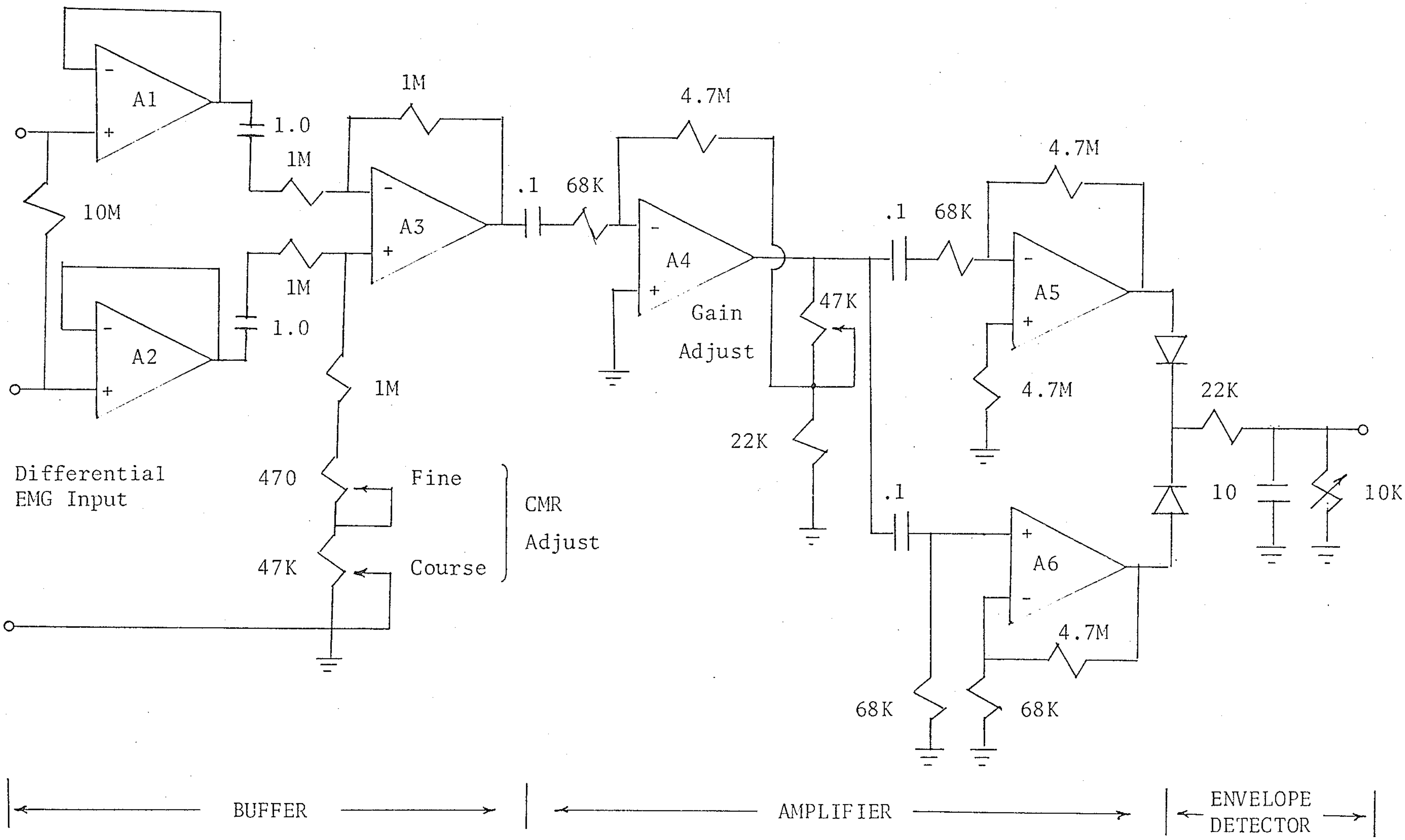


Fig. 24 Schematic of the Buffer, Amplifier, Envelope Detector.

0.5 volts. EMG tests were performed on the amputee and it was found that an EMG voltage of 0.5 millivolts could be produced voluntarily throughout the walking cycle (Fig. 25). Therefore, it was decided that the electronic system, including amplifier and envelope detector, must have an overall gain of at least 1000.

Since the envelope detector possessed a trade-off between gain and response time, an optimum potentiometer setting of 5KΩ was chosen for design purposes. This meant there was a signal attenuation of about 0.2 at the output which implied that the total gain of the two stage amplifier had to be at least 5000.

The first stage of amplification consisted of a variable gain amplifier. As seen in Appendix 5, with the 47K variable gain adjust potentiometer, this stage has an amplification range of -69.2 to -217.0. The second stage of amplification, in accordance with the preceding paragraph, would then have to possess an amplification of at least 70 times, which it was made to. This stage of amplification also employed a technique whereby the signal was passed through an inverting and a non-inverting amplifier, then adding together the positive portions of each. This is a technique of achieving single ended output full wave rectification, which minimized the spacings between peaks, and thus allowed for a more even DC level, after envelope detection.

This part of the system was built and bench tested. A small 0-2 mV. differential AC test signal was produced (Fig. 26) and fed into the input. It became immediately obvious, that this high gain system would require shielding, and the system was then mounted in a metal box. This test signal showed that the system functioned well, and was then used to check the frequency bandwidth of the circuit. The range was

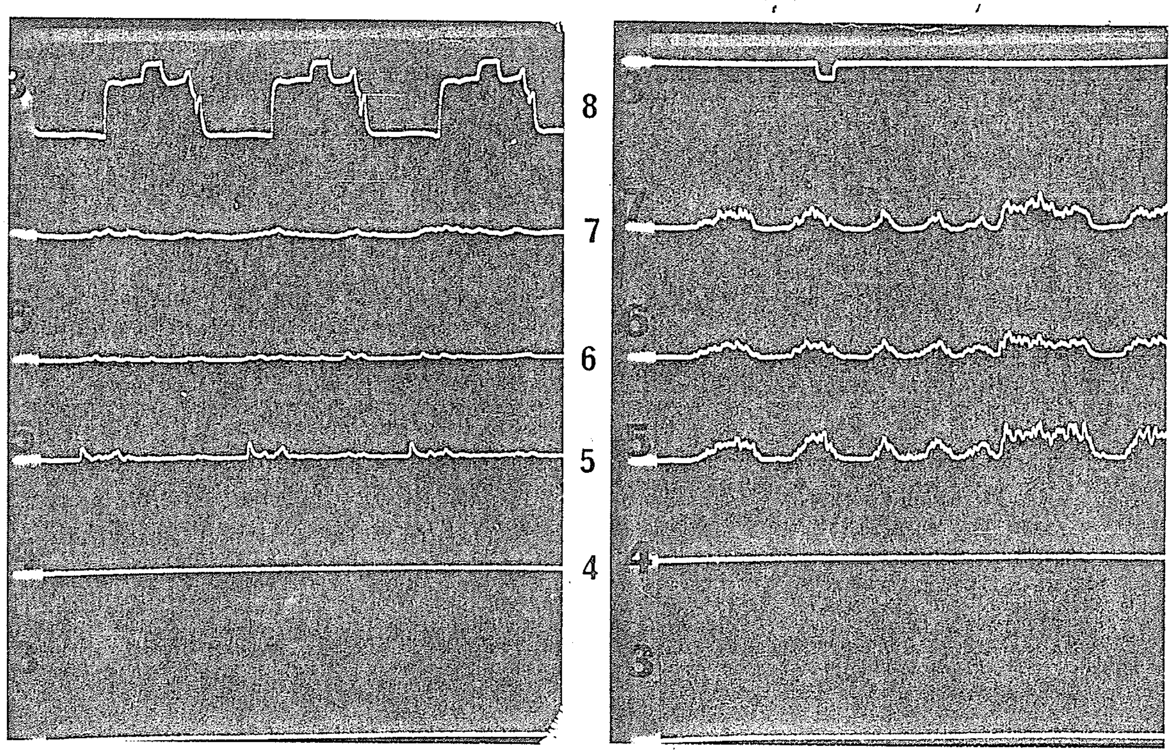


Fig. 25 Pictures showing test results determining muscle viability. Left is a normal walk and right is a voluntary isometric contraction while standing on both legs. The vertical scale is 1.0 V/cm. and the horizontal scale is 1.0 sec./cm.

Channel 4 unused

5 medial aspect of vastus lateralis

6 lateral aspect of vastus lateralis

7 hamstring muscle

8 output of a microswitch shoe used for temporal information. (The rising edge of the signal denotes heel strike.)

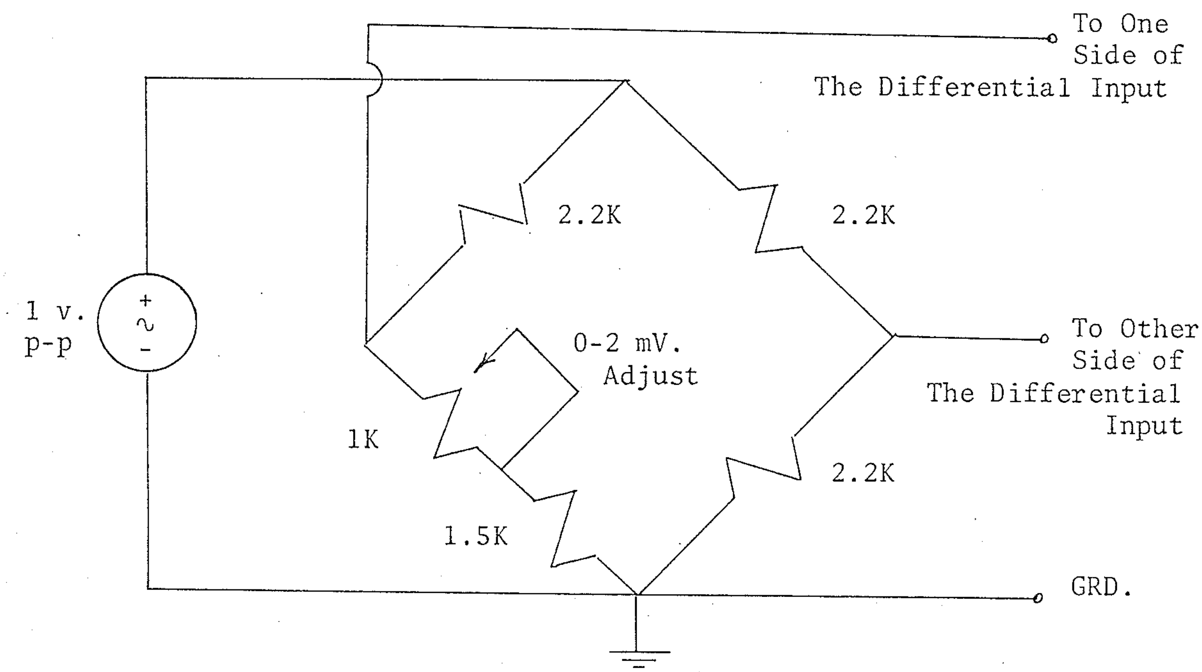


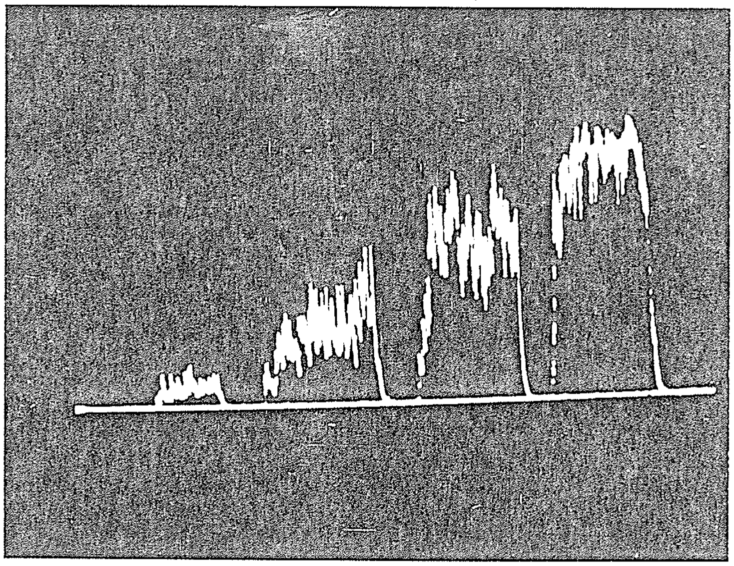
Fig. 26 Schematic of the 0 - 2 mV. Differential AC Test Circuit.

measured to be 30-600 Hz., which would pass EMG information, and eliminate high frequency noise and low frequency movement artifacts.

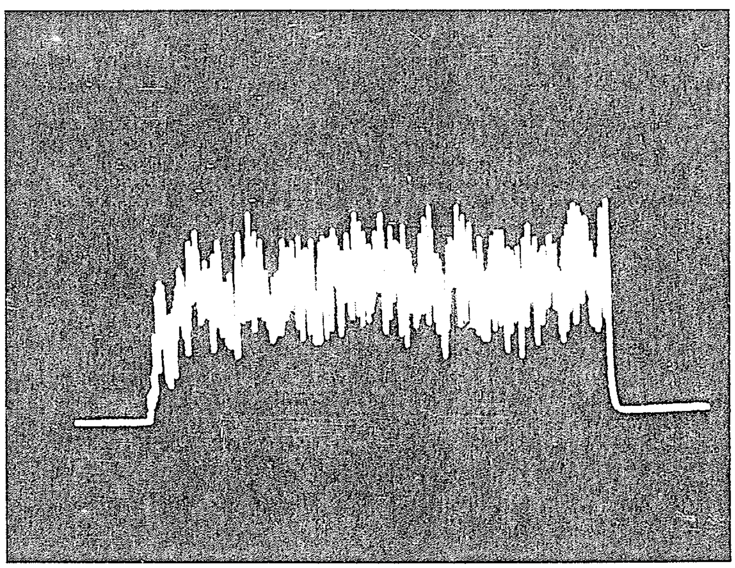
As a more realistic test, an actual EMG signal was fed into the inputs. Flexor carpi ulnaris was chosen, which is the biggest muscle on the medial aspect of the under side of the forearm. The wrist was flexed, and the signal observed at the output of the envelope detector (Fig. 27). The important thing noticed here, was that the DC level produced by EMG signals was not smooth, but rather very noisy. This suggested the need of a hysteresis effect in the design of the comparators for more stable detection, eliminating the possibility of valve chattering. Additional filtering was not implemented here because it would cause higher attenuation at the output and require even higher gains in the amplification stages - which wasn't practical. Otherwise, this section of electronics showed very good results.

