A PROSTHESIS FOR
ABOVE-KNEE AMPUTEE RUNNERS

TO LESLIE,

for her infinite love and encouragement

A PROSTHESIS FOR ABOVE-KNEE AMPUTEE RUNNERS

by

BARTON WILLIAM IRWIN, B.A.Sc., B.Ed.

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AUTHOR: Barton William Irwin, B.A.Sc. (University of Waterloo)
B.Ed. (University of Western Ontario)

SUPERVISOR: Professor W.R. Newcombe

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ABSTRACT

Although a number of above-knee amputees have expressed a desire to participate in recreational activities involving running, no currently available lower limb prosthesis has proven adequate in allowing amputees to achieve a natural, efficient, one-to-one running gait. Until recently, amputee runners such as Terry Fox and Steve Fonyo have adopted a variety of asymmetrical gaits, although the Terry Fox Jogging Prosthesis has allowed some amputees to achieve an inefficient one-to-one running pattern.

The objective is to design a conservative running prosthesis which will functionally imitate the intact limb during running activities. The prosthesis performance criteria were established for both stance and swing, based on an examination of non-amputee running biomechanics.

The prosthesis incorporates a shank unit assembly which linearly compresses upon heel-strike, absorbing the impact energies in a helical coil compression spring, and then uses a ratchet device to store these impact energies throughout the stance phase. In late stance, the natural dorsi flexion of the prosthetic foot initiates the release of the stored energies, propelling the amputee upward and

forward into the swing phase.

It is recommended that a comprehensive biomechanical gait analysis be performed on the prosthesis' operation to allow for optimization of its configuration and performance.

In conclusion, the features of this prosthesis will allow above-knee amputees to achieve a more natural, one-to-one running gait and participate more actively in activities involving running.

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I will miss my friends and colleagues at McMaster whose advice, encouragement and occasional good-natured ridicule made the good days enjoyable and the bad ones bearable.

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CHAPTER 1

INTRODUCTION

In recent years, several courageous Canadians have managed to emphasize the obstacles faced by the handicapped in our society, as well as their ability to triumph over them. Terry Fox, Steve Fonyo, and, most recently, Rick Hansen, have dramatically demonstrated the need for the disabled to lead rich, full lives in spite of their handicaps.

The loss of a limb, aside from the obvious physical impact, is also accompanied by serious emotional adjustments. In fact, the grief reaction following the loss of a limb has been compared to that of the loss of a spouse [54]. This impact can be lessened, however, if the amputee is able to obtain a reasonable degree of functional normalcy. Galway, [20], emphasizes that the ability of an amputee to participate as a coequal within a peer group can "contribute greatly to a sense of self-worth and independence". To this end, then, although no presently available prosthesis, "no matter how well engineered, fitted or constructed, can either functionally or cosmetically" [39] duplicate the missing limb, the field of prosthetic engineering continues to strive to develop ideal body component replacements [30].

Several surveys of lower limb amputees, [28], [40], have shown that most amputees do not resume a completely normal lifestyle, and many modifications are made. In many cases this failure to return to a normal lifestyle is a direct result of restrictions placed on the amputee by the functional limitations of the prosthesis. However, even within these limitations, the majority of amputees are eager to participate in activities that they routinely enjoyed prior to the amputation.

In a survey by Kegel et al., [28], over sixty percent of the lower extremity amputees participated in one or more recreational pursuits, with the most popular including fishing, swimming and dancing. A guide to sports and recreation for lower limb amputees, also by Kegel, [27], lists twenty-five activities, including skateboarding, mountain climbing, and sky diving. An incredibly diverse collection of specialty aids has been developed to allow amputees to participate in such activities. Swimming legs, and downhill ski equipment, including outriggers, are commonly dispensed. When amputees were asked what they could no longer do as a direct result of their amputation, however, the major problem area appeared to be running.

Although some might question the need for the lower-limb amputee to run, this activity has benefits beyond those associated with most recreational activities. Kegel et al., [28], states that "although many people did not need or want to run in their daily activities, this ability is extremely important in a 'fight or flight' situation". In addition, with society's increased awareness of physical fitness, there is a growing number of amputees who desire to run.

Running ability has developed quite successfully among below-knee (B/K) amputees, primarily due to two factors. The below-knee amputee's natural knee joint is intact, and thus the amputee is able to generate the complex bio-kinematic pattern of the lower limb necessary during running. Also, advances in prosthetic foot and socket design have enabled the amputee to run in relative comfort.

The goal of providing a prosthesis to above-knee amputees that would allow for a physiologically correct running gait has been less successful. The characteristic "hop-hop-step" running pattern of both Terry Fox and Steve Fonyo is representative of running patterns among above-knee amputees. In this form of gait, one step is taken on the prosthesis, followed by two "hops" on the sound limb. This occurs since currently available prostheses are designed for walking, and the associated motions in the same speed range. During running, there is insufficient time during the swing phase of the gait cycle to bring the leg forward into the position appropriate for weight bearing, or stance, thus necessitating the second step on the sound limb.

This form of "running" places high physical stresses on the amputee and is both functionally and cosmetically unacceptable. The objective of the following work is to develop a prosthesis which will allow active above-knee amputees to achieve a natural, and relatively comfortable, running gait. In the following pages, a detailed examination of the present state of above-knee prosthesis technology will be summarized, followed by a determination of the necessary design criteria for the new prosthesis, its subsequent design and development, and finally the conclusions and recommendations for future research.

CHAPTER 2

BACKGROUND

Medical amputations have been traced to the beginning of mankind, with the very first probably performed to remove gangrenous or severely damaged limbs. The manufacture of prostheses soon followed since "amputational surgery logically led to the need for a replacement or substitute for the amputated part so as to restore function" [45]. Rang and Thompson, [45], explain that "prosthesis is a Greek word meaning 'addition'". No record exists of when lower-limb prostheses were first used, but they most probably consisted of "wooden pegs, or splints and crutches" [45]. One of the oldest known artificial legs is made of wood, iron, bronze and leather and dates back to 300 B.C.. Lower-extremity prostheses remained functionally simplistic throughout the middle ages with little consideration for aesthetics [45]. The development of modern principles of lower extremity prosthetics, such as the suction socket, began in the eighteenth and nineteenth centuries, with many principles still in use today.

The modern history of artificial limbs, however, began with the completion of World War II. As stated by Mital and Pierce, [39], "the improvement of war time medical services had resulted in the survival of a large number of soldiers with amputations". Following the war, both in the United States and Canada, the first organized attempt to produce functional, inexpensive prostheses was undertaken. Since that time, substantial progress has been made in all facets of prosthesis development. Research continues, however, since the function and appearance of the missing limb has yet to be adequately duplicated.

The initial emphasis of post-war research was placed on engineering and design of prosthetic hardware. Wilson, [55], indicates, however, that "it soon became apparent that existing knowledge of human locomotion was insufficient to establish valid design criteria". Gait laboratories were established, and have subsequently provided significant data concerning the biomechanics of human gait. The result of this research has been the development of a diverse assortment of prosthesis designs and production methods. This variety allows the prosthetist greater flexibility in prescribing functional prostheses to the greatest number of patients.

One of the greatest successes of post-war research has been the development of the modular concept of prosthetic production. Traditional prosthetic raw materials have been wood and sheet metal, with the production of each prosthesis requiring many hours of hand work by skilled prosthetists [24]. This method of manufacture created delays in receipt of the prosthesis by the amputee, as well as difficulty in modifying the final prosthesis if deemed necessary.

Modular assembly is the name which has been given to a system of prosthesis production from pre-manufactured, internationally standardized components. The modular system introduces much needed flexibility into the field of prosthetics. The use of modular components provides a means of altering the configuration of the prosthesis without major structural modifications. It therefore becomes possible to adjust the alignment of the prosthesis during the course of a patient's treatment if that should prove desirable. Also, as an increasing number of manufacturers modify their products to conform to the modular system, the number of choices available to the prosthetist and patient increase. During the course of treatment, an infinite number of combinations and permutations of devices and alignments may be used until the "optimum" configuration is determined [24].

There are also several economic advantages to the use of the modular assembly system. With pre-manufactured standardized components, manufacturing may be done more efficiently and in economic quantities. The number of man-hours to produce a finished prosthesis is also significantly decreased. In fact, the critical factor in determining the time necessary to produce a prosthesis becomes the time necessary to produce a satisfactory socket. It then becomes a real possibility to provide a patient with a finished prosthesis in a matter of hours, and at a cost less than that of a traditional prosthesis [24].

Ross, [46], lists the basic components of an above-knee modular system as (see Figure 2.1):

(a) Socket and attachment hardware
(b) Knee mechanism, including flexion/extension control and alignment coupling
(c) Shin tube assembly
(d) Foot and/or ankle assembly
(e) Cosmetic cover

Although an above-knee amputee's ability to run has not been the primary thrust of prosthetic research in recent years, technological advances in the design of each of the above components are significantly improving amputee gait and walking comfort, and these advances should in turn prove beneficial in the development of a running prosthesis for above-knee amputees.

In the following sections, a review of the technological status of each of these components is presented.

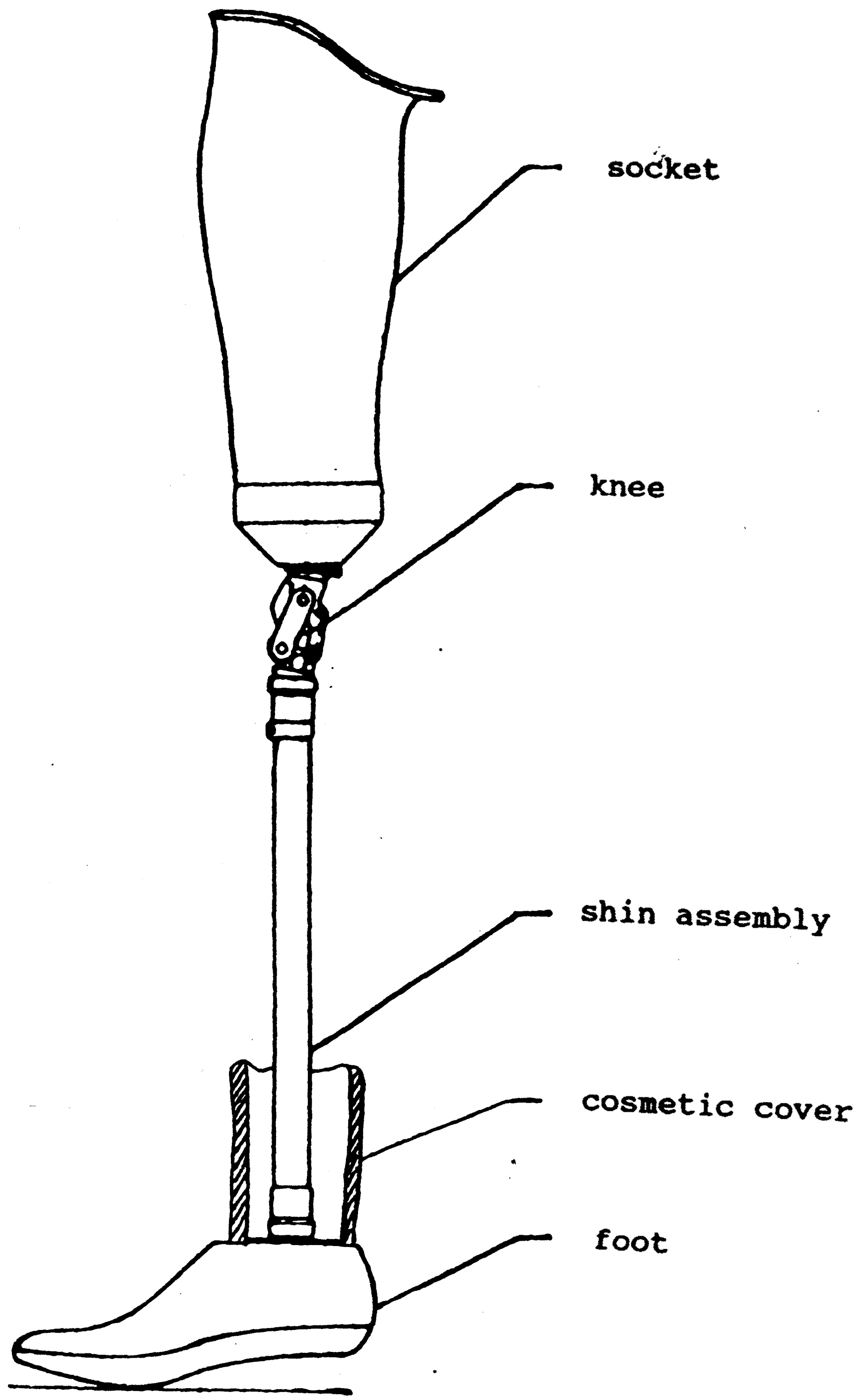


Figure 2.1 : Above-Knee Modular Prosthesis

2.1 Socket

The interface between the amputee and the socket represents the most critical component of any prosthetic system. As stated by Lyquist, [34], no prosthesis, regardless of design, will prove satisfactory to the user "unless the body weight is supported in a comfortable manner, and the user can adequately control the function of the prosthesis through that interface".