4.3. Comparator Design

The next step in the developement of the EMG control electronics was the comparator stage (Fig. 28). A positive reference voltage was established at the non-inverting input, using a double voltage divider (Appendix 5), and thereby at rest cause the output to saturate to the positive biasing level. The output of the envelope detector was fed into the inverting input, and as soon as this level exceeded the voltage level at the non-inverting input, the comparator saturated to the negative biasing voltage. Therefore, by designing three comparators with different reference levels of increasing voltages, and using the output of the envelope detector as a common input, three levels of contraction could be distinguished, other than rest, and so create a four state



A.



B.

Fig. 27 Envelope Detector Output Signal With An EMG Input To The Differential Amplifier. Both signals were the result of Flexor Carpi Ulnaris activity, both have a vertical scale of 0.1 volts/cm. and a horizontal scale of 1.0 sec./cm.

A. An attempt at flexing and holding different levels of contraction, the last signal being one of maximum contraction.

B. An attempt at holding a medium contraction for some length of time.

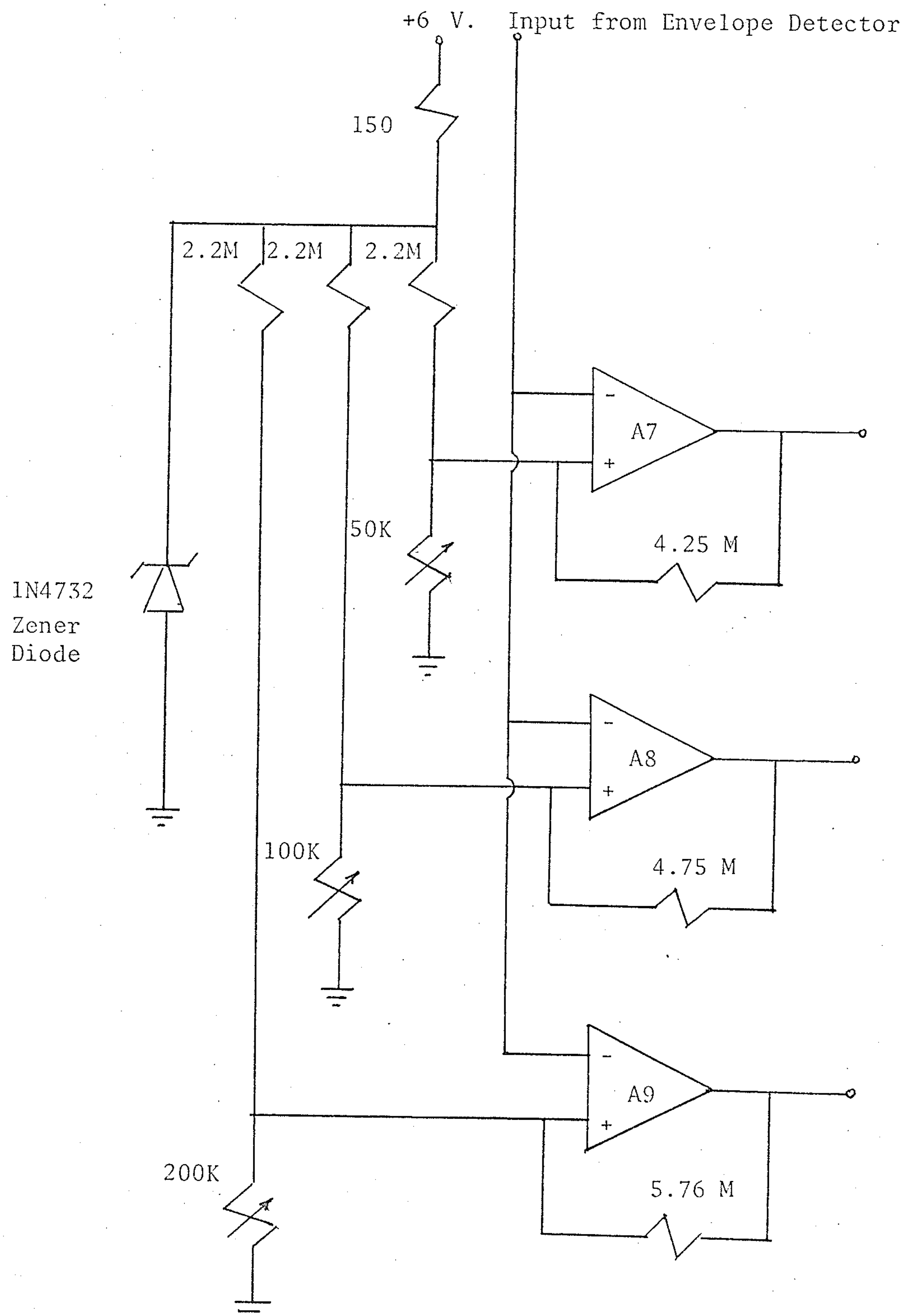


Fig. 28 Schematic of the Comparator Circuits

control system.

The hysteresis effect, suggested in the previous section, was accomplished by allowing the value of the feedback resistor to be in the order of the 2.2M resistor in the voltage divider circuit. The output, therefore, helped in establishing the reference voltage level, and once the comparator had switched states, immediately lowered the reference level at the input because of its negative polarity.

Referring to Fig. 27, it is seen that at low contractions, the noise has a small spread, whereas at high contractions, the noise spread is large. Measuring the widths of the noise levels, it was decided that state two should have a turn-on-turn-off voltage difference of about .18 volts, state three a difference of .25 volts and state four a difference of .30 volts. The comparators were then designed for the following levels: state two turns on at 0.1 volts and off at 0.02 volts, state three turns on at 0.2 volts and off at 0.05 volts, state four turns on at 0.3 volts and off at 0.10 volts.

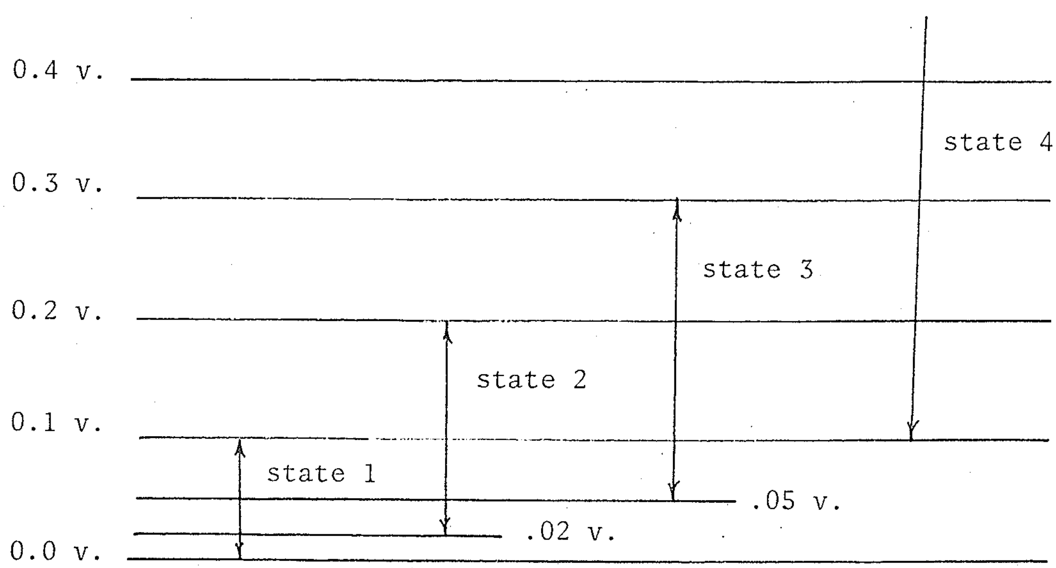


Diagram showing state voltage spreads.

The above diagram will now be used to help explain the comparator design. For example, state one (rest state) exists from zero volts to 0.10 volts when state two is activated. The turn off voltage for state two is 0.02 volts, however, state three will not be turned on until the voltage level exceeds 0.2 volts. Therefore, the state two voltage span is 0.20 - 0.02 = 0.18 volts. The voltage spans of states three and four can be explained similarly.

4.4. Logic Circuit Design

The outputs of the previously designed comparators are sensed by a simple logic circuit (Fig. 29), which, depending on the comparator states, energizes one or both power transistors connected to the outputs of the logic circuit. These transistors are used as power gates to the two solenoid valves in the hydraulic system. The logic circuit is comprised of an AND gate, and INVERTER, and a NOR gate. It was found that to attain enough power to switch the power transistors, and to use existing voltage supplies, it was easiest to design a discrete DTL (diode-transistor logic) circuit, also shown in Fig. 29.

When all comparators are off, no part of the logic circuit is activated, no power transistor switched, and therefore no solenoid energized. When comparator A7 is switched, suggesting state two exists in the system, the AND circuit and therefore the NOR circuit is activated and energizes the high resistance valve. When comparators A7 and A8 are on, state three, the INVERTER is activated, which energizes the low resistance valve. At the same instant, the INVERTER also deactivates the AND and NOR circuits and so de-energizes the high resistance valve. When all comparators are on, state four, the low

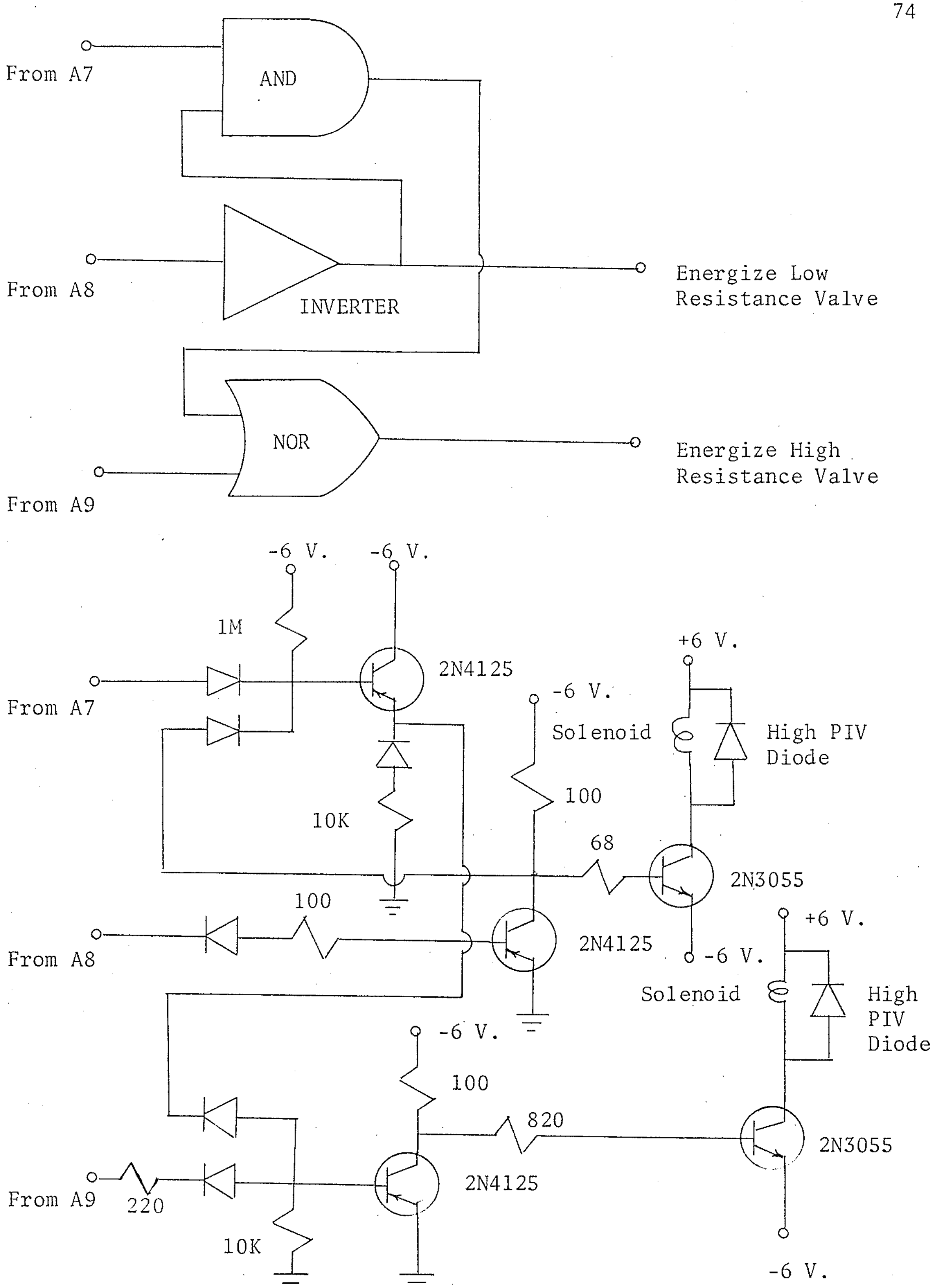


Fig. 29 Block Diagram and Schematic of the Logic Ciruit.

resistance valve remains energized and the NOR circuit is reactivated by comparator A9 and so re-energizes the high resistance valve. Now a lock situation exists in the system. The order is reversed when going from the lock to the rest state.

The whole EMG control electronics (Fig. 30) was then integrated in the shielded box. As was discussed earlier, provisions were made in this system to allow for an easy conversion to a three state control system. This was accomplished by breaking the connection between comparator A7 and the diode by a SPDT (single pole double throw) switch which connects the diode to either the comparator output (four state) or ground (three state). When the diode is grounded, the logic circuit can only be activated by comparators A8 and A9. Because the threshold setting resistors were chosen to be potentiometers, states two and three in the three state control system could be lowered and separated from each other, which provided a system which was much easier to control.

The circuit was again bench tested with the differential voltage source which showed that the circuitry was operating at the proper voltage levels. Then flexor carpi ulnaris was again chosen as an actual EMG site, and proved that the circuit could be successfully operated - three or four states - with just a few minutes of training.