Sockets must be individually designed for specific patients, and the design involves a number of physiological and biomechanical considerations. Stability of the stump within the socket must be high, as the amputee must exert considerable flexor and extensor moments through the stump to control the prosthesis and provide stability during gait. Relative motion between the skin and socket must be minimized to avoid tissue damage. Similarly, localized high unit pressures should be avoided [2]. The stump must be fitted into the socket in such a way that the muscles are able to function to maximum capacity. When contracting, the muscles shorten and increase in girth, and these muscles cannot be restricted without causing pain and loss of function, and thus the socket must provide sufficient space for muscle contraction [33]. To date, socket design and manufacture remains a time consuming "art", with success dependant on the manual skill and experience of the prosthetist.

Although sockets have traditionally been carved or machined from wood, with a plaster replica of the stump as a visual guide, the majority of sockets are now produced by laminating thermosetting resins and fiberglass over a positive plaster mould of the stump that has been appropriately modified by the prosthetist to provide areas of direct support and relief.

There have been suggestions that socket production could be standardized. The initial concept was that a series of standard sockets be made available, similar to different sizes of prosthetic feet, and that these sockets be dispensed to the majority of amputees with only minor modifications [18]. This concept has been expanded with the recent advances in the areas of computer-aided design (CAD) and computer-aided manufacturing (CAM).

Computer-aided socket design represents the future of socket production. Several systems have been developed, [15], [48], that allow for direct measurement of the stump's contours, either by direct tactile measurement or by measurement of the stump's silhouette. The positive image of the stump can then be appropriately modified on-screen by the prosthetist, similar to the modifications that would be manually performed on a plaster casting of the stump. The final design is then manufactured utilizing a computer-numerically controlled (CNC) milling machine.

The use of this technology should improve the quality of socket design in several ways. First, using CAD technology, the results of the fitting procedure would be reproducible. Also, if the patient expresses observations on a newly produced socket, it will be possible to directly compare the configuration of the new socket with previous ones. This quantitative data will eventually accumulate, allowing statistical determination of the most successful design practices. This could in turn lead to a universally standard approach to socket design to maximize socket comfort and functionality. Saunders et al., [48], summarizes the impact of CAD/CAM technology by stating "the automation of the shape management process will capture the important aspects of the current artisan methods and overcome their inefficiencies".

Another logical extention of the above technology would be the development of equipment which could physically simulate the fit of the socket for the patient before the final manufacturing is initiated [31].

The two types of sockets most commonly dispensed for above-knee amputees are the open-ended socket and the total-contact suction socket. Both are rigid walled sockets with suspension being maintained both by muscle tension and by the seal created by the flesh of the stump within the socket [33].

The major difference between the two types of sockets is the method of weight bearing. In the open-end socket, there is a distinct posterior brim, which is situated directly below the ischial tuberosity. The greater part of the body weight is carried through this interface, and to some degree through the gluteus maximus. There is no contact of the distal portion of the stump. In the total-contact socket, there is an attempt to provide uniform pressure between the stump and socket over the entire stump surface, including the distal portion. The ischial tuberosity is utilized, however to provide some weight bearing. It has been shown that the action of the total-contact socket during walking can improve circulation in the stump [33].

Other innovations in the area of socket design include the ISNY (Iceland, Sweden, New York University) socket system and the Contoured Adducted Trochanteric-Controlled Alignment Method (CAT-CAM).

In traditional sockets the body weight is carried through the rigid walls of the socket to the prosthesis and then to the floor. The ISNY socket is an ischial load bearing socket consisting of a posterior brim and a thin skeletal strut arrangement to transfer the load to the knee. This allows the remainder of the stump to be contained in a soft, flexible thermoplastic. A limited study has shown that the ISNY socket increases prosthetic comfort with no accompanying increase in stump-socket instability [32].

The CAT-CAM socket system has introduced totally new theories on load bearing and socket shape. Traditionally, quadrilateral sockets have been designed quite narrow in the anterior-posterior axis, (Figure 2.2), to force the ischial tuberosity into position on the posterior brim, while the medial-lateral dimension was quite liberal. The designers of the CAT-CAM system felt that the accepted standard medial-lateral dimension allowed excessive abduction of the femur within the socket during stance, resulting in asymmetric gait (Figure 2.3). In the CAT-CAM system, the medial-lateral dimension is decreased, while increasing the anterior-posterior dimension (Figure 2.2). Also, the posterior and medial walls of the socket are extended to encompass the ischial tuberosity (Figure 2.4). This ensures proper weight bearing placement of the ischial tuberosity by "locking" it into position while providing excellent lateral stability for the femur. In addition, the increased anterior-posterior dimension allows for greater functionality of the flexion and extension musculatures. Limited testing has indicated greater comfort and ease of locomotion with the use of this design [47].

Although there has been very little research specifically directed at the problem of sockets for amputee running, developments such as the use of computer design technology, and the ISNY and CAT-CAM socket systems will directly aid in the development of a socket suitable for running. The use of computer design for data collection will allow for the development of universal design theory. The

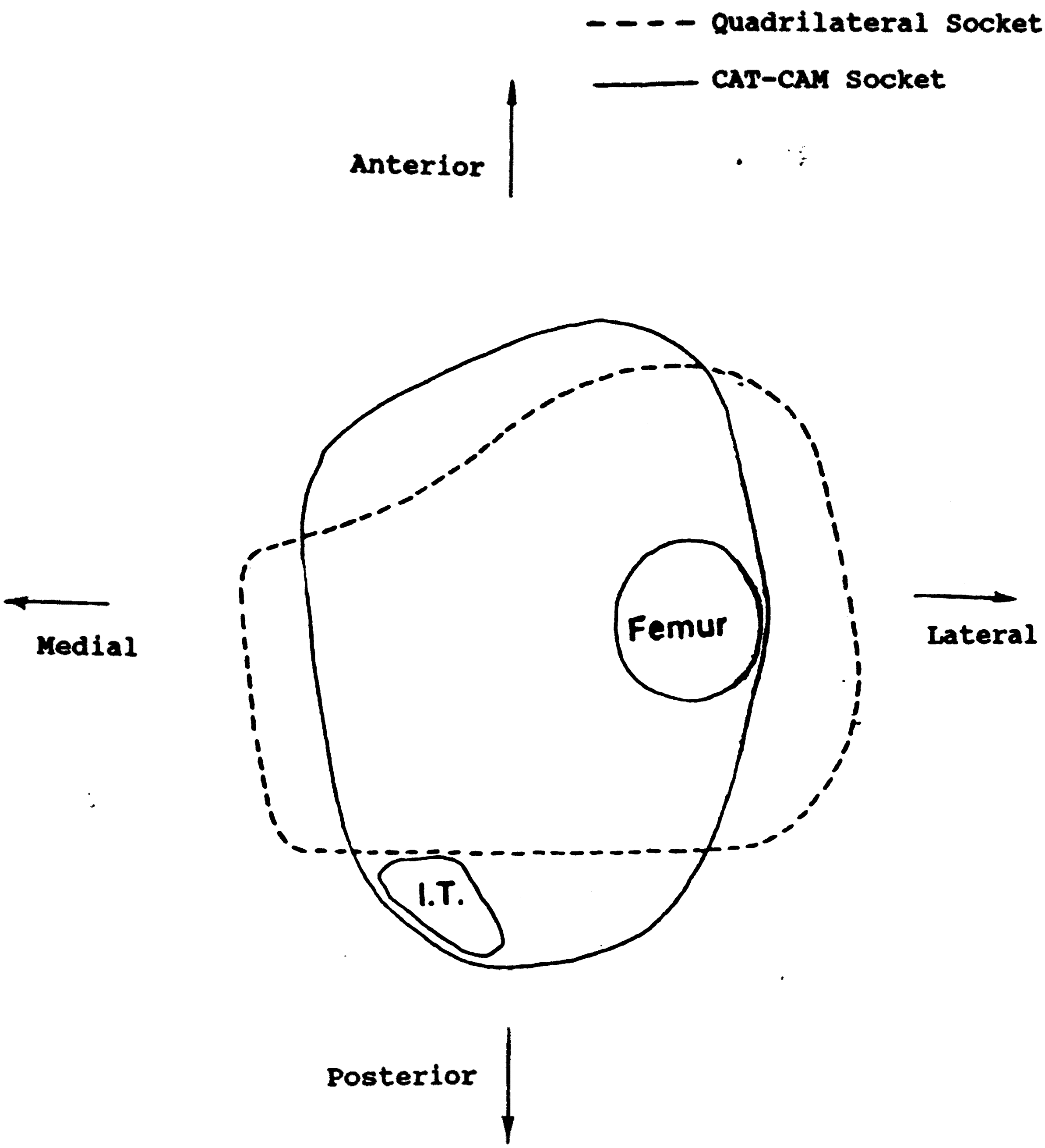


Figure 2.2 : Comparison of CAT-CAM and quadrilateral sockets in transverse view [47]

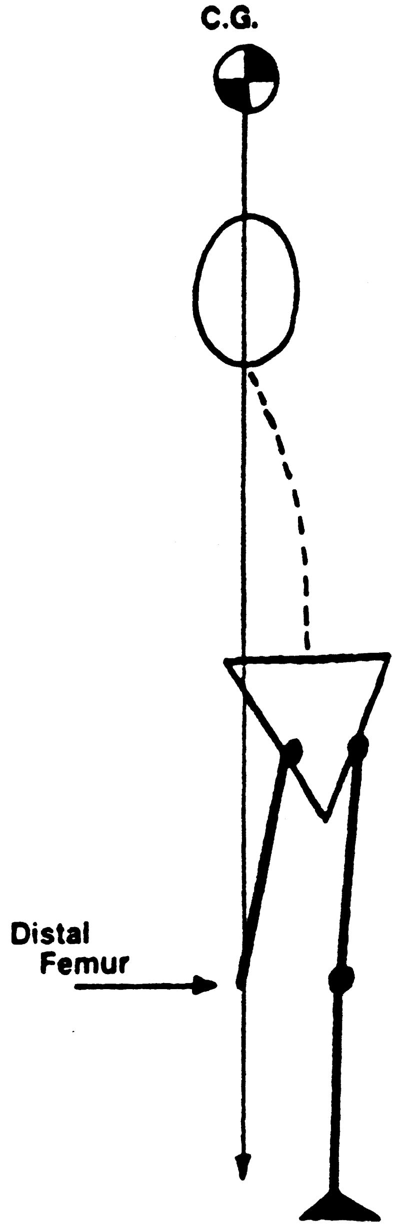


Figure 2.3 : Excessive femur abduction within quadrilateral socket [47]

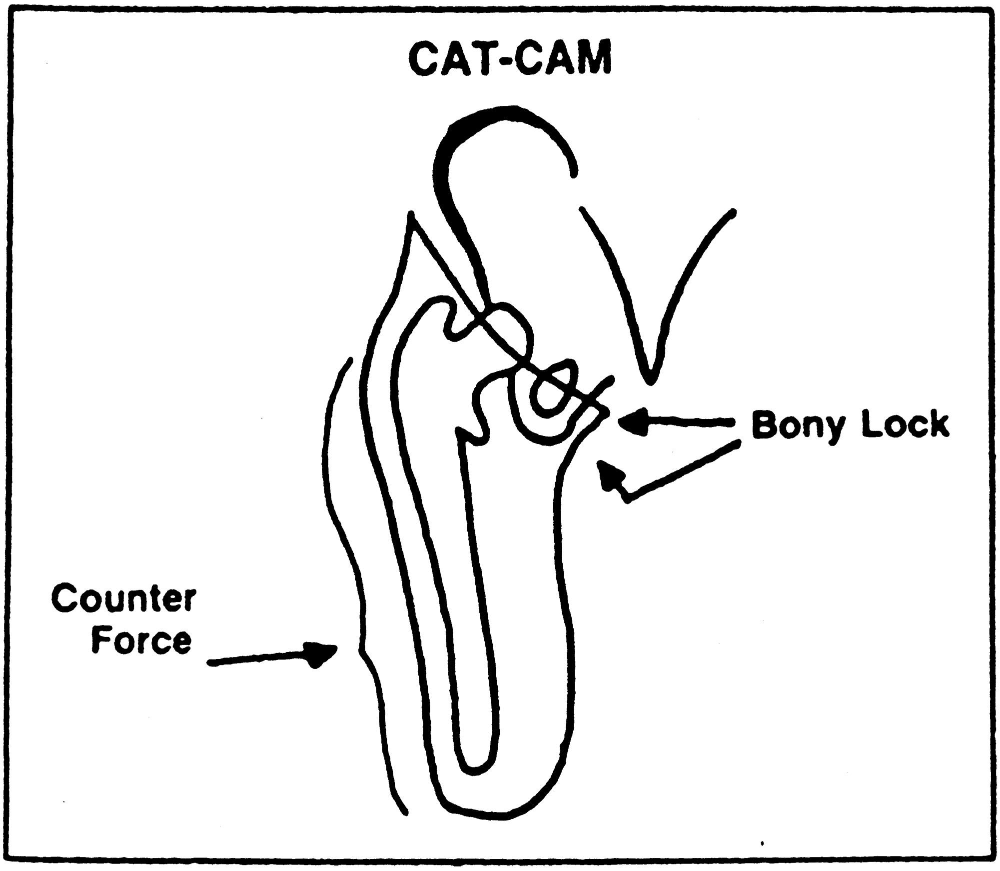


Figure 2.4 : "Locking" of ishial tuberosity
within CAT-CAM socket [47]

ISNY and CAT-CAM sockets, which provide greater user comfort while allowing for greater muscle functionality, offer alternatives to traditional sockets for use in running prostheses.