4.5. Power Pack

The solenoid valves used in this prosthesis design were chosen to fit a 12 VDC system. The high resistance valve was chosen with a rating of 12 VDC operating voltage and 6 watts power dissipation therefore requiring a current of 0.5 amps. The low resistance valve was

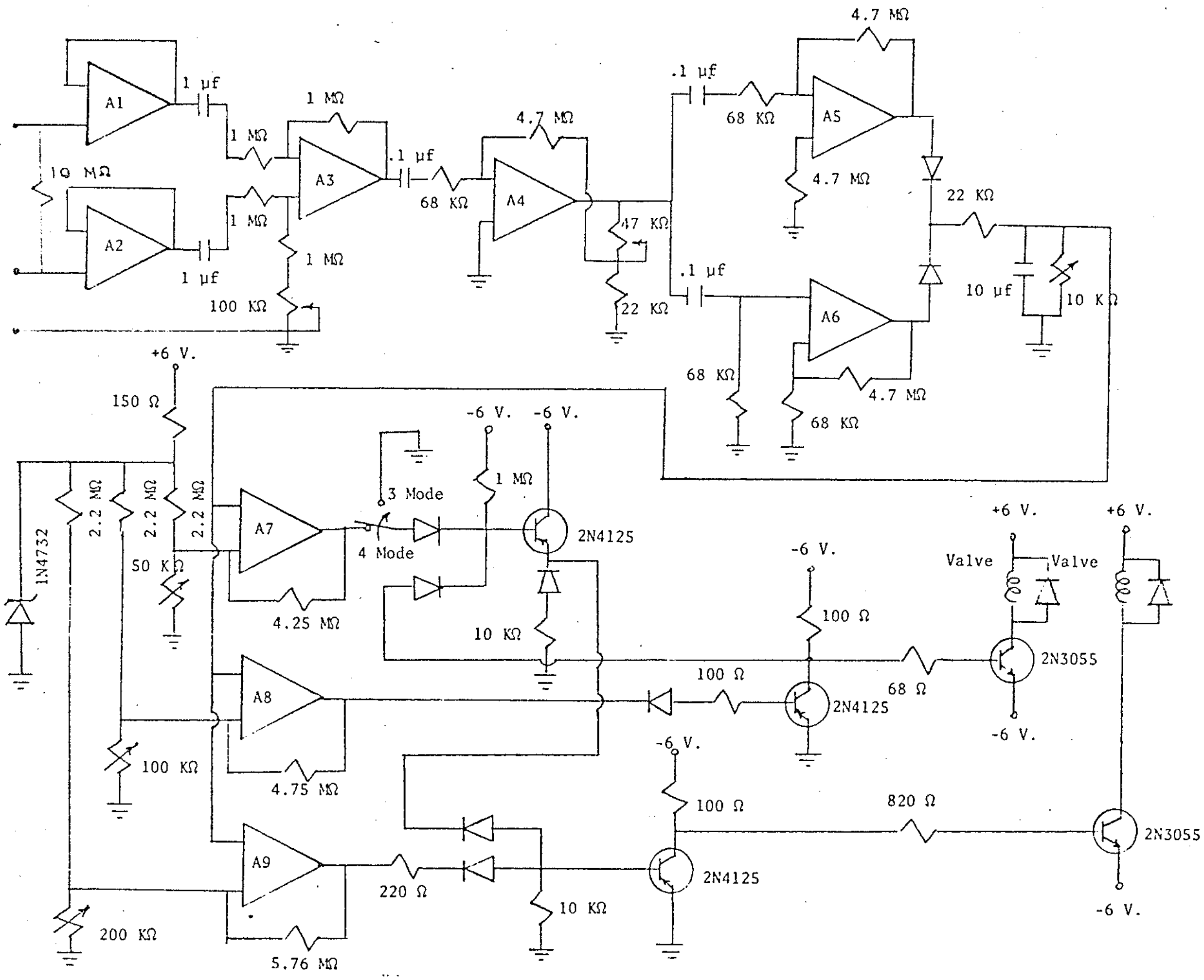


Fig. 30 Schematic of the complete EMG Control Electronics.

chosen with a rating of 10 VDC, in hope that this would allow for electrical switching at pressures higher than rated (as suggested by the manufacturer). It was rated at 10 watts and thus at 12 VDC drew 0.835 amps when operated. Since a high power rating is required of the battery pack, and because the leg would be used everyday, rechargeable nickel cadmium batteries were chosen to power the valves. These cells also have the added feature that the supply voltage would remain constant as long as the batteries have charged, and would then drop abruptly when discharged. Ten such cells, rated at 1.25 volts each, were stacked to give the required voltage. Their ampere hour rating was decided upon as follows.

According to the Cv curve fit (Fig. 12), and keeping in mind that the switch at heel contact was eliminated, it is seen that state two exists for 20% of the walking cycle, while state three 50%. State two would be associated with a 0.5 amp holding current and state three with a 0.835 amp holding current. It was estimated that the total walking time in one day might be in the order of three hours. Therefore, the ampere hour rating of the battery pack should be

$$3 \times 0.2 \times 0.5 + 3 \times 0.5 \times 0.835 = 1.5525 \text{ A. Hr.}$$

For amputee testing purposes, a 1.2 A-Hr. rating was used. A battery pack of a different rating would be used later, pending more accurate data on daily use.

The battery pack was also used in supplying bias and reference voltages to the electronic control system. Problems arose in the control system because of negative twelve volt spikes being reflected through the power pack when the valves were de-energized. To eliminate this phenomenon, a high PIV (peak inverse voltage) diode was added across the solenoids (Fig. 29 and 30), and a simple high frequency filter was connected to the plus and minus six volt terminals to ground to eliminate any spikes not eliminated by the diodes (Fig. 31). Since the energizing of the valves also causes a loading effect on the power pack, resulting in a voltage drop, and because the threshold voltages on the comparators are critical and must therefore be kept constant, a zener diode was added to the comparator circuit (Fig. 28). The 150Ω resistor at the top of the voltage divider was used to bias the diode for correct operation.

Careful consideration was always given to minimizing the power dissipation of the system. At rest, the electronic circuitry, three and four state, dissipated 0.24 watts. An on-off switch was added to the system (Fig. 31), which turned off the power to the electronics when the prosthesis was not used i.e. when sitting or standing still for long periods. Fuses were also incorporated into the plus and minus six volt lines to protect the battery pack from accidental short circuits in the electronics.

4.6. Summary

The following is a summary of what happens to an EMG signal as it passes from the input or source to the output (valve switching). A series of pictures is provided, Fig. 32, for a visual understanding. A differential EMG signal is fed into the control electronics and amplified (Fig. 32a). After full wave rectification by one part of the gain stage (Fig. 32b) the signal passes through an envelope detector (Fig. 32c) and is sensed by the comparators. From here on it shall be Assumed that state three exists in the system. Therefore comparators

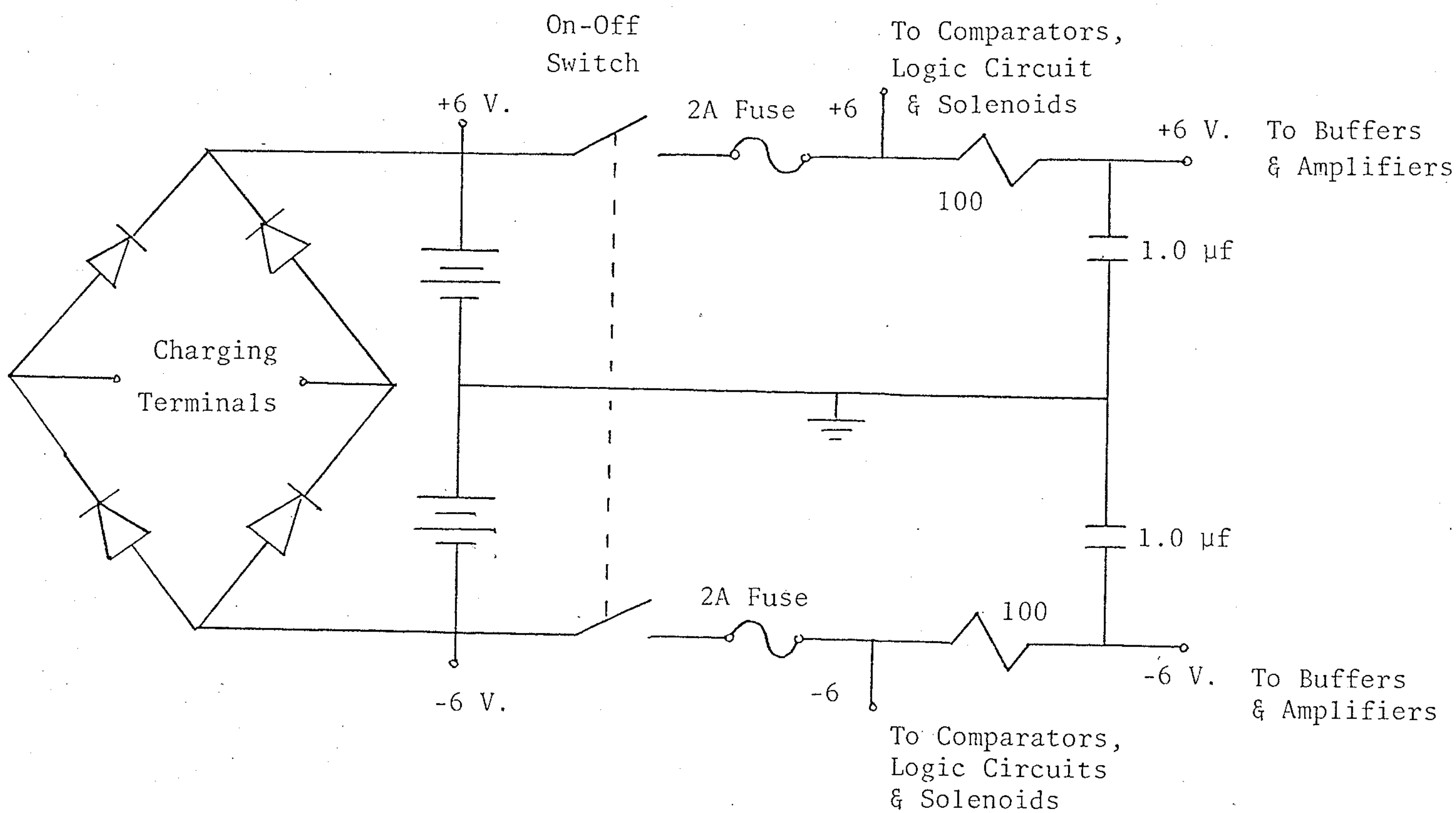
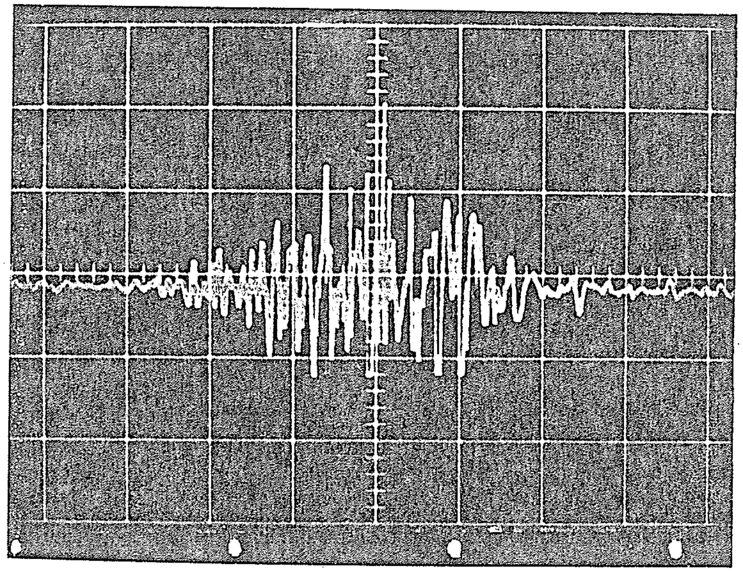
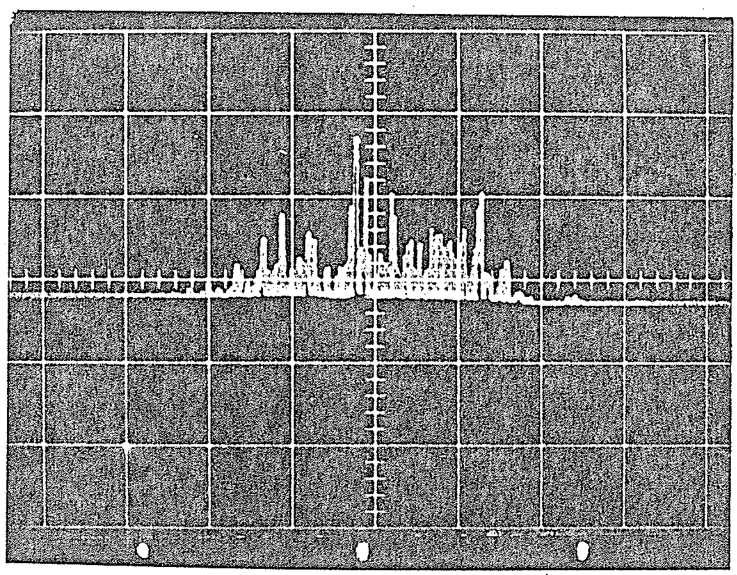


Fig. 31 Schematic of how the Power Pack is connected to the Electronics and Solenoids.



a. vert. 2v./cm. hor. .1sec./cm.

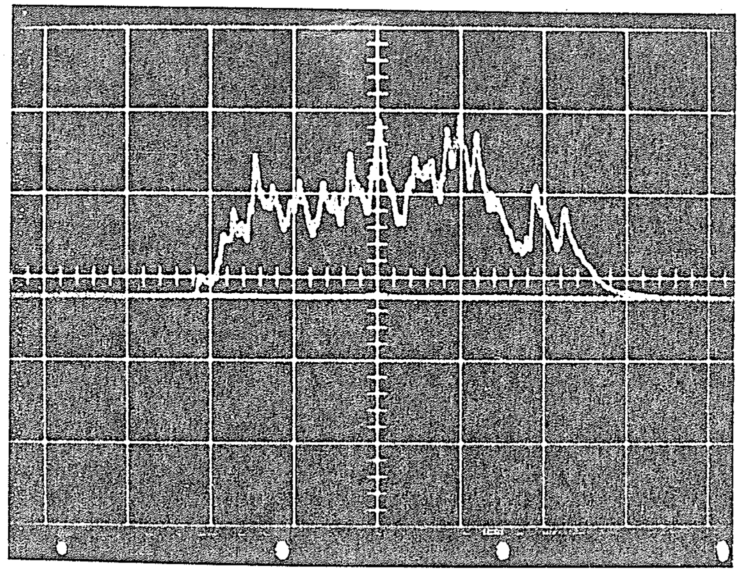


b. vert. 2v./cm. hor. .1sec./cm.