2.2 Knee Mechanisms

Extensive research has been done in the area of prosthetic knee function and design. The knee is second only to the biomechanical interface between the stump and socket in terms of effect on prosthetic functionality and is the primary determinant in the efficiency and kinematic correctness of the amputee's gait.

An artificial knee joint should provide two basic functions, stance phase stability and swing phase control [44]. The knee must be designed such that it will not collapse during the weight bearing phase of the walking cycle (stance), yet will allow flexion of the knee at toe-off with relative ease. During the swing phase of the walking cycle, it is desirable that the knee simulate the cumulative agonist-antagonist actions about the knee normally provided by the quadriceps and hamstring musculature in the intact knee [42].

2.2.1 Stance Phase Stability

Stance phase stability can be achieved by a number of different means, both voluntary and involuntary. Although it is always desirable to maximize the amount of voluntary knee control, (as this results in the "smoothest and most effortless gait" [43]), the amount of voluntary control made available will be dictated by the amputee's age, co-ordination and the general condition of the stump (including length, musculature and range of motion).

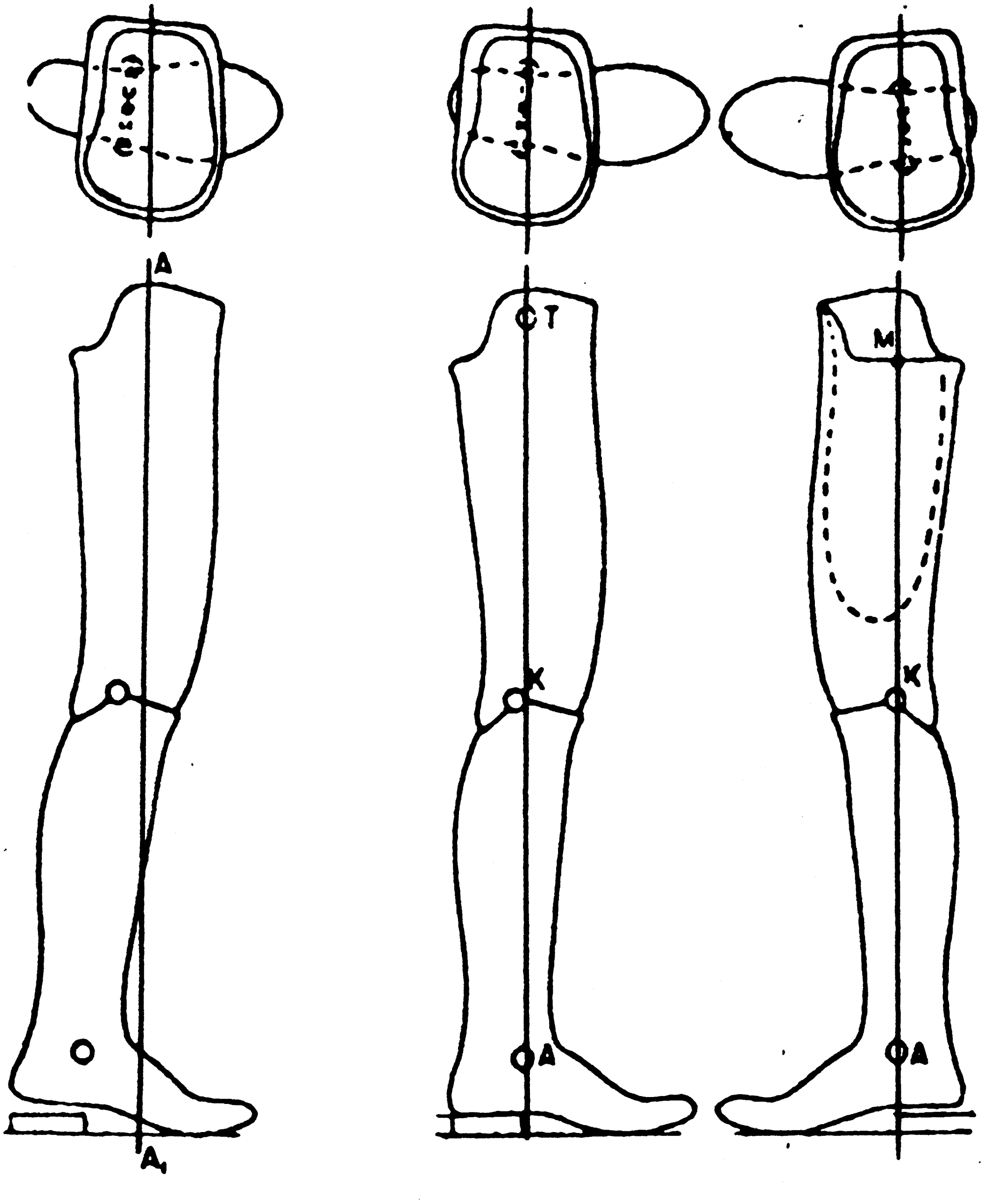
The most extreme solution to knee instability is the use of a knee which remains locked in a fully extended position throughout the walking cycle, but can be unlocked to allow the amputee to sit. This, of course, leads to a very unnatural and awkward gait, and is rarely necessary. A more common alternative, especially for elderly patients, are so-called "safety knees", where a mechanical friction brake, or hydraulic cylinder is incorporated into the knee to provide active resistance to knee flexion during stance [16].

The primary principle employed in all above-knee prostheses, (regardless of other additional means to achieve stability), is summarized by Radcliffe, [42], and is referred to as "alignment stability" [42]. Alignment stability results from the placement of the knee axis posterior to the load line of the prosthesis, and thus forcing the knee into full extension during stance [43].

Alignment stability can be ensured by the use of several different alignment techniques (Figure 2.5). The German alignment technique is based on a plumb line co-incident with the mid-point of the lateral brim and the bisector of the length of the foot. The posterior distance of the knee axis placement is then dictated by the particular knee and foot in use. A similar plumb line technique is used in the United States, where the trochanter is used as the upper reference point and is in direct line with the ankle joint, with this known as the TKA line. The knee is then placed on or behind this line (typically six millimeters posterior) [42]. Errors in both systems can be induced by the difficulty in accurately locating the precise location of the contact point of the head of the trochanter. This has led to the development of a modified American system, (MKA), which uses the mid-point of the interior medial wall of the socket as the upper reference point, and thus the knee is located using the medial aspect of the prosthesis.

To prevent lateral movement (whip) of the foot during the swing phase, the knee axis is typically aligned in five degrees external rotation, with the medial end of the knee axis approximately six millimeters forward of the lateral end. Thus, for a typical, active amputee, the medial end of the knee axis can be placed directly on the MKA line [42].

Although designing the prosthesis such that the knee axis is posterior to the load line throughout stance will



(a) German (b) TKA reference (c) MKA reference

Figure 2.5 : Bench alignment systems [42]

ensure stance stability, it has the disadvantage of making the prosthetic knee difficult to flex under even a light load, and can result in poor gait and difficulty in negotiating uneven terrain [43].

The average above-knee amputee retains a reasonable amount of strength in his hip flexors and extensors, as well as an appreciable range of motion, and this musculature can and should be used for voluntary control of knee stability [43].

The characteristics of a given prosthetic knee are thus defined by the relation of the knee axis of rotation (i.e. the instant center of the thigh-shank rotation) to the load line, the braking moment or torque generated by the prosthetic knee and the hip moment (voluntarily supplied by the amputee) [42].

A simplistic mechanical model of this system is illustrated in Figure 2.6, where the combination of a force, P, plus a moment, M, is equivalent to a single force, Q, offset at a distance d. It is with these basic mechanical principles that Radcliffe develops the free body diagram of the combined stump and prosthesis, (immediately at heel strike), shown in Figure 2.7(a). This illustrates the floor reaction forces and moments exerted through and about the hip [44]. The relationship between the shear force, V, and the hip moment is determined by balancing the moments about the hip joint.

Magnitude

$Q = P$

$Q * d = M$

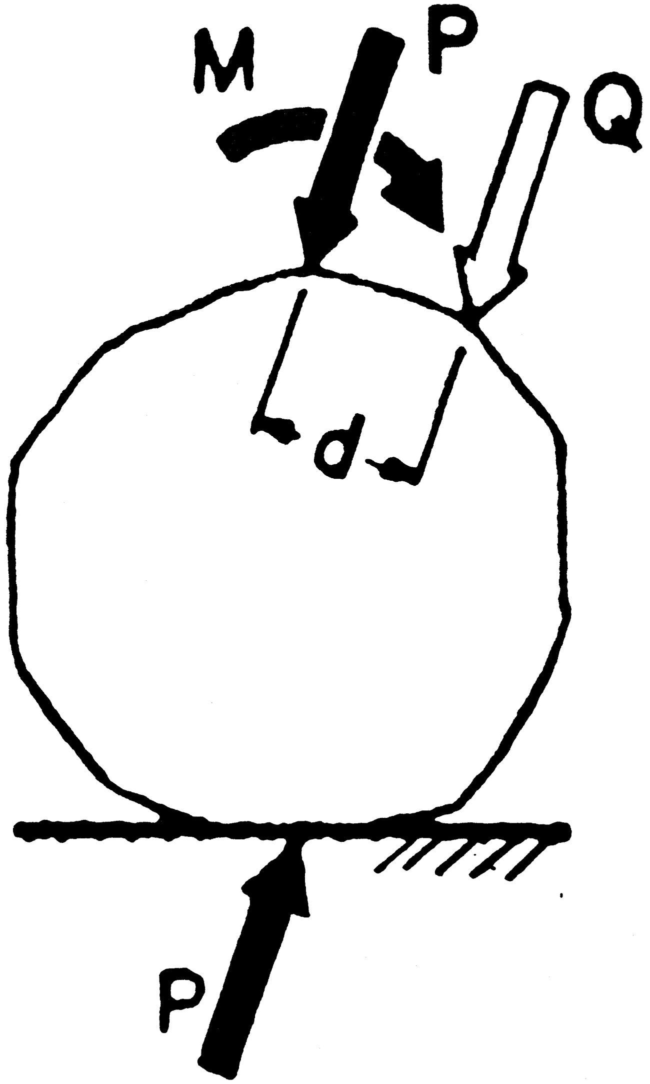


Figure 2.6 : Simple mechanical model of prosthesis loading [42]

Forces

Equivalent Load Line

(note posterior shift)