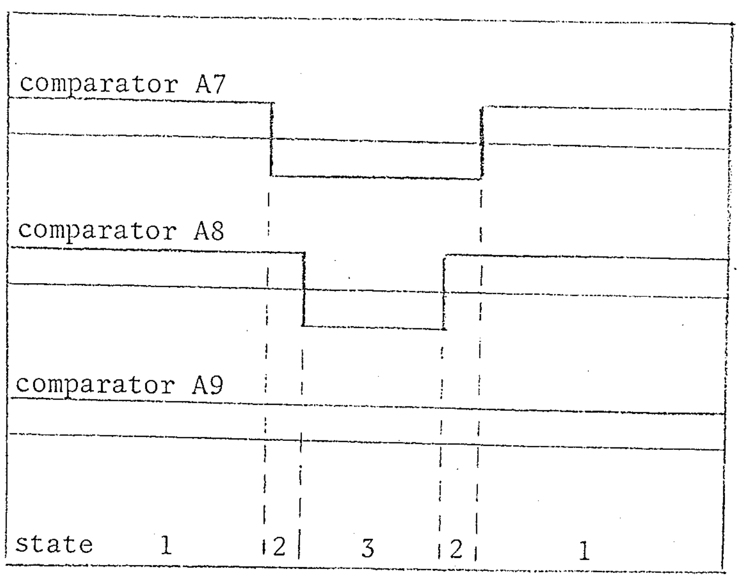
Fig. 32 Pictures and diagram showing the outputs of various sections of the control electronics.

a. after amplification of the EMG input

b. after rectification



c. vert. .1v./cm. hor. .1sec./cm.



d. vert. 5v./cm. hor. .1sec./cm.

Fig. 32 (cont'd)

c. after envelope detection

d. output of comparators

A7 and A8 have switched to -6.0 volts (bias voltages were ± 6 v) while comparator A9 remains at +6.0 volts (Fig. 32d). The inverter has been activated which not only energizes the low resistance valve, but also prevents the high resistance valve from energizing by deactivating the AND gate and therefore the NOR gate. And so, state three which consists of only the low resistance valve energized, has been obtained.

CHAPTER V

AMPUTEE TESTING

5.1 Preliminary Amputee Tests

The amputee was a male, 46 years old, 5' 7 3/4" tall, 140 pounds, very active and very enthusiastic. He was amputated about 6" above the knee in 1965 (Fig. 33), put on a pneumatic training prosthesis 13 days after the amputation, and then fitted with a standard hydraulic prosthesis about 6 months later. The amputee, showing a keen interest in the project, was considered invaluable with respect to feedback concerning the operation of the system.

As was mentioned earlier in Chapter 4, the first test performed on the amputee was an EMG test on the quadriceps and hamstrings, to discover muscle viability and therefore possible EMG sites to be used as input supplies to the control electronics. The level of contraction that could be obtained was also used in the determination of the amplification required by the amplifiers. The test consisted of attaching wet electrodes to a quadricep muscle (vastus lateralis) and a hamstring muscle, having the amputee put on his prosthesis, and then do controlled contractions while walking (Figs. 25, 34). The EMG activity was transmitted to a storage oscilloscope via a portable FM-FM telemetry system [11] available at the Winnipeg Shriners Hospital Locomotion Laboratory.

The amputee's muscle signals were observed during normal walking, and voluntary isometric contractions while standing still on both legs (Fig. 25). These results showed that there were distinguishable EMG

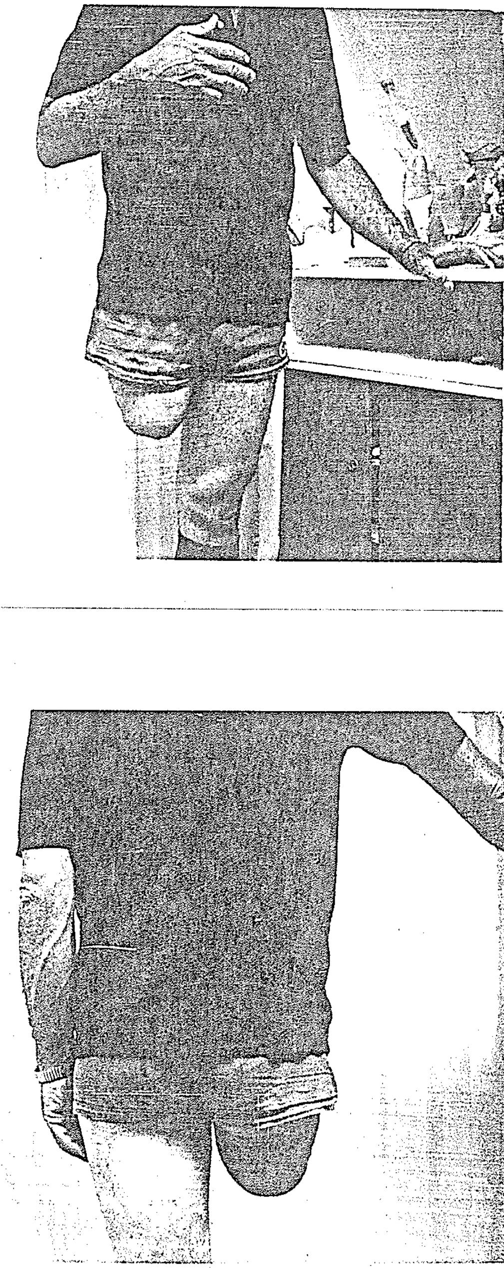


Fig. 33 Pictures showing the front and back view of the stump.

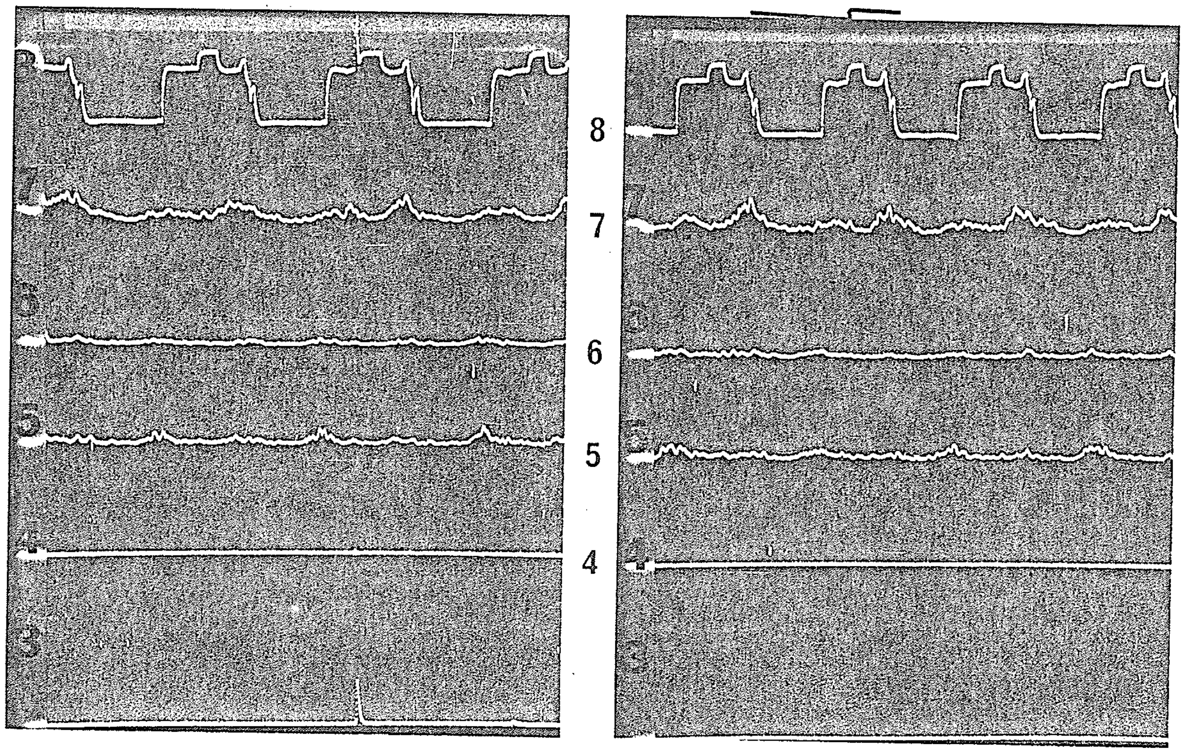


Fig. 34a Pictures showing test results determining muscle viability. Left is a walk with a voluntary contraction at push off and right is also a voluntary contraction at push off, but after a short rest. The vertical scale is 1.0 V./cm. and the horizontal scale is 1.0 sec./cm.

Channel 4 unused

5 medial aspect of vastus lateralis

6 lateral aspect of vastus lateralis

7 hamstring muscle

8 output of a microswitch shoe used for temporal information. (The rising edge of the signal denotes heel strike.)

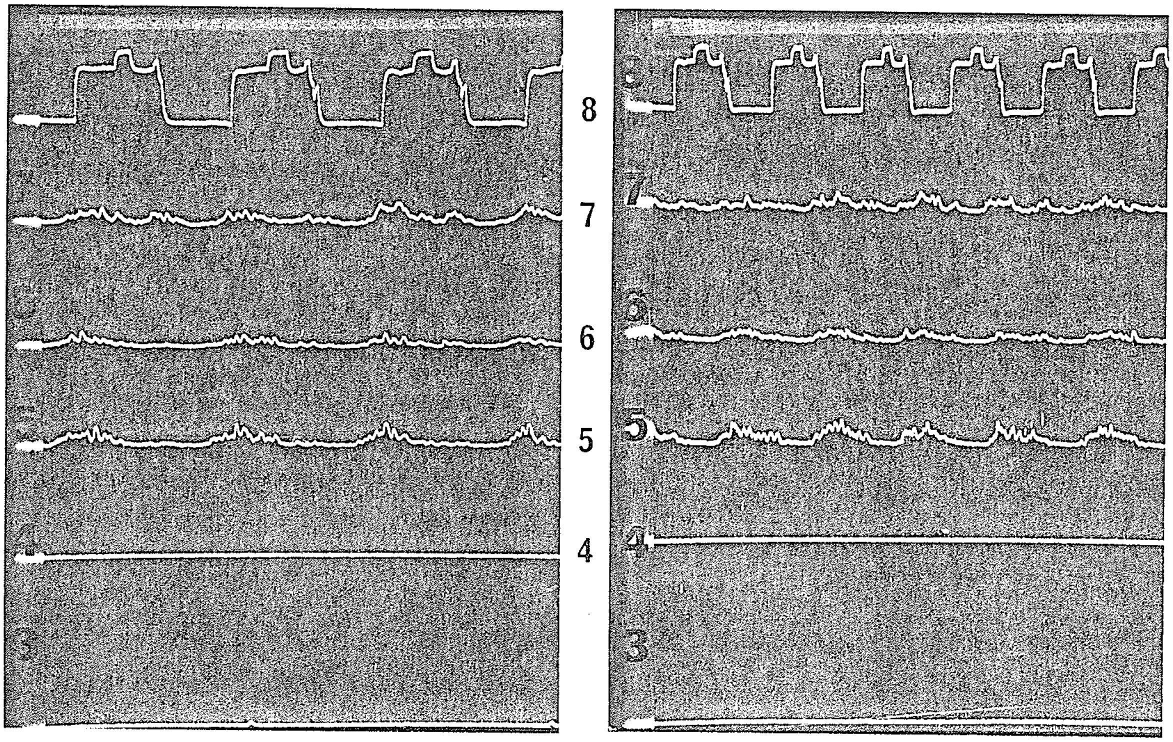


Fig. 34b Pictures showing test results determining muscle viability. Left is a walk with a voluntary contraction at heel strike and right is a fast walk with voluntary activity during swing phase. The vertical scale is 1.0 V./cm. and the horizontal scale is 1.0 sec./cm.

Channel	4	unused
	5	medial aspect of vastus lateralis
	6	lateral aspect of vastus lateralis
	7	hamstring muscle
	8	output of microswitch shoe

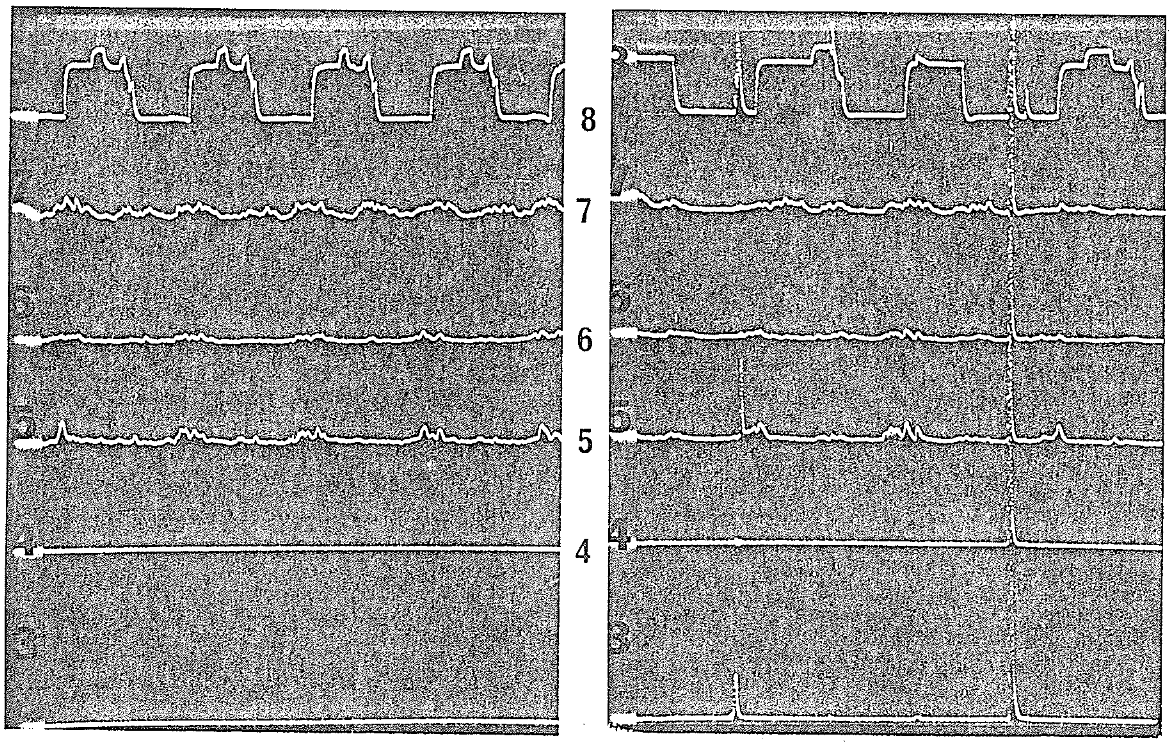


Fig. 34c Pictures showing test results determining muscle viability. Left is a fast walk with no voluntary contraction and right is a normal walk down an incline. The vertical scale is 1.0 V./cm. and the horizontal scale is 1.0 sec./cm.

Channel 4 unused

5 medial aspect of vastus lateralis

6 lateral aspect of vastus lateralis

7 hamstring muscle

8 output of microswitch shoe

signals produced in the stump, and their amplitude suggested a gain of about 1000 by the amplifiers.

Other tests performed included having the amputee walk and voluntarily contract at push-off, at heel strike, and during swing phase (Fig. 34). A fast walk and a normal walk down a ramp were also done. The results were very good in that he could contract his muscles voluntarily at just about any point in the walking cycle, although the strength of contraction was only about 0.5 mV. This low amplitude of contraction was to be expected, since he really had not used his stump muscles to any great degree, for almost ten years.

Continued testing was done with an exploring electrode while seated (results not recorded). These tests showed that the quadricep muscle, rectus femoris, yielded the strongest EMG signals - approximately 1.0 mV. Together with the fact that in a normal leg, stability of the knee is achieved by quadricep control, it was decided that rectus femoris would be the first control site attempted in testing the control electronics.

The four state control circuit was mounted on a belt (Fig. 35), which kept it out of the way while performing the next tests. These tests included evaluating both the hydraulics and electronics on the amputee separately. The amputee wore the prosthesis minus the electronics in order to familiarize himself with it alone. This was possible because without the electronics, or with the electronics turned off, both bypass valves are open and the leg functions similar to other hydraulic prostheses.

The results were as follows. Even though the prototype weighed 10 pounds, and the standard hydraulic prosthesis weighed 7½ pounds, the amputee felt no noticeable weight difference. The damping cylinder in

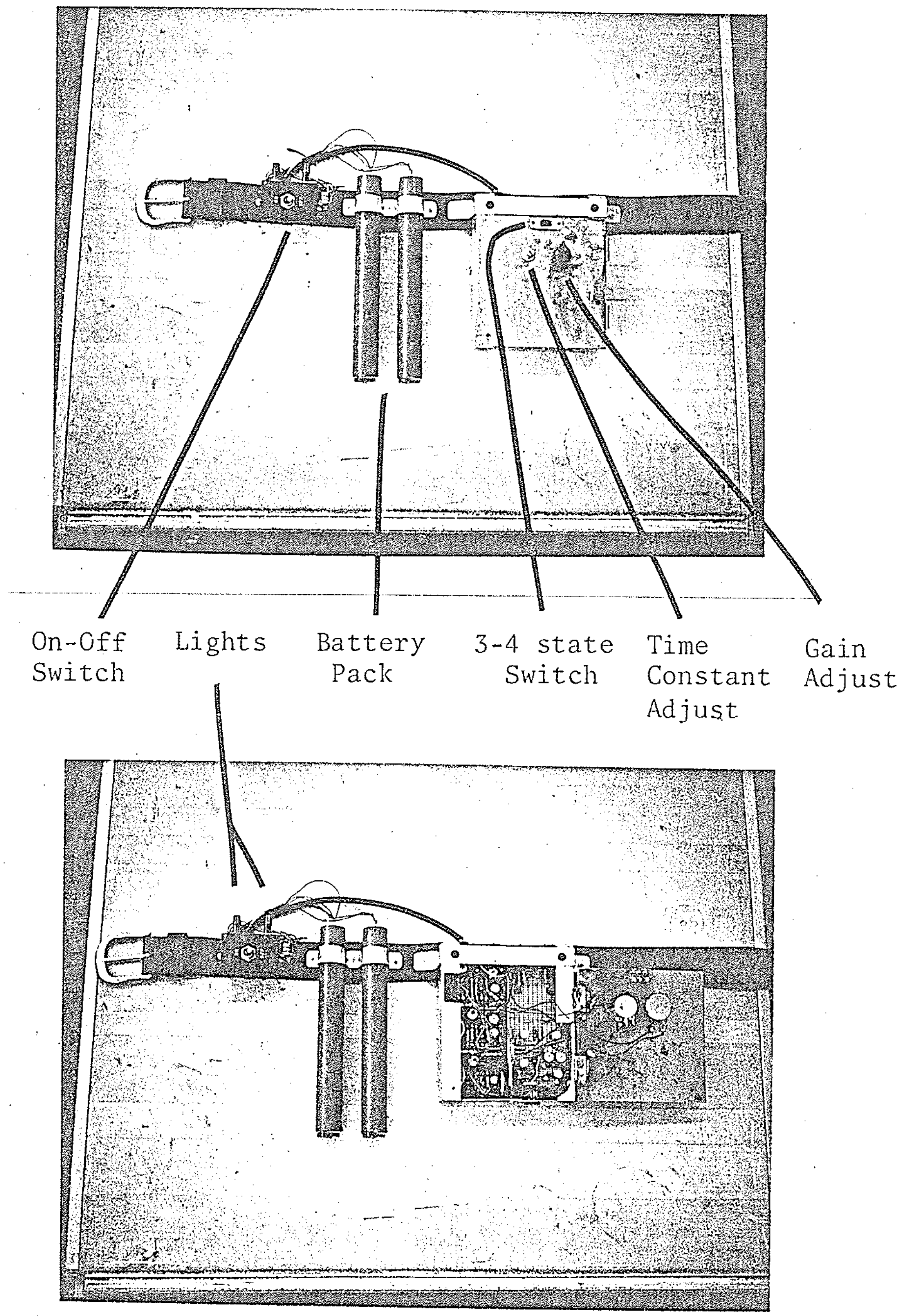


Fig. 35 Pictures showing the Four State Control System mounted on a belt.

the knee, however, was ordered with a cushion at the top end which hindered good knee flexion at heel contact and toe off considerably, and was therefore eliminated.

Wet electrodes were placed on rectus femoris and the electronic circuit was tested. It was discovered that the EMG signals were weaker than expected, but improved greatly when the amputee was given something to contract against, i.e. his prosthesis socket wall. It was also found, that there was no easy way to determine how many levels of control the amputee was actually operating and how accurately, since only an oscilloscope was connected to the valve leads, and switchings were too fast and undecodable to the naked eye. Two low power 12 VDC lights were then connected to the valve leads and used as indicators of state (rest state both off, state two one on, state three the other on, and state four both on).

With these modifications, the amputee was retested with the complete system. Although the cushioning effect was eliminated and knee flexion improved, the leg still seemed too stiff to walk on. The hydraulic fluid viscosity was measured at this time and was found to be extremely viscous. Viscosity tests were then performed on a number of available fluids with the results shown in Appendix 6. Brake fluid was chosen for the next test because it was much less viscous than the oil and also possessed the added characteristics of operation at high pressures and throughout the normal atmospheric temperature range.

As for the electronics, when seated, the amputee was able to control the four states quite well after only a short training period of about 15 minutes. However, during gait, the threshold level on comparator A7 had to be raised because the resting noise level in rectus femoris was higher. Since the threshold level of comparator A9 had to be lowered somewhat to make a lock possible, not much range was left for the existance of state three and so the amputee began experiencing difficulties in controlling four states accurately because of their close proximities. Therefore it was decided to use three state control in further tests, until the amputee had increased the strength of his EMG signals and has developed better control of his range.

A problem was also encountered with the site of EMG pickup. Although good control was achieved by rectus femoris while seated, during gait, when the hip was flexed, an uncontrolled lock appeared consistantly at various times during the walking cycle. This implied that rectus femoris was still being used for too actively in hip flexion and therefore could not be used as a voluntary control site to operate the control circuit. Because it was noticed that the amputee was using his hamstrings to a far greater degree for prosthesis stability than his quadriceps, a hamstring muscle was the next site tried.

Another important fact becoming evident at this time was that the wet electrodes and their leads were cutting into the skin on the stump. This was due to the fact that the amputee was now actually walking with the complete electro-hydraulic system, and the excessive pressures developed between the stump and the socket were causing the electrodes to imbed themselves into the skin. It was decided that to eliminate this ill effect, the electrodes would next be mounted into the socket, flush with the wall.

The electrodes used in the socket were dry electrodes. The 10 MΩ resistor across the differential input (Fig. 24) was removed to increase the differential input impedance and thereby the input sensitivity. It was thought, that if the dry electrodes worked, no electrode preparation would be required and thus create a better system. A new socket was contructed from the one the amputee normally used, and the electrodes were epoxied into the socket wall over the semimembranosis (a superficial medial hamstring muscle). Since a suction socket was being used, care had to be taken when mounting the electrodes so as not to allow any air leaks.

With the electrodes in the socket and the brake fluid in the hydraulic system, the amputee was again tested. The hydraulic system was still a little stiff and so the brake fluid was replaced by kerosene, which was even less viscous (Appendix 6). It was also found that the amputee could not control the operation of the electronics whatsoever. Noise and oscillations presented problems in the control circuit which were traced to the dry electrodes themselves. The dry skin in contact with the electrodes was introducing too much noise into the differential inputs, and by exceeding input specifications of the operational amplifiers (Appendix 5), caused oscillations to occur throughout the system which could not be eliminated.

The 10 MΩ resistor was replaced and the dry electrodes were substituted with wet electrodes. These were also epoxied into place (Fig. 36), but allowed for a small hole at the back of the electrodes which was used as a means for applying EMG paste to the electrodes, once the stump was in the socket. After injection of the paste, the holes were plugged so no fluid could escape through them.

The amputee was retested with the new system. The leg still appeared somewhat stiff, but the stiffness at this point was attributed to the resistance to flow developed by the long length of hydraulic

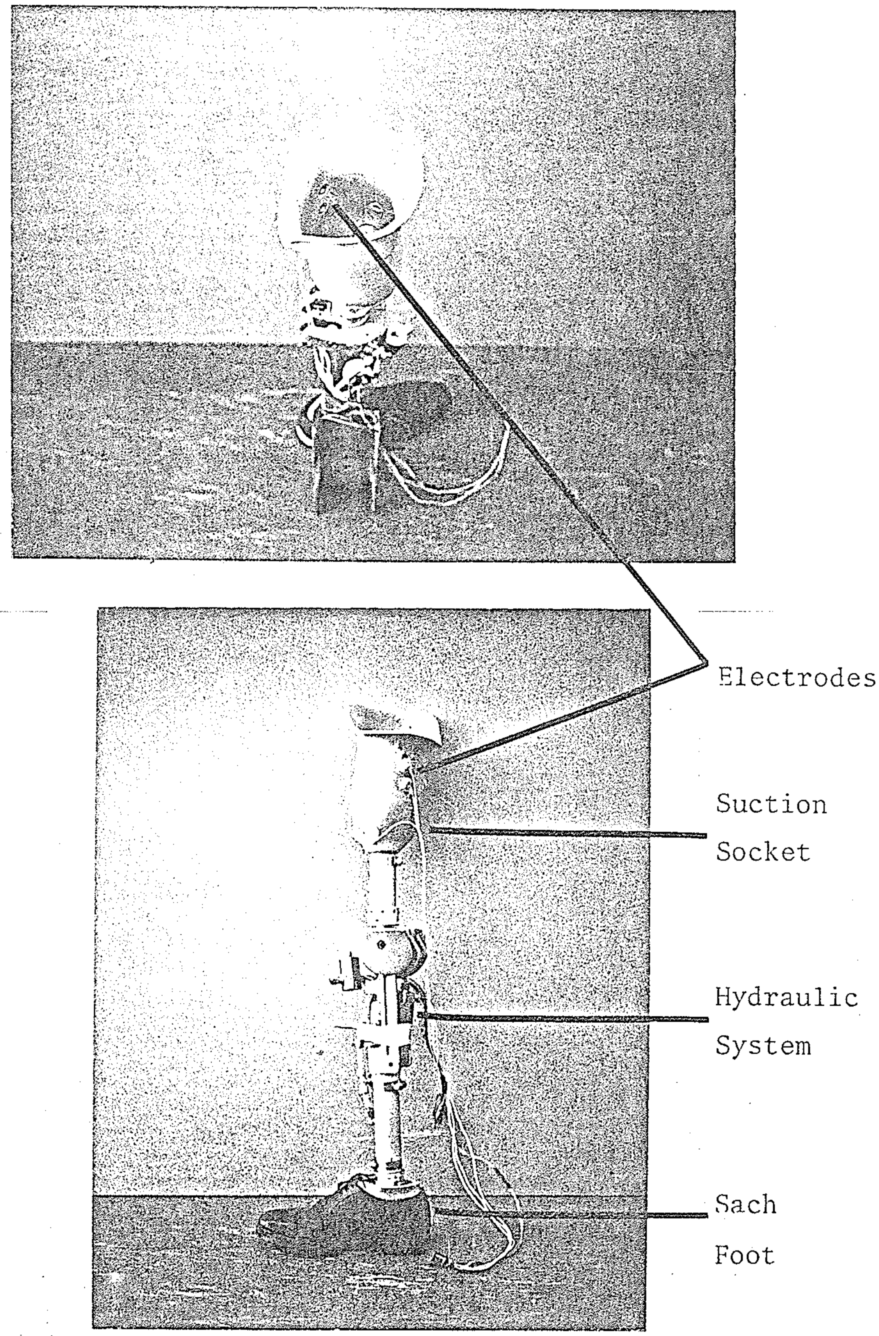


Fig. 36 Pictures showing the complete prototype prosthesis.

tubing used. Since future work entails eliminating this lengthy tubing, it was hoped that the stiffness problem would also disappear. The hydraulic system was therefore considered adequate for these preliminary tests.