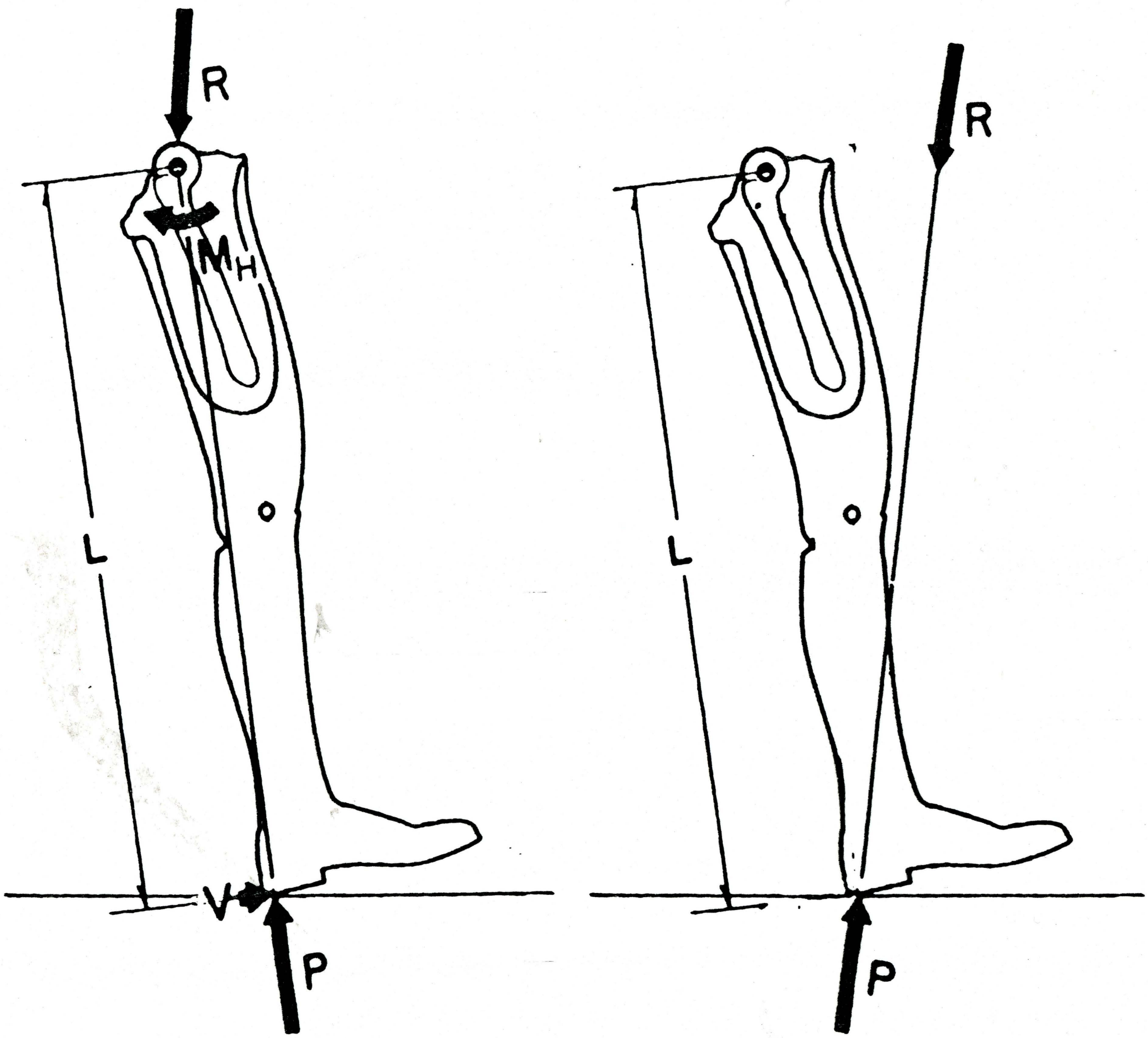


Figure 2.7 : Loading of prosthesis at heel strike [44]

Using these principles it can be seen that the load line at heel-strike or toe-off is related to the load, P, carried through the hip joint, and the hip moment, M_H. The hip musculature is able to control the line of action of the resultant force between the foot and the floor. With no hip moment, the load line would pass behind the knee, resulting in instability. With the application of a hip moment, an equivalent single force, Q, is generated at an offset distance, d, effectively moving the load line anterior to the knee, resulting in stability. An extension moment moves the load line anterior to the hip while a flexion moment moves it posteriorly. These principles can be directly applied to the fitting and placement of the prosthetic knee.

The earliest and simplest form of the prosthetic knee is the single axis, or simple hinge joint. Stability is achieved by properly locating the knee axis with respect to the hip/heel line.

In a healthy, active amputee who can generate a significant hip moment, the prosthetic knee will be stable at heel strike whenever the knee center is aligned behind the line PQ as shown in Figure 2.8(a). The knee can be flexed voluntarily to achieve toe-off if the knee's center of rotation is located ahead of the line P'Q' as shown in Figure 2.8(c). When these lines are superimposed, (Figure 2.8(b)), they define a common region, S, known as the "zone of voluntary stability" [23]. As stated by Radcliffe, in such

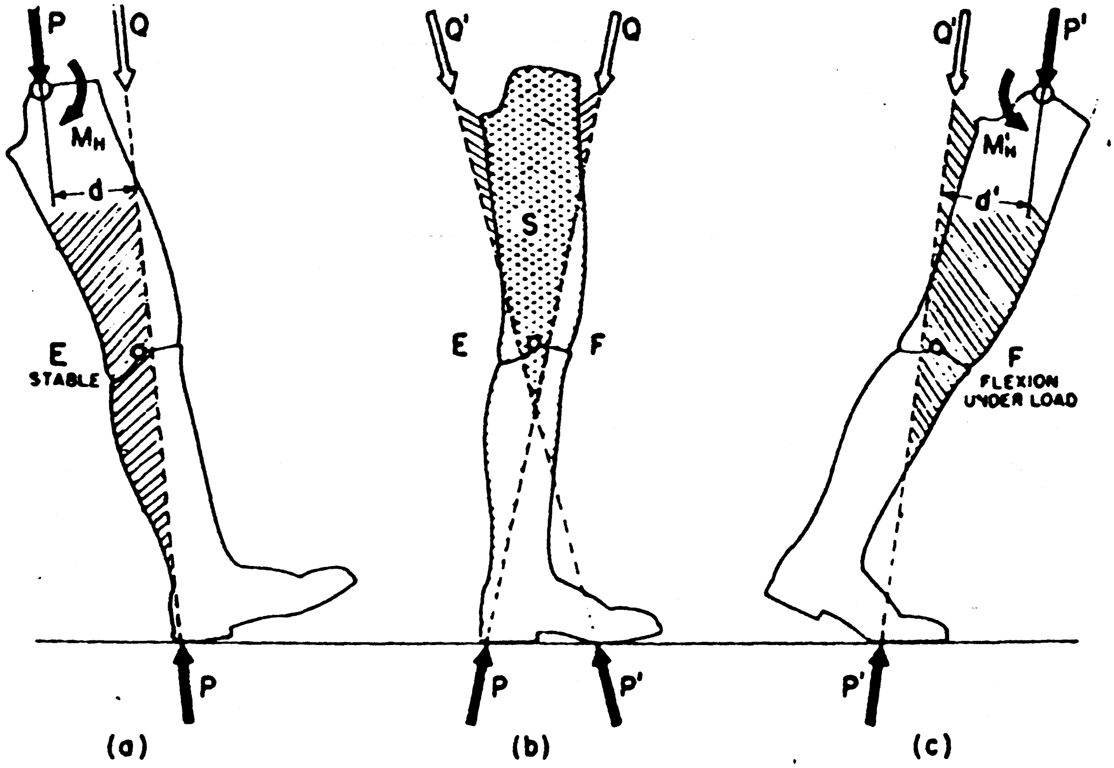


Figure 2.8 : Zone of voluntary stability [42]

case this zone allows a "considerable variation in the alignment of the prosthetic knee while maintaining stability at heel contact and ease of knee flexion at toe-off. In such a case, "the prosthetist aligns the knee as a compromise between the necessary knee stability at heel strike and the contradictory requirement that it should be possible to purposely cause knee instability prior to toe-off" [44].

A much more common situation is illustrated in Figure 2.9. In this instance, the amputee either has reduced hip moment capabilities or prefers to use his hip musculature at less than maximum capacity. Thus the zone of voluntary stability is dramatically reduced [42]. The prosthetist must then place the knee axis posterior to the load line to ensure stance stability at heel strike, at the expense that the knee is difficult to flex at toe-off.

Several types of prosthetic knees have been developed to overcome the inherent difficulties of the single axis knee. As mentioned previously, "safety knees" incorporate braking devices, (mechanical friction, hydraulic and pneumatic resistance), and are capable of producing a support moment at the knee during stance. This support moment can actually reduce the required hip moment for stance stability to near zero (Figure 2.10). These mechanisms are used for geriatric patients or for patients with minimal hip musculature. The drawback to such designs, however, is that the friction which acts to resist knee flexion during stance

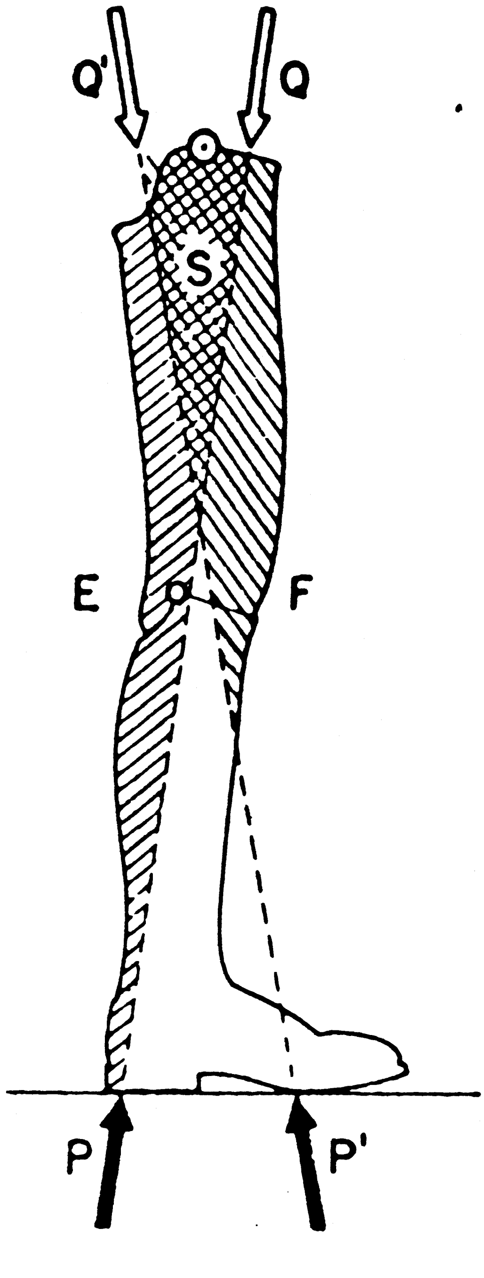
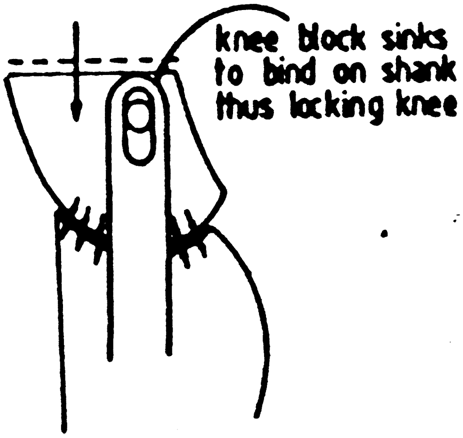
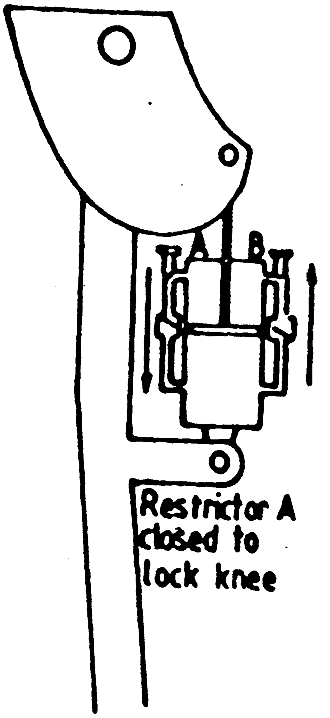


Figure 2.9 : Stability diagram for a typical above-knee amputee [42]



Friction stabilized knee



Resistance stabilized knee

Figure 2.10 : "Safety" knees [19]

will also tend to resist knee flexion at toe-off, and thus the hip moment necessary to initiate knee flexion is actually greater than that required in a non-brake mechanism.

More recently, a number of knee mechanisms known as polycentric knees, have been developed. In these devices, the instantaneous center of rotation of the knee is not a fixed point, but moves relative to the thigh as the angle of knee flexion increases or decreases [42]. This action duplicates the polycentric kinematic rotation of the normal human knee. The most common means of achieving the polycentric knee action is the use of a four-bar linkage (Figure 2.11).

As previously mentioned, for some amputees, (see Figure 8), the zone of voluntary stability is reduced, and is actually a significant distance above the knee axis, necessitating the placement of the knee for stance stability in a position that counteracts voluntary knee flexion. Radcliffe, [42], illustrates the advantage of the polycentric knee by stating:

> "... the ability of the amputee to control knee stability is influenced by the instant center of the knee rotation above the floor. A high knee center provides improved leverage for voluntary control of knee stability. Single axis knees provide little or no opportunity to make use of this fact because any significant change in the vertical position of the knee joint is cosmetically unacceptable. The polycentric knee, on the other hand, can be designed with the initial instantaneous center of rotation located above the usual knee joint and well within the

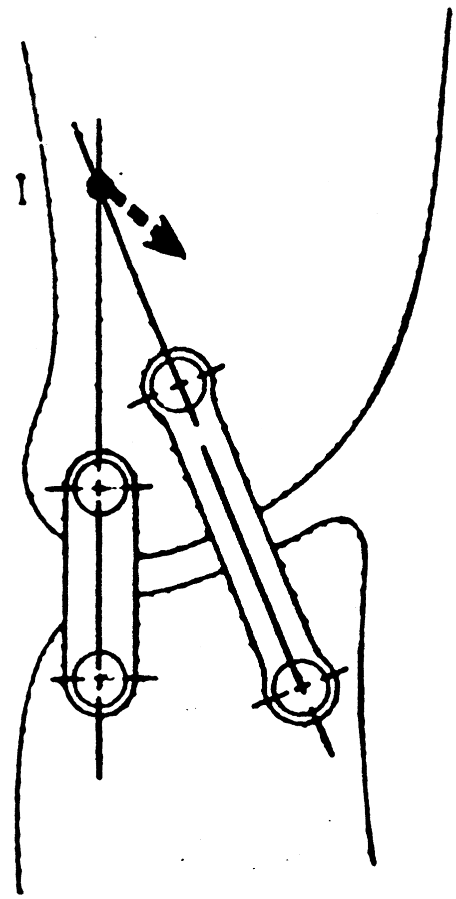


Figure 2.11 : Four-bar linkage polycentric knee [44]

> zone of voluntary stability. The prosthetist then has the option of emphasizing either stability at heel strike or ease of knee flexion by appropriate placement of the knee mechanism within the prosthesis."