Further tests on the electronics showed that the control circuit had unfortunately retained its oscillatory behavior. The trouble was traced to the fact that the filter on the bias supply to the amplifiers was not holding the voltage levels constant enough in the presence of the loading effects of the solenoids. It was suspected that this varying supply voltage was causing the amplifiers to oscillate. To eliminate this problem, the 1.0 μf capacitors in the supply filter (Fig. 31) were replaced by 4.7 V. zener diodes. This arrangement held the bias supply voltages constant and still retained the capacitor characteristic of eliminating any spikes reflected through the power pack by the solenoids.

5.2. Final Amputee Tests

With the zener diodes in place, the amputee was asked to try to operate the control circuit again. The test was a success. Within just a few minutes of training he was able to control the three states quite accurately. The threshold levels of the two comparators were adjusted for easiest operation - state two was set to .08 V. and state three was set to .20 V. - and the amputee walked with the complete system (Fig. 37), to discover if he could operate the three states just as accurately during gait. This again, after only a few minutes of training was accomplished easily.

Another test performed was having the amputee walk 'normally',

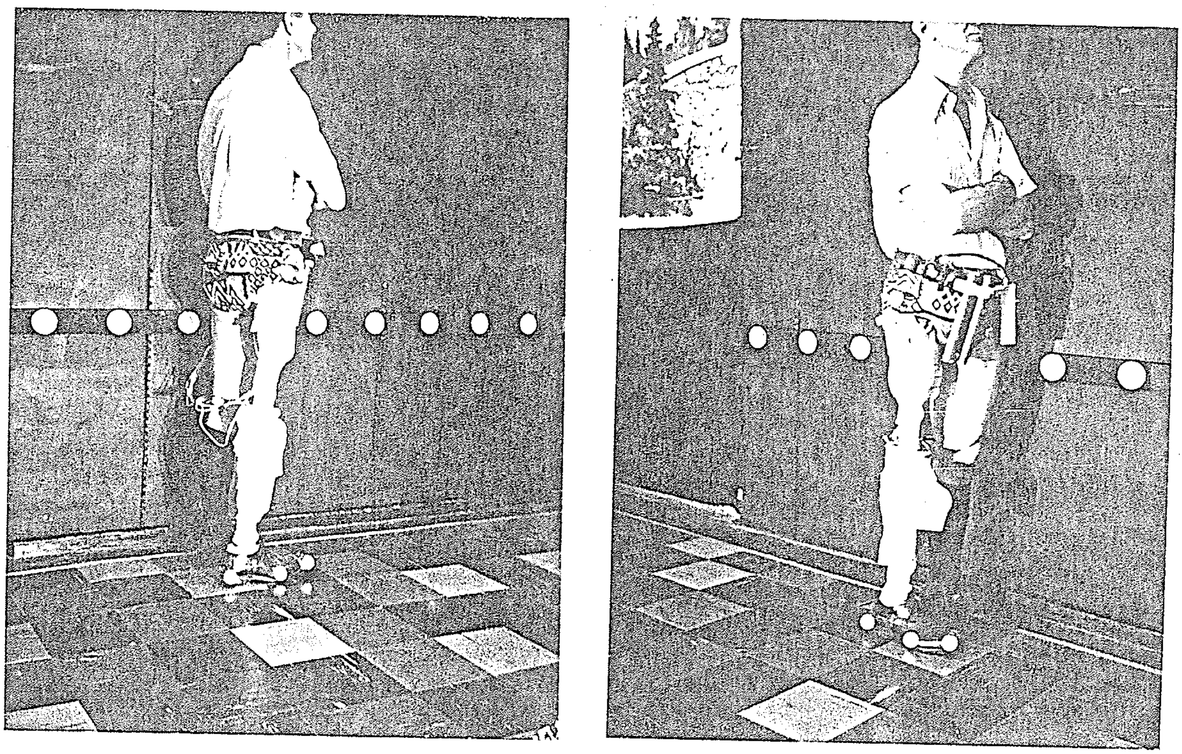


Fig. 37 Pictures showing the amputee actually wearing the complete system. The belt has been turned around so that the lights on the belt can be readily seen while the amputee is walking.

i.e. not consciously trying to activate any particular state. This was done to discover how naturally the leg was being operated by the amputee by observing, via the lights on his belt, what states were being activated at the different phases of the walking cycle. State two appeared twice in the walking cycle, once at heel contact and once at push off. This result showed that the correct states were being operated automatically at the right times, which led to the anticipation that this system required minimal extra training. The amputee was actually subconsciously activating a higher level of resistance to flexion when needed.

Also, at one point while the amputee was walking, a stumbling situation was encountered, and the amputee, again automatically, locked the knee preventing a fall. The amputee was very pleased with the test results and thus the prototype was considered a success.

5.3. Future Amputee Tests

Although this voluntary controlled variable resistance prosthesis has proved successful, much future work is yet to be done. A new socket will be contructed, since some discomfort was experienced by the amputee with the present one, and the hydraulic system will be integrated into a single unit reducing the size and weight of the whole system. The electronics will be microminiaturized and mounted in the prosthesis itself. Now with the reduced size of the hydraulic system, a cosmetic covering would also be possible for the leg.

Once adequate control is attained, a detailed locomotion study will be performed to fully evaluate the performance of the prosthesis, and thus point out what is still lacking. An attempt would then be made to compensate for any difficiency by adding, modifying, or redesigning the system. Locomotion tests would be conducted again, and after a few cycles of this loop, it is hoped that a maximum function prosthesis will emerge.

Other considerations could be the monitoring of actual forces in the damping cylinder piston rod and comparing the results with the predicted values. Also, doing a knee moment study on the prosthesis, it would be interesting to discover the contributions to this moment by each term in the moment expression [9]. Adding a bio-feedback mechanism would give the amputee a sense of prosthesis position without looking [10]. Another addition might be the incorporation of some energy storage device (i.e. a spring in the damping cylinder) for conservation of energy. More work could also be done in trying a dry electrode system again, instead of wet electrodes, which would minimize preparation and maintenance.

CHAPTER VI

CONCLUSION

A voluntary controlled variable resistance lower limb prosthesis was designed from a detailed analysis of normal gait. A prototype, consisting of a standard endoskeletal prosthesis fitted with a hydraulic damper across the knee, bypassed by two solenoid valves, was built. The system was then tested on an actual amputee who voluntarily controlled the damping resistance across the knee with a discrete three state control system. The voluntary control arises from the fact that the input to the control electronics comes from EMG signals derived from residual thigh muscles in the stump.

The amputee was very pleased with the performance of the prosthesis because it was so natural to operate. He was especially pleased when he automatically locked the knee in an inadvertant stumbling situation encountered in the lab, and recovered from a potential fall. Because the amputee was able to voluntarily control the resistance to flexion, and also voluntarily control the lock, aiding in recovery from potential falls, the prototype was considered a success.

APPENDIX 1

Glossary of some medical terminology.

BONES:

Femur - the strong bone in the thigh

Tibia - the strong and massive bone on the medial side of the shank.

Fibula - the smaller bone on the lateral side of the shank.

MUSCLES:

Calf Group - the group of muscles in the posterior shank which act as strong plantar flexors of the ankle.

Hamstring Group - consisting of Biceps Femoris, Semitendinosis, and Semimembranosis, in the posterior thigh, acting as flexors of the knee, and extensors of the hip.

Quadriceps Group - consisting of Rectus Femoris, Vastus Lateralis, Vastus Medialis, and Vastus Intermedialis, in the anterior thigh acting as strong extensors of the knee.

Pretibial Group - the group of muscles in the anterior shank acting as dorsiflexors of the ankle.

MOTIONS: (see Fig. 4)

Flexion - is a bending motion.

Extension - is a straightening motion.

Abduction - is a sideways motion away from the midline of the body.

Adduction - is a sideways motion toward the midline of the body.

Medial Rotation - is the motion, for example, when the elbow is flexed to 90 degrees but kept applied to the side, and then the forearm and hand are moved medially.

Lateral Rotation - is the opposite motion to medial rotation.

Dorsiflexion - is the turning up of the foot at the ankle.

Plantar Flexion - straightens the foot from a dosiflexed position.

Inversion - bending the foot medially at the ankle.

Eversion - bending the foot laterally at the ankle.

DIRECTIONS:

Anterior or Ventral - toward the front.

Posterior or Dorsal - toward the back.

Superior - toward the head.

Inferior - toward the feet.

Lateral - away from the midline of the body.

Medial - toward the midline of the body.

APPENDIX 2

The following is the computer program used in calculating the necessary design data for the above knee prosthesis.

```
$JOB WATFIV
C
C    THIS PROGRAM CALCULATES IMPORTANT CHARACTERISTICS
C    ASSOCIATED WITH THE KNEE DURING GAIT.
C    XCYC  IS % OF ONE STRIDE IN GAIT
C    XANG  IS ANGLE SHANK MAKES WITH LEG IN DEGREES
C    XRAD  IS ANGLE SHANK MAKES WITH LEG IN RADIANS
C    XVEL  IS ANGULAR VELOCITY OF KNEE IN DEG/SEC
C    XVEL1 IS ANGULAR VELOCITY OF KNEE IN RAD/SEC
C    XCYL  IS VELOCITY OF PISTON IN DAMPING CYLINDER
C    XMOM  IS MOMENT ABOUT THE KNEE
C    XFOR  IS FORCE CYLINDER MUST CREATE TO CAUSE SUCH A MOMENT
C    RATE1 IS FLOW RATE A VALVE (CV=1) CAN HANDLE WITH XFOR ACROSS IT
C    RATE2 IS FLOW IN CYLINDER DICTATED BY VELOCITY & AREA OF PISTON
C    R     IS RESISTANCE TO FLOW ASSUMING LINEAR VALVES
C    C1    IS RESISTANCE TO FLOW ASSUMING NONLINEAR VALVES
C    XX    IS TOTAL LENGTH OF CYLINDER & PISTON ROD
C    CINV  IS INVERSE OF C1 USED TO COMPARE WITH R
C    XREL  XX-MIN LENGTH (WHEN KNEE BENT 90 DEGREES)
C          THIS IS USED FOR TEST JIG CAM DIMENSIONS
C
     DIMENSION XANG(41),XVEL(41),XVEL1(41),XRAD(41),XCYC(41),
    *          XMOM(41),XCYL(41),XFOR(41),RATE1(41),RATE2(41)
     Dimension R(41),C1(41),XX(41)
C
C    HERE KNEE ANGLES & MOMENTS ARE READ IN
C    ALSO DAMPING CYLINDER PLACEMENT IS READ IN
C         XD IS DIST BETWEEN KNEE JOINT & LEG ATTACHMENT
C         XL IS DIST BETWEEN KNEE JOINT & SHANK ATTACHMENT
C          B IS REF ANGLE LEG ATTACHMENT MAKES WITH HORIZONTAL
C    RATE1 IS READ IN FROM MANUFACTURERS TABLES
C    AND CV IS CONDUCTANCE OF VALVE TO BE COMPARED WITH RATE2
C
     READ(5,*) (XANG(I),I=1,41),(XMOM(J),J=1,41)
     READ(5,*) XD,XL,B
     READ(5,*) (RATE1(I),I=1,41),CV
C
C    ANGLES ARE CONVERTED TO RADIAN MEASURE
C    AND ANGULAR VELOCITIES ARE CALCULATED
C
     DO 1 K=1,41
     XRAD(K)=XANG(K)/57.3
1    XCYC(K)=(K-1)*2.5
     XVEL(1)=(XANG(2)-XANG(1))/.033
     XVEL1(1)=(XVEL(1)/57.3)
     XVEL(41)=(XANG(41)-XANG(40))/.033
     XVEL1(41)=XVEL(41)/57.3
```

```
      DO 2 L=2,40
      XVEL(L)=(XANG(L+1)-XANG(L-1))/.067
2     XVEL1(L)=XVEL(L)/57.3
C
C     HERE NOTHING BUT HEADINGS ARE WRITTEN OUT
C
      WRITE(6,22)XD,XL
22    FORMAT('1',//5X,'DATA: RADIUS WHICH THIGH CONNECTION OF CYL MAKES
     *WITH KNEE = ',F5.3,' INCHES'/10X,'DISTANCE FROM KNEE OF LOWER CONN
     *ECTION OF CYL = ',F6.3,' INCHES')
      WRITE(6,24)B,CV
24    FORMAT(' ','    ASSUMING A ',F5.1,' DEGREE REFERENCE ANGLE, AND A
     *WALKING SPEED OF 90 SPM.'/'     THE VALVE CONSIDERED HERE HAS A CV
     * OF ',F6.3)
      WRITE(6,3)
3     FORMAT(' ',//14X,'KNEE  POSITION',5X,'KNEE  VELOCITY',5X,'CYL VEL.
     *',2X,'MOMENT',3X,'F ON CYL',2X,'FLOW RATES (GPM)'/5X,'% CYCLE',2X,
     *'ANGLE',2X,'RADIANS',2X,'DEG / SEC',2X,'RAD / SEC',2X,'FT / SEC',
     *2X,'FT - LB',3X,'IN LB.',5X,'CYL',4X,'VALVE'/5X,7('-'),2X,5('-'),
     *2X,7('-'),2X,9('-'),2X,9('-'),2X,8('-'),2X,7('-'),3X,6('-'),5X,
     *3('-'),4X,5('-'))
C
C     NOW ALL OTHER DATA IS CALCULATED AND WRITTEN OUT
C
      DO 4 M=1,41
      T=B/57.3+XRAD(M)
      PHI=ATAN(XD*COS(T)/(XL-XD*SIN(T)))
      ALPHA=T-PHI
      XCYL(M)=XD*XVEL1(M)*COS(ALPHA)/12.0
      RX=XL*SIN(PHI)/12.0
      XFOR(M)=XMOM(M)/RX
      RATE2(M)=XCYL(M)*3.12
      C1(M)=ABS(RATE2(M))/RATE1(M)
      RATE1(M)=RATE1(M)*CV
      XX(M)=SQRT(XD**2.0+XL**2.0-2.0*XL*XD*COS(1.57+T))
4     WRITE(6,5)XCYC(M),XANG(M),XRAD(M),XVEL(M),XVEL1(M),XCYL(M),XMOM(M)
     *          ,XFOR(M),RATE2(M),RATE1(M)
5     FORMAT(' ',4X,F7.3,2X,F5.1,2X,F7.3,2X,F9.2,2X,F9.4,2X,F8.3,2X,
     *      F7.3,3X,F6.1,3X,F6.3,2X,F6.3)
      WRITE(6,51)
51    FORMAT('1',////5X,'% CYCLE',2X,'RESISTANCE',2X,'CV NEEDED',2X,
     *'INV. CV',2X,'CYL LEN',2X,'LEN-MIN'/5X,7('-'),2X,10('-'),2X,
     *9('-'),2X,7('-'),2X,7('-'),2X,7('-'))
      DO 55 N=1,41
      CINV=1.0/C1(N)
      R(N)=XFOR(N)/(12.0*XCYL(N))
      XREL=XX(N)-6.375
55    WRITE(6,58)XCYC(N),R(N),C1(N),CINV,XX(N),XREL
58    FORMAT(' ',4X,F7.3,2X,F10.2,3X,F8.6,2X,F7.2,2X,F7.3,2X,F7.3)
C
6     STOP ; END

$ENTRY
```

DATA: RADIUS WHICH THIGH CONNECTION OF CYL MAKES WITH KNEE = 1.125 INCHES
DISTANCE FROM KNEE OF LOWER CONNECTION OF CYL = 7.500 INCHES
ASSUMING A 0.0 DEGREE REFERNECE ANGLE, AND A WALKING SPEED OF 90 SPM.
THE VALVE CONSIDERED HERE HAS A CV OF 0.100

	KNEE	POSITION	KNEE	VELOCITY	CYL VEL.	MOMENT	F ON CYL	FLOW RATES (GPM)						
% CYCLE	ANGLE	RADIANS	DEG / SEC	RAD / SEC	FT / SEC	FT - LB	IN LB.	CYL	VALVE	RESISTANCE	CV NEEDED	INV. CV	CYL LEN	CAM-DIM
0.000	0.0	0.000	-151.52	-2.6442	-0.245	5.000	53.9	-0.765	0.750	-18.33	0.101985	9.81	7.583	1.834
2.500	-5.0	-0.087	-149.25	-2.6048	-0.238	3.000	32.9	-0.741	0.575	-11.54	0.128902	7.76	7.485	2.029
5.000	-10.0	-0.175	-119.40	-2.0838	-0.186	1.000	11.2	-0.579	0.330	-5.04	0.175470	5.70	7.387	2.225
7.500	-13.0	-0.227	-104.48	-1.8233	-0.160	-9.000	-102.9	-0.498	0.970	53.73	0.051314	19.49	7.328	2.343
10.000	-17.0	-0.297	-89.55	-1.5629	-0.133	-20.000	-235.1	-0.415	1.450	147.30	0.028614	34.95	7.250	2.499
12.500	-19.0	-0.332	-44.78	-0.7814	-0.065	-32.000	-382.1	-0.204	1.850	486.48	0.011038	90.60	7.212	2.577
15.000	-20.0	-0.349	-14.93	-0.2605	-0.022	-44.000	-529.8	-0.067	2.200	2040.66	0.003068	325.94	7.192	2.615
17.500	-20.0	-0.349	22.39	0.3907	0.032	-42.000	-505.7	0.101	2.100	-1298.60	0.004821	207.41	7.192	2.615
20.000	-18.5	-0.323	44.78	0.7814	0.066	-38.000	-451.9	0.205	2.000	-573.02	0.010252	97.55	7.221	2.557
22.500	-17.0	-0.297	44.78	0.7814	0.066	-28.000	-329.1	0.207	1.750	-412.43	0.011854	84.36	7.250	2.499
25.000	-15.5	-0.271	52.24	0.9117	0.078	-16.000	-186.0	0.245	1.300	-197.60	0.018824	53.12	7.280	2.441
27.500	-13.5	-0.236	52.24	0.9117	0.080	-9.500	-108.9	0.248	1.000	-114.16	0.024807	40.31	7.319	2.363
30.000	-12.0	-0.209	52.24	0.9117	0.080	-5.000	-56.8	0.250	0.740	-58.96	0.033841	29.55	7.348	2.304
32.500	-10.0	-0.175	59.70	1.0419	0.093	-1.000	-11.2	0.290	0.330	-10.08	0.087735	11.40	7.387	2.225
35.000	-8.0	-0.140	59.70	1.0419	0.094	2.000	22.2	0.293	0.470	19.75	0.062243	16.07	7.427	2.147
37.500	-6.0	-0.105	44.78	0.7814	0.071	7.000	77.1	0.221	0.850	90.50	0.026051	38.39	7.466	2.068
40.000	-5.0	-0.087	14.93	0.2605	0.024	13.000	142.5	0.074	1.200	500.02	0.006177	161.90	7.485	2.029
42.500	-5.0	-0.087	-14.93	-0.2605	-0.024	15.500	170.0	-0.074	1.300	-596.18	0.005701	175.39	7.485	2.029
45.000	-6.0	-0.105	-59.70	-1.0419	-0.095	17.000	187.2	-0.295	1.350	-164.83	0.021670	45.73	7.466	2.068
47.500	-9.0	-0.157	-89.55	-1.5629	-0.140	14.000	156.4	-0.437	1.250	-93.11	0.034929	28.63	7.407	2.186
50.000	-12.0	-0.209	-126.87	-2.2141	-0.195	6.000	68.1	-0.608	0.800	-29.13	0.076023	13.15	7.348	2.304
52.500	-17.5	-0.305	-164.18	-2.8653	-0.243	-6.000	-70.8	-0.758	0.800	24.29	0.094718	10.56	7.241	2.518
55.000	-23.0	-0.401	-186.57	-3.2560	-0.263	-17.000	-210.3	-0.821	1.345	66.59	0.061055	16.38	7.135	2.730
57.500	-30.0	-0.524	-208.96	-3.6467	-0.273	-21.000	-280.1	-0.853	1.500	85.35	0.056875	17.58	7.005	2.990
60.000	-37.0	-0.646	-223.88	-3.9072	-0.267	-22.000	-322.3	-0.832	1.600	100.69	0.052011	19.23	6.881	3.238
62.500	-45.0	-0.785	-238.81	-4.1676	-0.249	-17.000	-284.9	-0.776	1.400	95.49	0.055416	18.05	6.751	3.498
65.000	-53.0	-0.925	-216.42	-3.7769	-0.190	-5.000	-99.5	-0.592	0.800	43.73	0.073983	13.52	6.636	3.729
67.500	-59.5	-1.038	-119.40	-2.0838	-0.088	0.000	0.0	-0.273	0.000	0.00	********	0.00	6.555	3.890
70.000	-61.0	-1.065	-7.46	-0.1302	-0.005	1.000	24.9	-0.016	0.370	-397.86	0.004404	227.06	6.538	3.923
72.500	-60.0	-1.047	59.70	1.0419	0.043	1.000	24.2	0.135	0.365	46.66	0.036872	27.12	6.549	3.901
75.000	-57.0	-0.995	119.40	2.0838	0.094	1.000	22.1	0.294	0.360	19.54	0.081704	12.24	6.585	3.831
77.500	-52.0	-0.908	149.25	2.6048	0.134	1.000	19.4	0.418	0.350	12.09	0.119457	8.37	6.649	3.702
80.000	-47.0	-0.820	149.25	2.6048	0.149	1.000	17.4	0.466	0.340	9.72	0.137147	7.29	6.721	3.559
82.500	-42.0	-0.733	179.10	3.1257	0.197	1.500	23.8	0.614	0.420	10.08	0.146275	6.84	6.798	3.404
85.000	-35.0	-0.611	208.96	3.6467	0.256	2.000	28.5	0.799	0.470	9.26	0.170098	5.88	6.916	3.169
87.500	-28.0	-0.489	223.88	3.9072	0.300	2.500	32.6	0.936	0.520	9.05	0.179918	5.56	7.041	2.917
90.000	-20.0	-0.349	238.81	4.1676	0.346	3.000	36.1	1.080	0.560	8.70	0.192851	5.19	7.194	2.615
92.500	-12.0	-0.209	223.88	3.9072	0.344	7.000	79.5	1.073	0.870	19.26	0.123363	8.11	7.348	2.304
95.000	-5.0	-0.087	208.96	3.6467	0.333	12.000	131.6	1.038	1.200	32.97	0.086472	11.56	7.485	2.029
97.500	2.0	0.035	74.63	1.3024	0.121	11.000	118.1	0.378	1.150	81.14	0.032908	30.39	7.622	1.757
100.000	0.0	0.000	-60.61	-1.0577	-0.098	5.000	53.9	-0.306	0.750	-45.83	0.040794	24.51	7.583	1.834

CORE USAGE OBJECT CODE = 4656 BYTES,ARRAY AREA = 2132 BYTES,TOTAL AREA AVAILABLE = 39056 BYTES

DIAGNOSTICS NUMBER OF ERRORS = 0, NUMBER OF WARNINGS = 0, NUMBER OF EXTENSIONS = 0

COMPILE TIME = 0,58 SEC,EXECUTION TIME = 0.74 SEC,DATE = 73/113,TIME=13:37:40

Table showing a Typical Output Page of the Computor Program listed on the previous two pages.

APPENDIX 3

TEST JIG

The test jig was designed to test the actual performance of the hydraulic system. The jig causes the piston in the damping cylinder to function as if it were in a prosthetic leg, going through a number of walking cycles.

A 1/6 hp. DC motor (3)(see included diagram) was mounted on a metal frame (12). A half wave rectifier (4) is connected to the field of the motor. This allows the motor to run at its 60 VDC rating. A feedback mechanism (8) is then connected to the armature to create a constant motor speed in the presence of varying torques. This feedback mechanism also incorporates a speed control mechanism to vary motor speed. The motor is run at speeds close to rated value, 1100 RPM, and so the speed is reduced through a pulley arrangement (13,14,15), which also amplifies motor torque. Through this speed reduction, pulley #3 (15) is made to rotate at 45 RPM which corresponds to a walking speed of 90 SPM.

On the same drive shaft (18) as the pulley, a computer designed cam (5) was attached. This cam was so designed so as to move the center of the driving lever (10), to which the damping cylinder (2) is connected, up and down to pump it in the same manner as it would be in a prosthetic leg. The bottom of the cylinder is fastened to a platform (16), adjustable in height and welded to the main frame, holding the damping cylinder in place while being pumped. One revolution of the driving cam (5) corresponds to one walking cycle, therefore by rotating this cam at 45RPM, the cylinder is forced to operate at 45 SPM also.

One end of the driving lever (10) is connected to the main frame

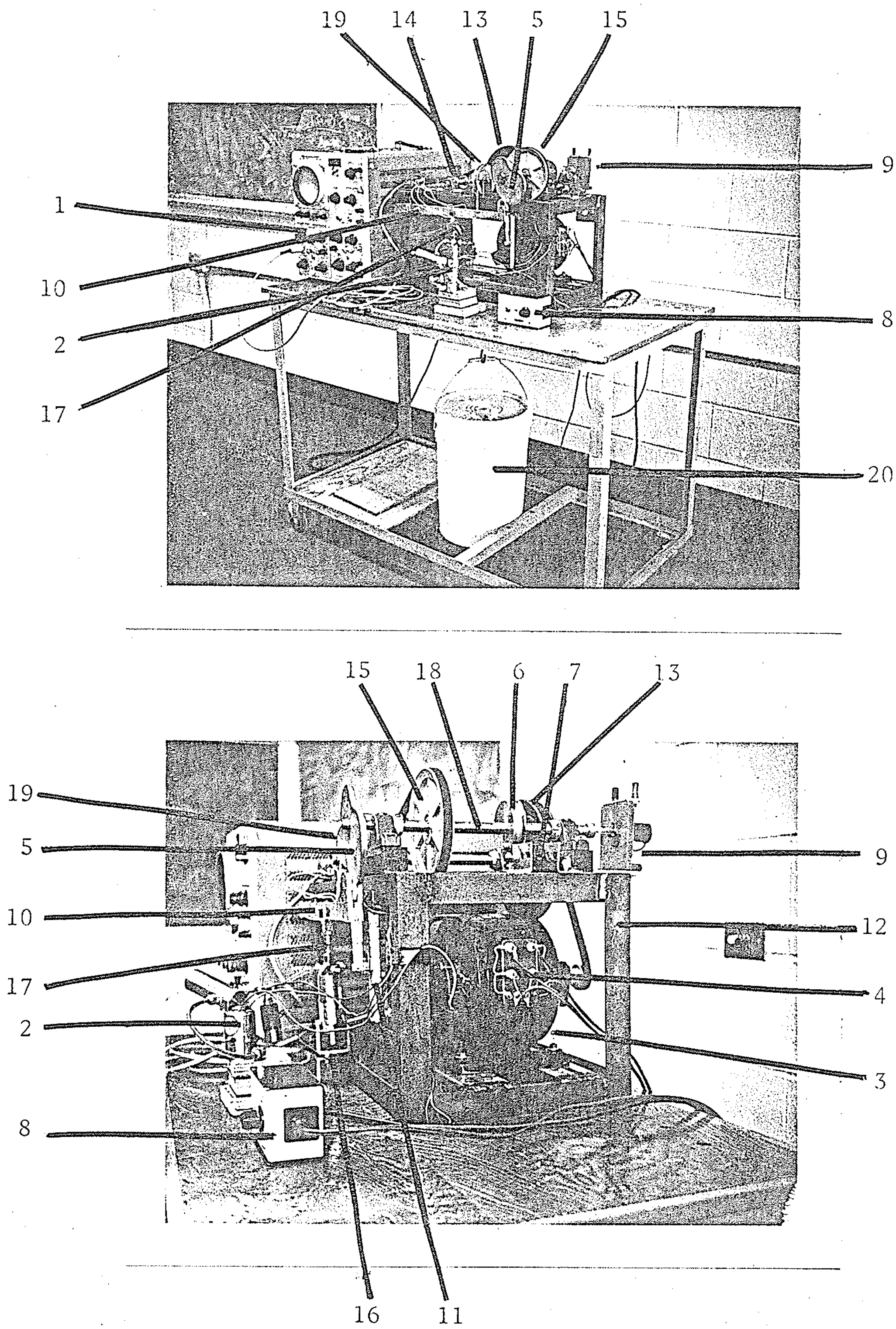


Diagram of the Test Jig

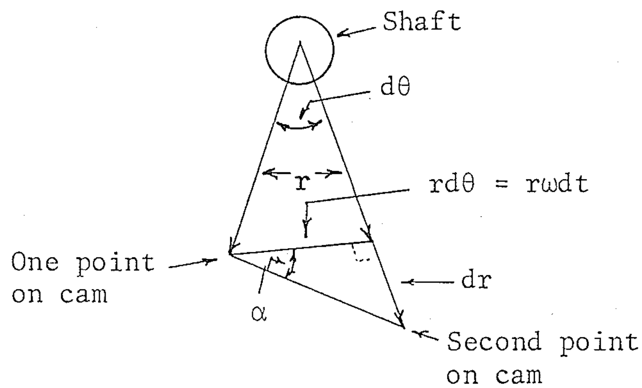
via a pivot. The other end of the lever rides the cam with a 1" diameter nylon bushing. In order to keep the driving lever on the cam, a heavy weight (20) is attached to the lever over a pulley (19).