Judge and Fisher, [26], have introduced yet another concept into the field of prosthetic knee design. Rather than requiring a fully extended stable knee throughout stance, they have performed experimental work with a knee, (termed the "bouncy knee"-see Figure 2.12), which actually allows limited knee flexion throughout stance, as occurs in normal gait. A leaf spring placed posterior to the knee axis resists knee flexion, and can be adjusted to permit varying degrees of flexion. Limited testing has shown the knee can improve amputee gait and prosthetic control, especially when walking down slopes. The possibility of incorporating such a device into commercial knee units is now being investigated.

2.2.2 Swing Phase Control

Fernie and Ruder, [16], summarize the fundamental problem associated with prostheses which do not include a swing phase control by stating, "a free-swinging prosthetic shank behaves as a pendulum with a fixed frequency of oscillation that depends upon the effective length of the shank. This is disturbing to the amputee who sometimes wishes

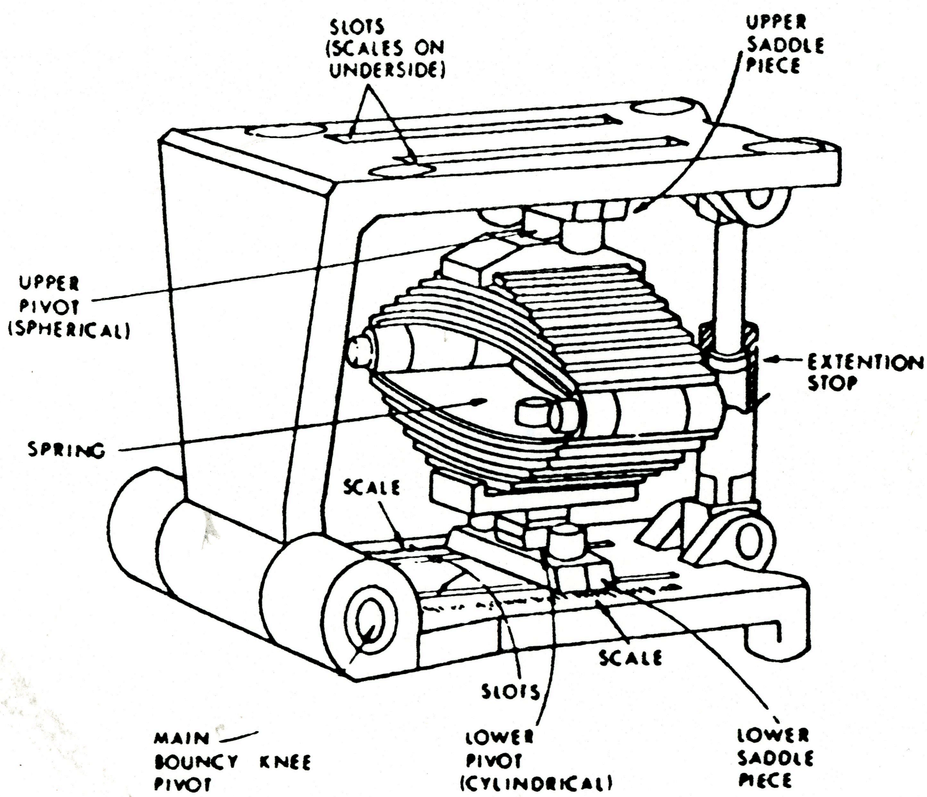


Figure 2.12 : "Bouncy knee" unit [26]

to walk faster and is held back by this physical restriction." If an amputee is to be able to walk at a variety of speeds with a reasonably naturally appearing gait, a mechanical substitute for the lost action of the quadriceps and hamstrings about the knee joint during swing must be provided.

During the swing phase, the quadriceps act in tension, to slow and control the amount of knee flexion in early swing, and to aid and accelerate knee extension in late swing. The hamstrings act in late swing to resist and control extension of the knee and prevent an abrupt impact of the knee into full extension. This is the knee moment pattern that must be generated by the prosthetic knee in order to cause the shank/foot to swing through space with a motion which approximates that of a normal person.

Design criteria for swing phase control mechanisms have been based on biomechanical data collected in laboratory gait studies performed on normal runners. Figure 2.13 illustrates the resistance patterns around the knee during an average swing phase, consisting of the three major components previously mentioned: resistance of the quadriceps to knee flexion, (limiting it to a maximum of approximately sixty-five degrees) [42], initial forward acceleration of the shank by the quadriceps and the final action of the hamstrings to decrease the rate of extension. In normal level walking, the knee begins to flex before the load is

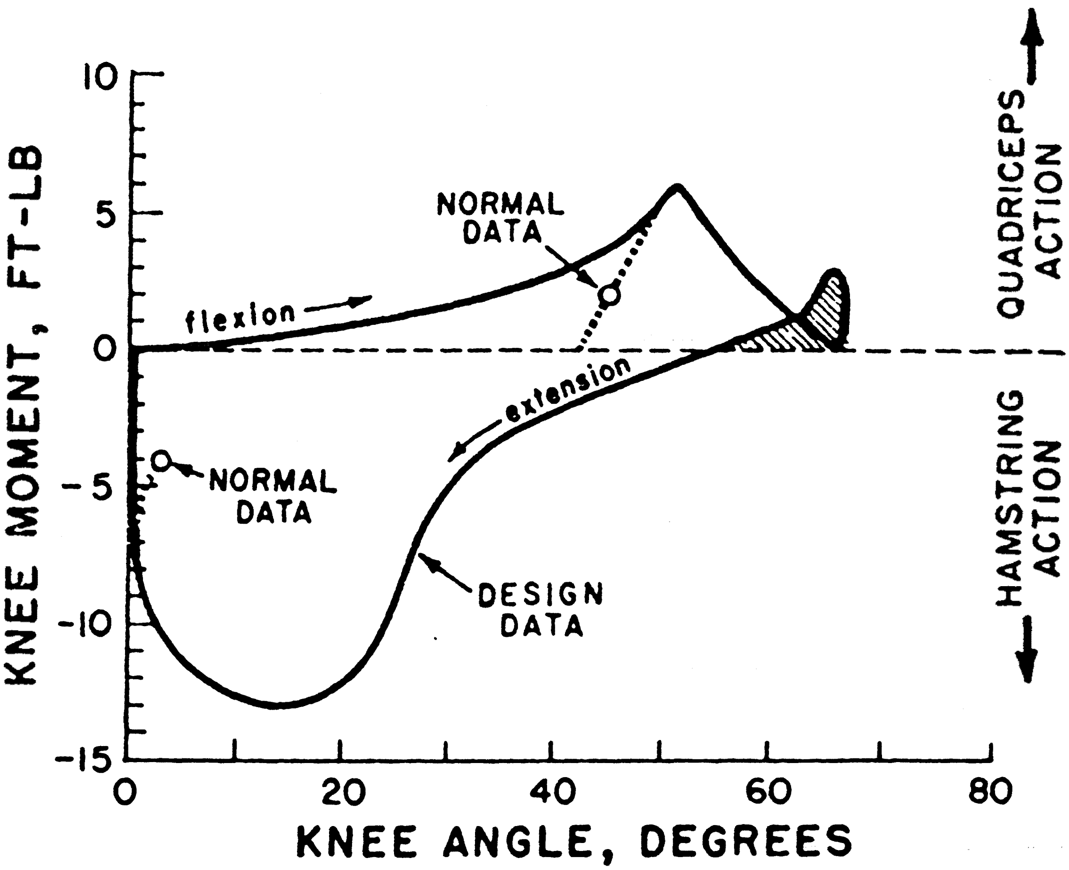


Figure 2.13 : Prosthetic knee moment versus knee angle during swing phase [42]

transferred to the opposite foot, and has flexed to approximately forty-five degrees prior to toe-off. In a standard above-knee prosthesis, knee flexion is not possible prior to toe-off. Thus Radcliffe, [42], explains that "in order to allow for this the normal data curve" is modified "as shown in the heavy line labelled 'design data'. The design data curve serves as an idealized design objective and as a basis for comparison of various swing control devices" (see Figure 2.13).

Various devices have been devised to attempt to accurately simulate this action.

The resistance characteristics of a constant friction device with an elastic extension bias is shown in Figure 2.14. When properly adjusted, the amputee is able to walk reasonably well at the speed for which it is adjusted, but the device lacks automatic adjustment for changes in walking speed [42].

Hydraulic knee units with non-linear resistance patterns, (Figure 2.15), have been used with success. A properly adjusted hydraulic swing control will allow an amputee to walk over a wide range of speeds with a gait pattern closely approximating the normal [42].

Pneumatic devices, (Figure 2.16), have been devised which rely on air compression characteristics and a controlled leak rate. These devices also approximate the desired characteristics at a wide range of walking speeds

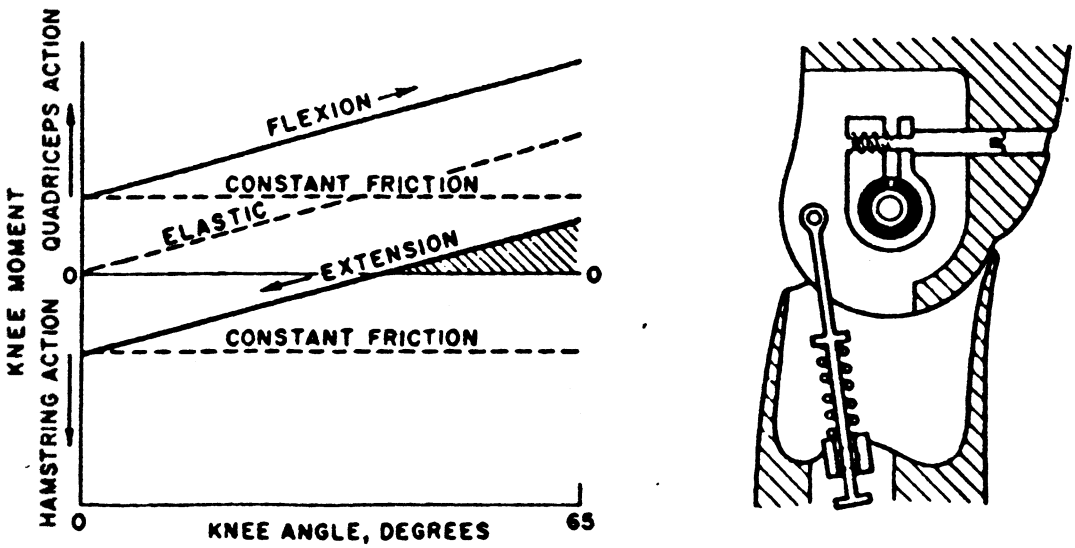


Figure 2.14 : Mechanical swing phase control with linear extension bias [42]

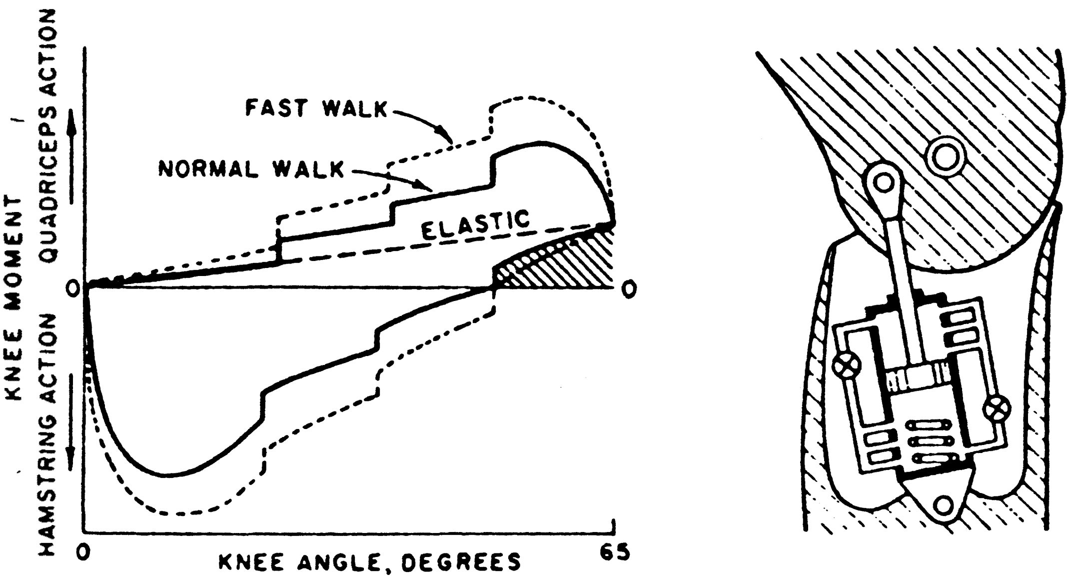


Figure 2.15 : Hydraulic swing phase control with linear extension bias [42]