Now, as the piston of the damping cylinder is moved up and down, the valves have to be switched on or off at specific times in the walking cycle (Fig. 12). In order to do this accurately, a second cam (6) was added to the drive shaft (18). This cam actuates two micro-switches (7) at the correct times in one walking cycle which energizes the correct valve combination.

When the system was working properly, measurements were taken. Force was measured by using two strain gages (17) mounted on opposite sides of the piston rod in the damping cylinder, and then have leads running from them to the Type Q Unit (1) which converts strain into a force (after calibration). The RPM of the drive shaft was monitored via the tachometer (9) and the velocity of the piston rod was found by differentiating the output of the linear potentiometer (11) attached to the end of the driving lever (10).

APPENDIX 4

To prove that flow varies directly as angular velocity of the shaft, assume an incremental angular shaft displacement, $d\theta$, resulting in an incremental displacement change, dr, of the end of the driving rod due to the cam.



$rd\theta$, can be considered as a straight line segment because $d\theta$ is assumed small. Therefore,

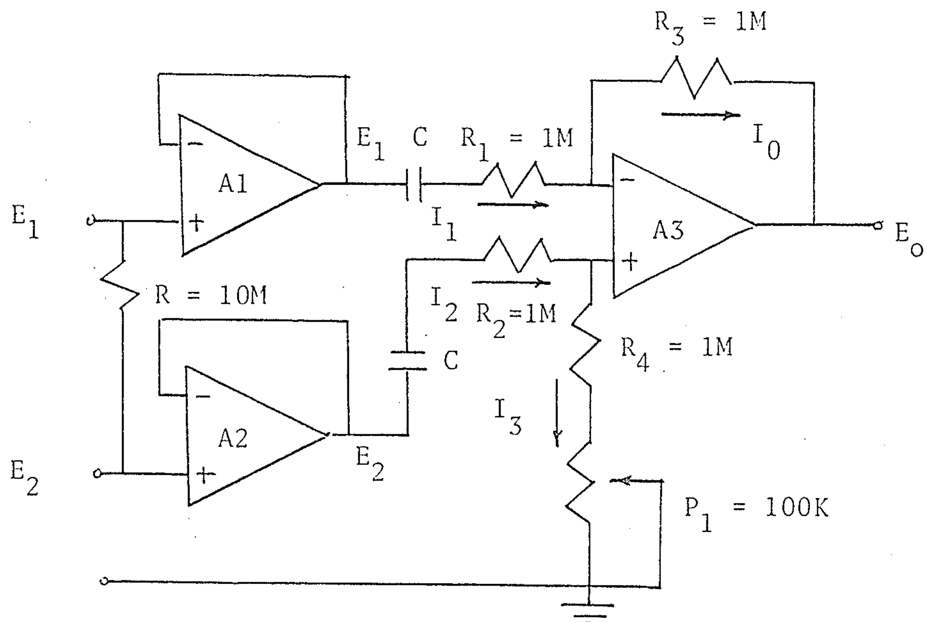
$$\tan \alpha = \frac{dr}{r\omega dt}$$

$$\frac{dr}{dt} = v = r\omega \tan\alpha .$$

α can be associated with a cylinder flow rate, and since α and r depend only on the geometry of the cam, it can be seen that flow varies directly with angular velocity ω.

APPENDIX 5

The following is a more detailed discussion of the function of some of the integrated circuits used in the EMG control electronics as discussed in Chapter 4.



A1 and A2 are both μA741 operational amplifiers connected as voltage followers, therefore having unity gain. Their input impedances are given in the specs as 400MΩ in this configuration, thus making the differential input impedance 10MΩ. A3 is a CA3078 operational amplifier. Its differential voltage gain is derived as follows:

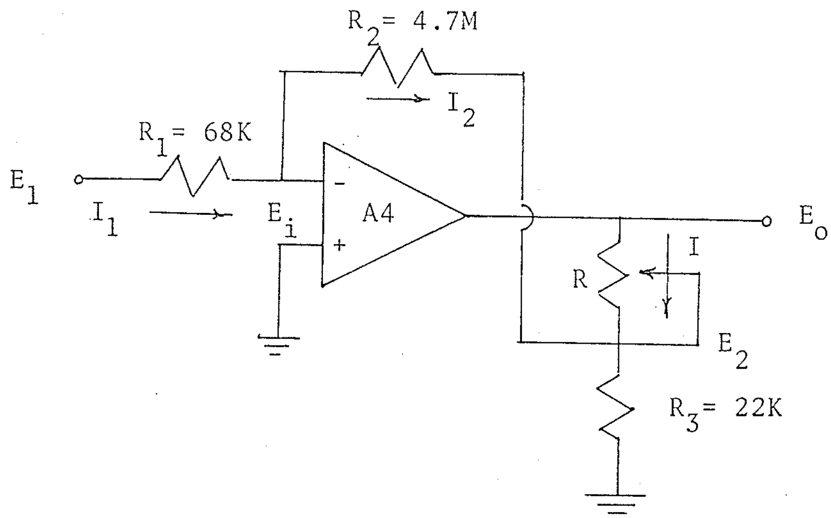
$$I_1 = I_o \Rightarrow \frac{E_1 - E_f}{R_1} = \frac{E_f - E_o}{R_3} \qquad \text{(defn. of an op amp)}$$

$$E_o = E_f\left(1 + \frac{R_3}{R_1}\right) - \frac{R_3}{R_1}E_1$$

Since $E_f = \frac{R_3}{R_2 + R_3}E_2$ and $R_1 = R_2 = R_3 = R_4$

$$E_o = \left(\frac{R_3}{R_2 + R_3}\right)\left(1 + \frac{R_3}{R_1}\right)E_2 - \frac{R_3}{R_1}E_1 \Rightarrow \frac{E_o}{E_2 - E_1} = 1$$

P_1 is just a pot which balances the impedances and improves CMR.



A4 is another CA3078 operational amplifier.

$I_1 = I_2$ (defn. of op amps)

$$\frac{E_1 - E_i}{R_1} = \frac{E_i - E_2}{R_2}$$

$$E_2 = \frac{R_3}{R + R_3} \qquad \text{if } I_2 << I$$

$$\frac{E_1 - E_i}{R_1} = \frac{E_i - R_3 E_o / (R_2 + R_3)}{R_2}$$

Since $E_o = -A\,E_i$, $E_i = -\frac{E_o}{A}$

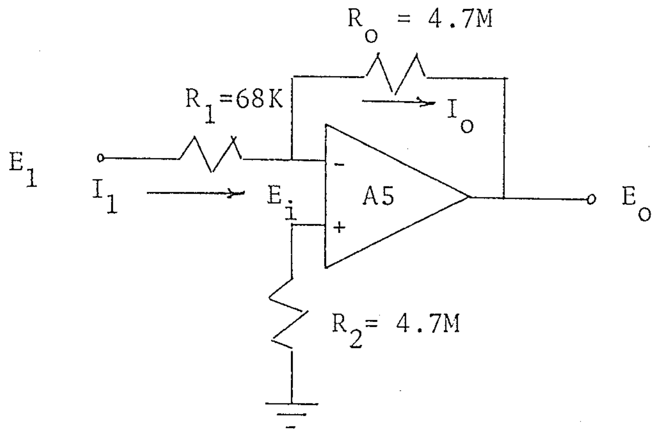
and as $A \to \infty$, $E_i \to 0$

Therefore, $\frac{E_1}{R_1} = -\frac{R_3\,E_o}{R_2\,(R + R_3)}$

$$\frac{E_o}{E_1} = -\frac{R_2\,(R + R_3)}{R_1 R_3}$$

Thus , when $R = 0$, gain = - 69.2

when $R = 47K$, gain = - 217

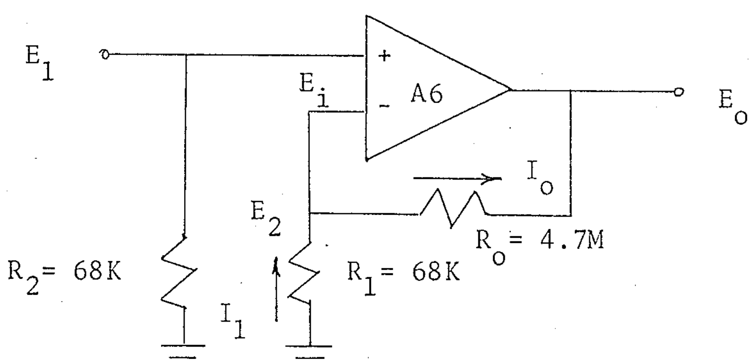


$I_1 = I \quad \Rightarrow \quad \frac{E_1 - E_i}{R_1} = \frac{E_i - E_o}{R_o}$ (defn. of op amps)

As shown previously , as $A \to \infty$, $E_i \to 0$

Therefore $\frac{E_o}{E_1} = - \frac{R_o}{R_1} = - 69.2$

R_2 merely controls output offset by balancing the impedances at the inputs, reducing the current sucked by the negative input from the output.

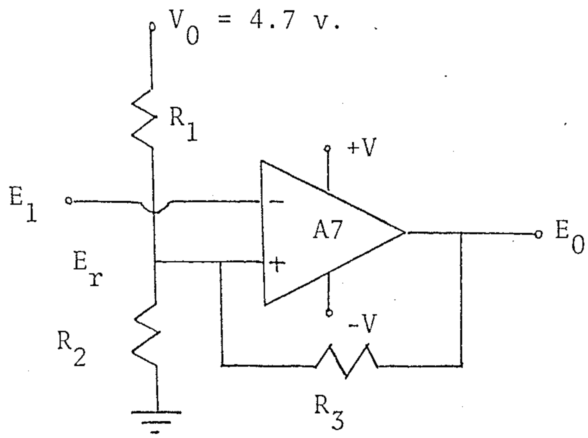


$I_1 = I_o \quad \Rightarrow \quad \frac{E_2}{R_1} = \frac{E_o - E_2}{R_o}$ (defn. of op amps)

$E_2 = E_1 + E_i$ and as $A \to \infty$, $E_i \to 0$

$\frac{E_1}{R_1} = \frac{E_o - E_1}{R_o}$ and $\frac{E_o}{E_1} = \frac{R_o + R_1}{R_1} = 70$

Again R_2 exists to balance input currents to avoid output offset levels.



Comparator circuit.

Normally E_1 is 0 , $E_r > 0$, therefore with positive feedback, the output saturates to a voltage determined by the biasing voltage of the comparator (+V , -V). In this circuit V = 6 v., and at rest the comparator voltage $E_0 = +6$ v. E_r is thus:

$$E_{on} = \frac{R_2 V_o}{R_1 + R_2} + \frac{R_2 V}{R_2 + R_3}$$

If E_1 exceeds E_r then the output is negative and with positive feedback, saturates to the negative bias voltage ie. $E_o = - V$. This, however, changes E_r so the comparator will not turn off until

$$E_{off} = \frac{R_2 V_o}{R_1 + R_2} - \frac{R_2 V}{R_2 + R_3}$$

Hysteresis is established because R_3 is not much greater than R_1. Thus knowing V_o, selecting an R_1, and then deciding upon the turn on and turn off voltages, R_2 and R_3 can be solved for by the equations:

$$R_2 = \frac{(E_{on} + E_{off})\ R_1}{2V_o - (E_{on} + E_{off})}$$

$$R_3 = \frac{R_2(2V - E_{on} + E_{off})}{E_{on} - E_{off}}$$

In the three cases here, $R_1 = 2.2M$, and $V = 4.7v$. The turn on and turn off voltages are:

comparator A7,	.10v.	and	.02v.
comparator A8,	.20v.	and	.05v.
comparator A9,	.30v.	and	.10v.

Thus R_2 and R_3 for:

comparator A7, are	28.5K	and	4.25M
comparator A8, are	60K	and	4.75M
comparator A9,	97.8K	and	5.76M.

Characteristics of the operational amplifiers used:

	CA3078	µA741
Supply voltage	± 14v.	± 18v.
Differential input voltage	± 6v.	± 30v.
DC input voltage	± 14v.	± 15v.
Output short circuit duration	No limit	No limit
Device dissipation	500 mW.	310 mW.
Temperature range operating	0 to 70°C	0 to 70°C
storage	-65 to 150°C	-55 to 125°C
Input resistance	.87 MΩ	2.0 MΩ
Output resistance	800 Ω	75 Ω
Transient response Rise time	2.5 µsec	.3 µsec

APPENDIX 6

Viscosity of several fluids tested given in Saybolt Universal Seconds and centistokes. The room was 75°F. and the fluid was heated to 100°F.

1 centistoke (cs) = .01 stokes

1 stoke = cgs. unit of kinematic viscosity

Conversion from Saybolt Seconds to stokes follows:

for $t < 32$	tables were used	
for $32 < t < 100$	$\nu = .00226t - 1.95/t$	stokes
for $t > 100$	$\nu = .00220t - 1.35/t$	stokes

	Saybolt Seconds	centistokes
Hydraulic fluid	110.6	22.95
Octoil 's'	65.7	11.86
Brake fluid	52.0	8.00
Kerosene	29.9	1.089
Water	25.6	1.0038

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