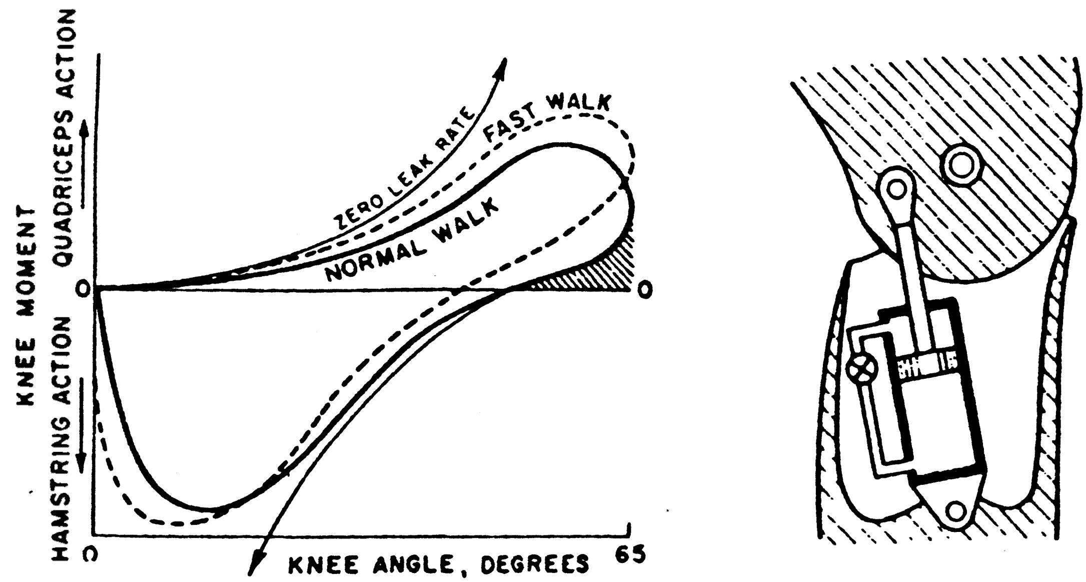


Figure 2.16 : Pneumatic swing phase control

[42].

2.3 Foot and/or Ankle Assemblies

As stated by Goh et al., [21], "a wide variety of prosthetic feet and ankle mechanisms have been designed to date, some incorporating ingenious mechanisms capable of imitating functions and movements of the normal foot and ankle complex. However, the complexity of most of these designs, their excessive maintenance requirements and unacceptably high mass have prevented their wide use. The two most common prosthetic feet used nowadays are the uniaxial (single-axis) type, and the solid ankle cushion heel (SACH) foot".

The uniaxial foot, (Figure 2.17), consists of a carved wooden keel with a single, articulating ankle joint. Rotation of the ankle occurs only in the plane of dorsi-plantar flexion; no medio-lateral motion or rotation about the vertical axis is allowed. Two rubber bumpers, (one in front and the other to the rear of the axis), provide the restraining and restoring moments, (as well as physical restraint), about the ankle for plantar and dorsi flexion movements, simulating the normal foot and ankle function during walking.

The SACH foot, on the other hand, contains no moving parts (see Figure 2.18). Instead, the heel consists of a

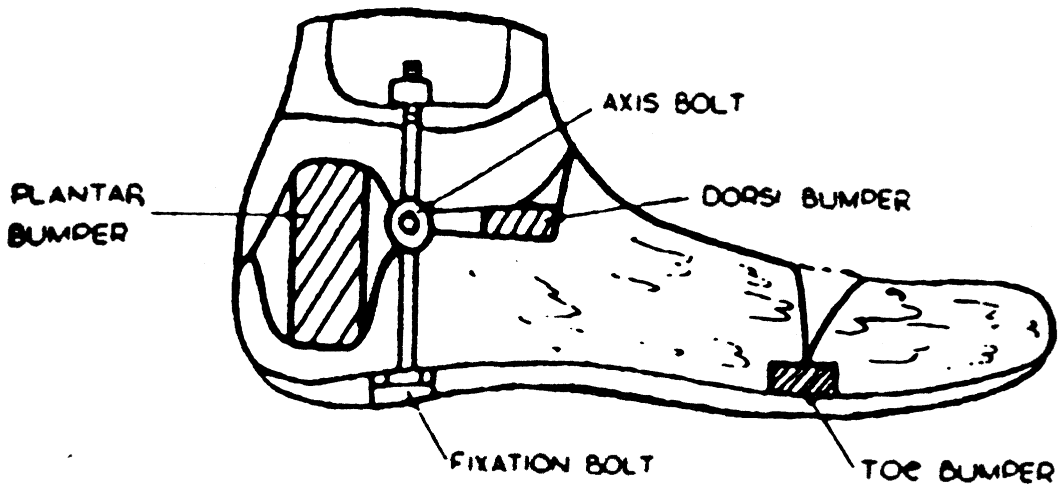


Figure 2.17 : Conventional single-axis foot [39]

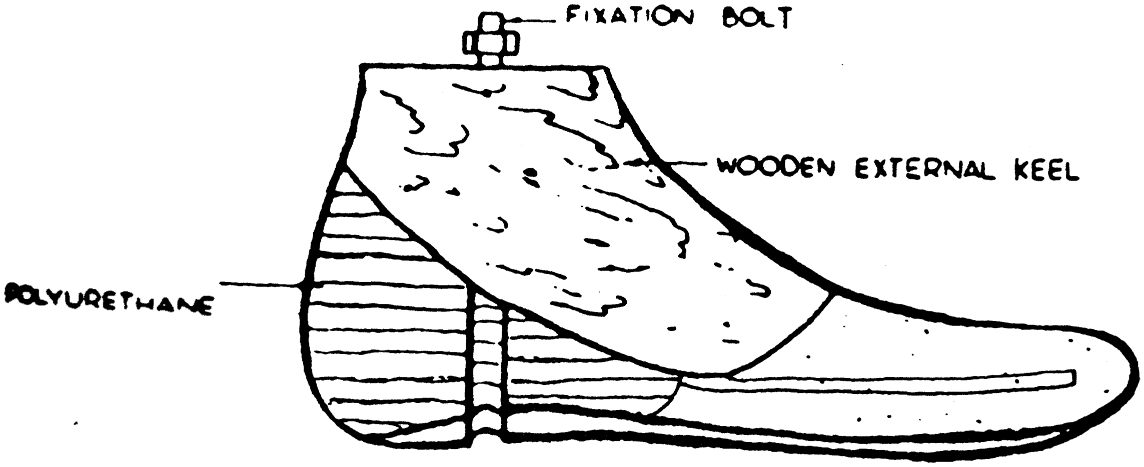


Figure 2.18 : SACH foot [39]

wedge of molded polyurethane in conjunction with an internal wooden heel, "shaped at the ball of the foot to provide a smooth, rolling action" [21]. The heel wedge compresses as it absorbs the impact at heel strike and simulates plantar flexion of the ankle, although no true rotation of the ankle joint occurs. Also, as in the single-axis foot, there is no eversion or inversion of the ankle.

In neither case does the prosthetic toe of the prosthetic foot function in the same way as the metatarsalphalangeal joint (MTP) of the normal foot. James and Stein explain that, [25], "instead of an actual joint rotation, the material at the fore of the foot deforms under the weight of the amputee and simulates the flexion of the MTP joint".

A recent study on the comparison of SACH and single-axis feet concluded that the interchanging of the two feet in a prosthesis does not effect the gait pattern of the amputee [21].

An alternate foot design which is enjoying increased usage is the Gressinger multi-axial foot. It incorporates rubber bumpers, similar to the uniaxial foot, however it permits limited rotation in the medio-lateral direction in addition to the dorsi-plantar plane. This foot can improve amputee gait over rough or uneven ground, and is usually prescribed to younger, more active amputees.

Other recent foot designs include the Stationary

Attachment Flexible Endoskeleton (SAFE) foot, which "changes its flexibility to perform different functions during the time course of the stance phase" [25]. The Copes bionic ankle employs a mechanical bearing which allows a limited amount of multi-axial ankle movement with a flexible heel section [25].

A tri-axial hydraulic ankle, developed by Mauch, has been introduced which incorporates damping into the ankle function and provides limited rotation around the vertical axis as well as the medio-lateral and anterior-posterior [36].

Two recent innovations in the area of prosthetic foot design, the Seattle foot and the Flex-Foot, are the first prosthetic feet which have been developed for the more active amputee, including running. Both designs are based on the premise of incorporating a "spring" into the foot to absorb deflectional energies generated during roll-over and then releasing that energy at toe-off, thus simulating the forward propulsion which is generated at the ankle joint in the normal runner (Figure 2.19).

The Seattle foot design utilizes a Delrin keel, (Figure 2.20), which acts as leaf spring within the foot. A Kevlar pad is added in the toe section in an attempt to reduce toe breakage which had occurred in earlier prototypes. The Seattle foot was also the first prosthetic foot to date which was made from life cast molds, making the shape of

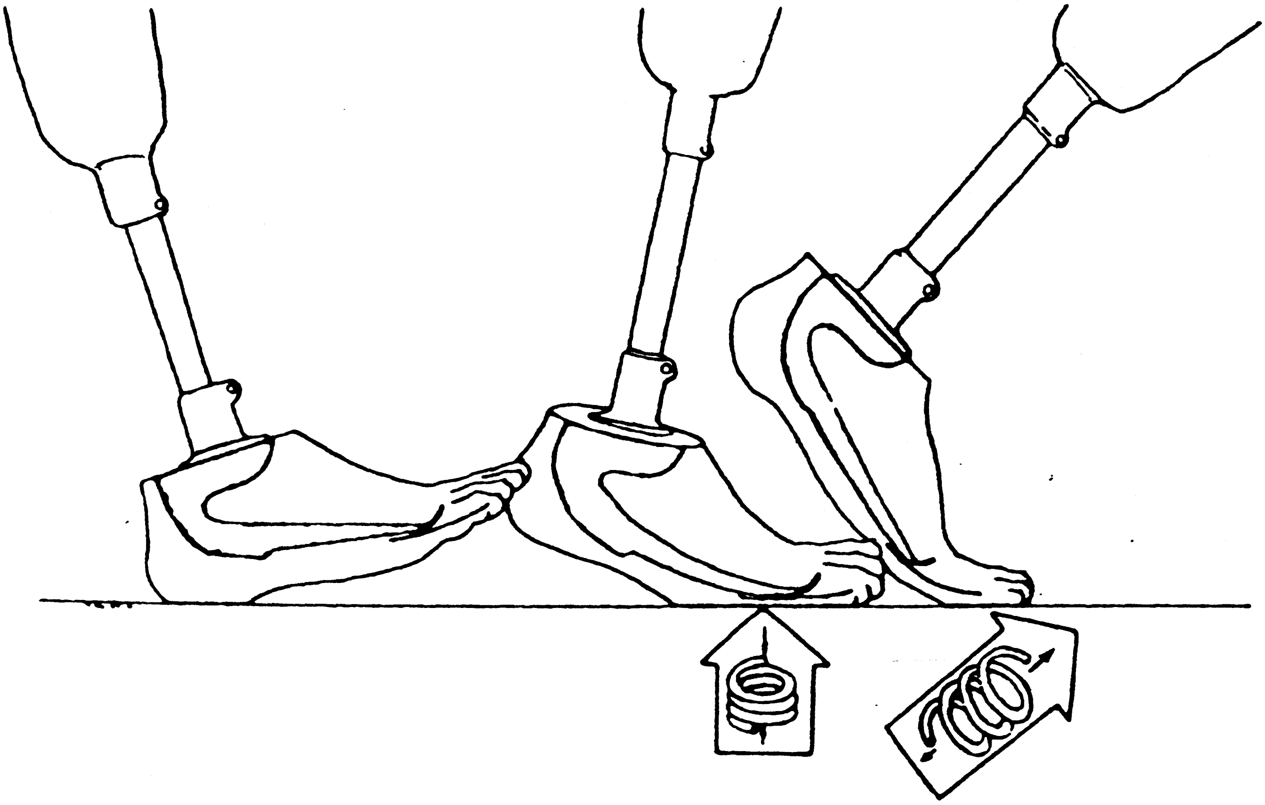


Figure 2.19 : Functioning principle of Seattle Foot [4]

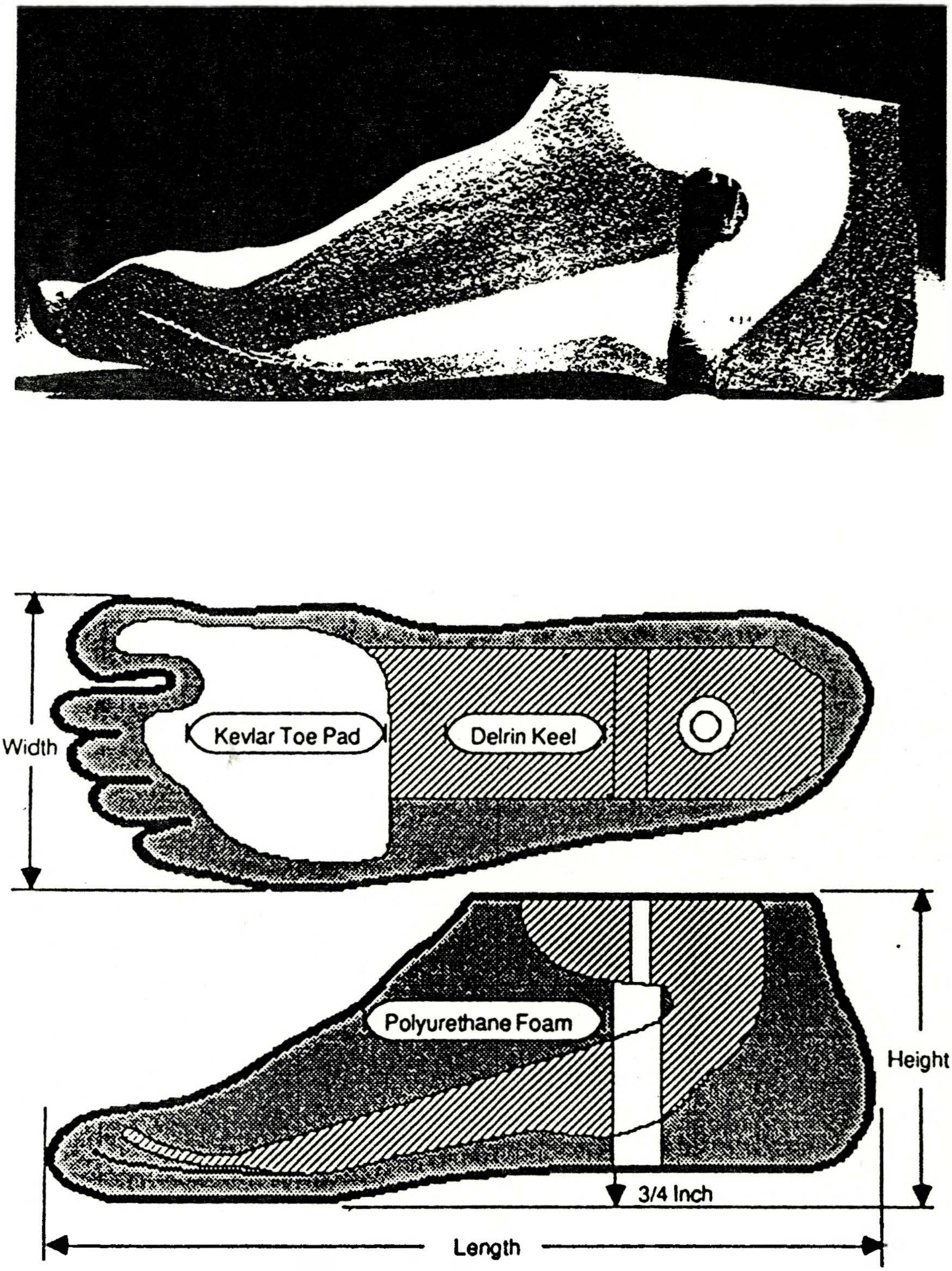


Figure 2.20 : Construction of the Seattle Foot [49]

the prosthetic foot anatomically correct [49].

Extensive testing has indicated wide acceptance of the foot by amputees, due to enhanced performance, greater endurance, and, in most cases, increased comfort [4].

The Flex-Foot is functionally similar to the Seattle foot, in that it incorporates carbon-graphite composite struts that act as leaf springs (Figure 2.21). The Flex-Foot, however, also features a leaf spring in the heel to aid in the absorption of impact energies. Similar to the Seattle foot, the Flex-Foot is popular both with young active amputees involved in strenuous athletic activities, and with older, less active amputees who wish to walk with less fatigue [17].

The Flex-Foot, (and to some degree the Seattle foot), has been designed for the below-knee amputee runner. Since the below-knee amputee has his knee joint and accompanying musculature intact, the ability of these prosthetic feet to simulate the "push-off" normally provided by the intact ankle effectively restores the minimum biomechanical components necessary for the amputee to run. The biomechanical requirements of the above-knee amputee runner are more complicated, however, but these feet show promise for future incorporation into a total above-knee running prosthesis.

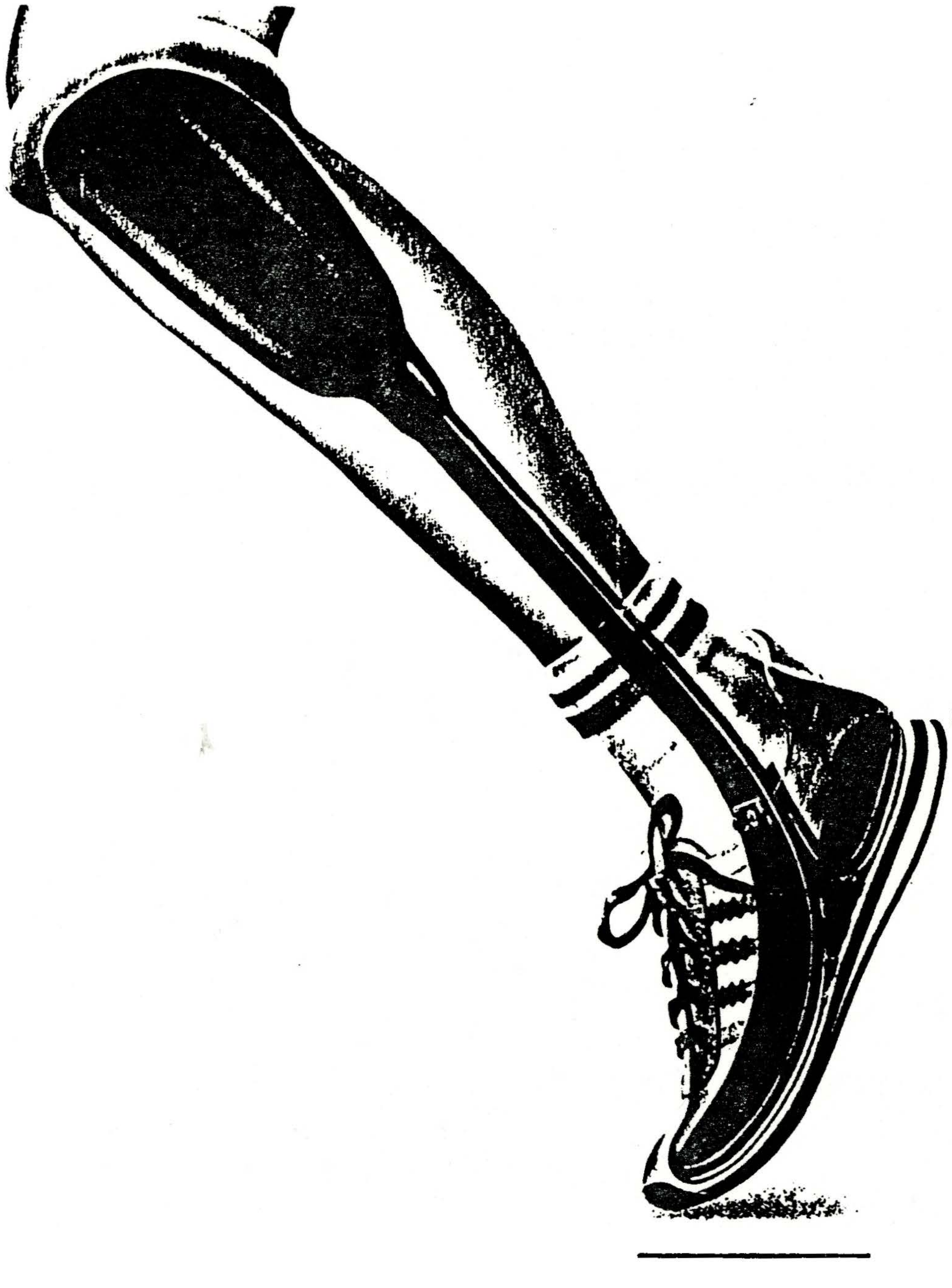


Figure 2.21 : The Flex-Foot [17]

2.4 Above-Knee Running Prostheses

As previously illustrated, improvements in areas such socket and knee design hold promise for use in running prostheses. The problem arises, however, that very little progress has been made in the development of a total running prosthesis for above-knee amputees.

Advances in socket design such as the ISNY or CAT-CAM sockets indicate that amputees should be able to endure the rigors of running while maintaining greater stump comfort than ever before. Sophisticated swing control mechanisms incorporated into recently designed prosthetic knees should accommodate the accelerated stance and swing patterns of running. Finally, prosthetic feet such as the Seattle foot are attempting to imitate the forward propulsion normally generated at the foot at toe-off. However, such technologies must be incorporated into radically new designs if a functional above-knee prosthesis is to evolve.

There have been several prostheses developed to date for the purpose of above-knee amputee running. They include the Terry Fox Running Prosthesis, developed at Chedoke-McMaster Hospitals in Hamilton, Ontario, Canada with funding from the War Amputations of Canada, and the Physiological Jogging Prosthesis, [6], developed at McMaster University, also in Hamilton.

A major obstacle in the use of conventional prostheses for running is the inadequate swing phase action of the leg. There is insufficient time in the running stride, (from toe-off to subsequent heel strike of the prosthesis), for the prosthetic shank to obtain sufficient ground clearance for swing through, yet still reach full extension to result in a stable knee at heel-strike. The conventional prosthesis used by Terry Fox on his Marathon of Hope was fitted with an extension assist, (an elastic strap passing in front of the knee, see Figure 2.22), that was to minimize the knee flexion during swing through and hasten the knee into full extension prior to heel-strike. The extension assist did not perform adequately and thus Terry was forced to use the "hop-hop-step" running gait.

One proposed solution to this problem has been the idea of a "telescoping" prosthesis. This idea had been mentioned as early as 1967 by Kenedi [29]. A group at Chedoke Hospital has also done preliminary work on a telescoping prosthesis [52]. In the telescoping leg design, the prosthesis has no knee joint. Instead, the leg shortens itself directly following toe-off and is locked in this position during swing through providing adequate ground clearance. At the end of swing, the leg fully extends once again in preparation for heel strike. Neither the practicality or the functionality of such a design has been proven in field studies.

Terry Fox also attempted to design a running

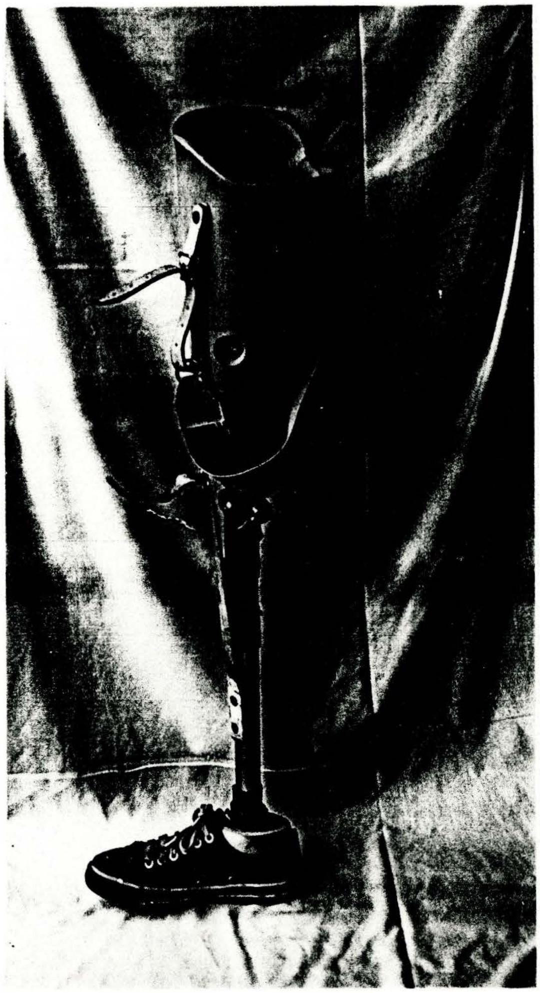


Figure 2.22 : Conventional prosthesis used by Terry Fox

prosthesis in which the telescoping aspect served a very different purpose. He and a British Columbian garage mechanic produced a crude prosthesis which has since been referred to as the "pogo" leg (Figure 2.23). Once again the prosthesis did not include a knee joint, but consisted solely of a socket, a spring assisted telescoping shank and a block acting as a foot. The design concept was that the telescoping shank would act as a "shock absorber" and absorb the impact energies at heel strike. Terry Fox was dissatisfied with the design, stating that he felt like he was "running in a hole". This was probably indicative of the use of a spring of insufficient stiffness in the shank. In addition, had the design been pursued, it is quite likely that the leg would not have allowed proper ground clearance during swing.

Based on Terry Fox's original concept of a prosthesis capable of significant impact absorption, design work began at Chedoke Hospital, sponsored by the War Amputations of Canada. The result of that work is the Terry Fox Jogging Prosthesis (see Figure 2.24).

The focal point of the prosthesis is the replacement of the standard aluminum prosthetic shank with a telescoping tube and spring in tandem. This new shank mechanism was produced in such a way as to be compatible with other modular components. The unit was combined with a standard, open-ended suction socket, (similar to that with

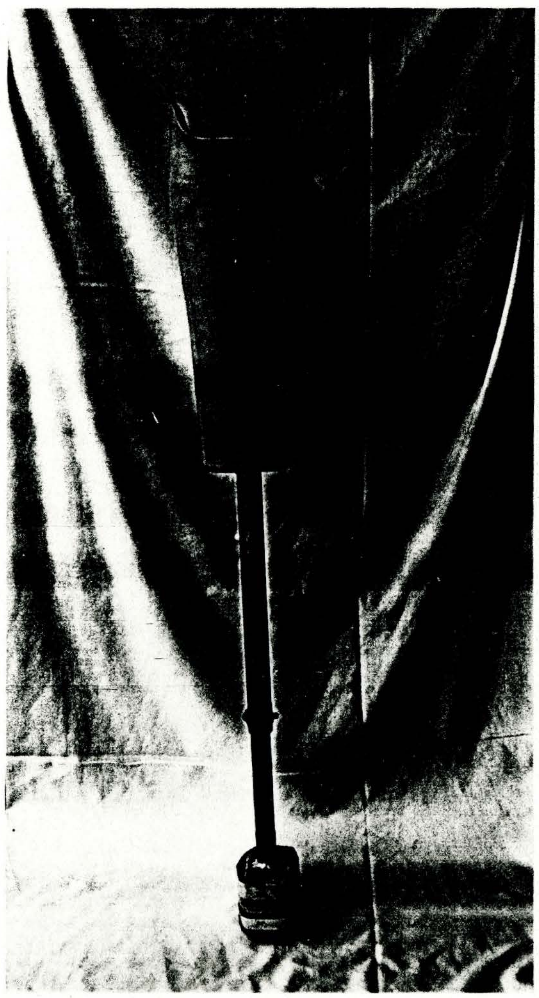


Figure 2.23 : Terry Fox's "pogo" leg

Figure 2.24 : Terry Fox Jogging Prosthesis

which the test subject was familiar), a Gressinger multi-axial foot, and a Teh-lin pneumatic, four-bar polycentric knee with swing phase adjustability of flexion and extension.

The leg was to function such that the shin tube assembly would absorb the impact forces upon heel strike, and then release the impact forces later in stance to propel the amputee forward. The swing phase control was adequate to allow a reasonably natural swing through of the prosthesis.

A biomechanical evaluation of the prosthesis' function was undertaken at the gait lab at the University of Waterloo. The conclusions were that although ground reaction forces were excessive as compared to those of normal runners of similar weight, and the prosthesis' spring collapsed to its solid height under the impact loads, the prosthesis did provide forward propulsion to the amputee and allow him to achieve a normal stride consisting of alternate periods of single support and non-support [7].

In contrast, the Physiological Jogging Prosthesis, [6], (see Figure 2.25), attempts to allow the above-knee amputee to achieve a one-to-one running pattern by trying to simulate as close as possible the kinematics and dynamics of the lower limbs of a non-amputee jogger. The simulation is based on the analyses of power, moment-of-force and ankle patterns at the hip, knee and ankle joints [6].

The design incorporates a spring posterior to a

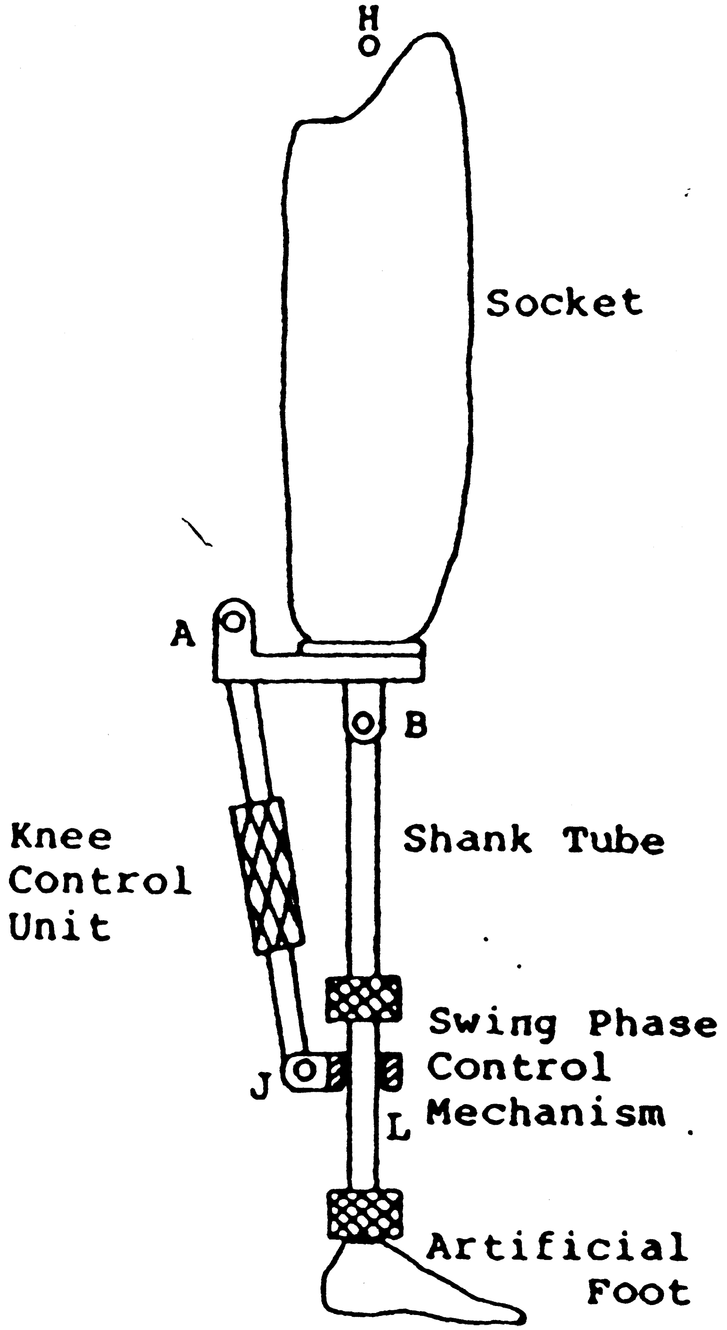


Figure 2.25 : Physiological Jogging Prosthesis

single axis knee. The spring absorbs the impact forces at heel strike in conjunction with limited knee flexion, similar to the action that occurs in a normal jogger. The spring mechanism is combined with a latching system which is designed to allow the shank to swing freely during non-support periods, yet remained locked during stance. Insufficient testing and development has occurred to date to prove the functionality of the design.

In summary, although technical advances have been made in almost all areas of prosthesis design, there has been insufficient progress made in the area of running prostheses for above-knee amputees. It is the purpose of this work to attempt to develop a prosthesis to allow above-knee amputees to run efficiently, in comfort, and with a naturally appearing running gait.

CHAPTER 3

RUNNING

Running is often regarded as a natural extension of walking, and in the broadest sense that is true. When power requirements for terrestrial locomotion, (calculated from oxygen consumption), is plotted against speed for a man, (Figure 3.1), a break occurs, indicating a change of gait. Extrapolation indicates that walking would require more power than running at speeds higher than 2.3 m/s, (5.2 mi/hr), while running would require more power than walking at lower speeds. Humans automatically choose the gait which requires less power at the speed in question [3].

In the most clinical sense, human gait is the movement of the body over the ground by being alternately supported and projected by each leg and foot [8]. As stated by Adelaar, [1], "the phases of gait in walking are the key to understanding the running cycle".

The basic unit of locomotion is the stride and is commonly measured from the initial ground contact of one foot to the subsequent touchdown of the same foot (eg. ipsilateral footstrike to ipsilateral footstrike). A step

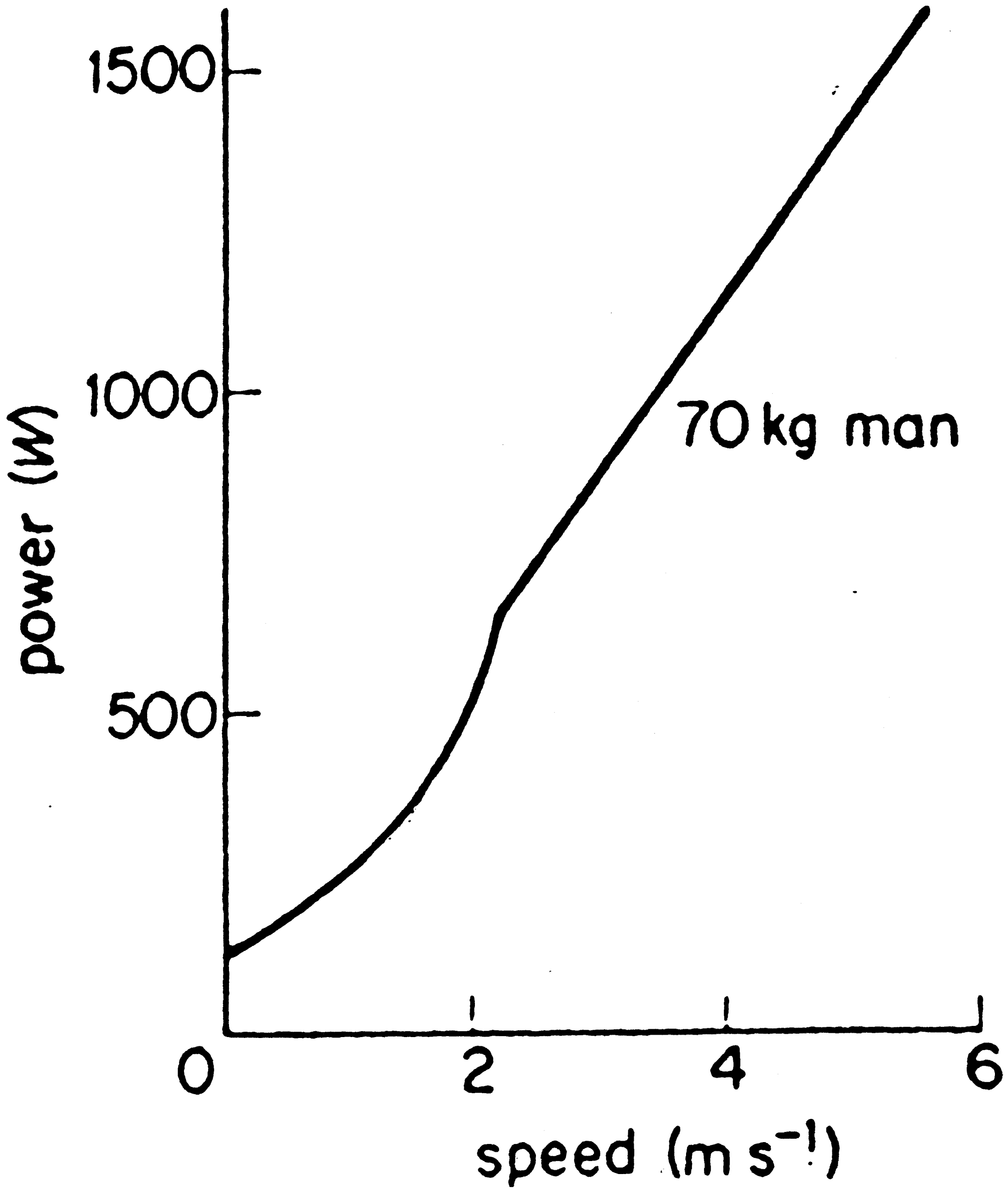


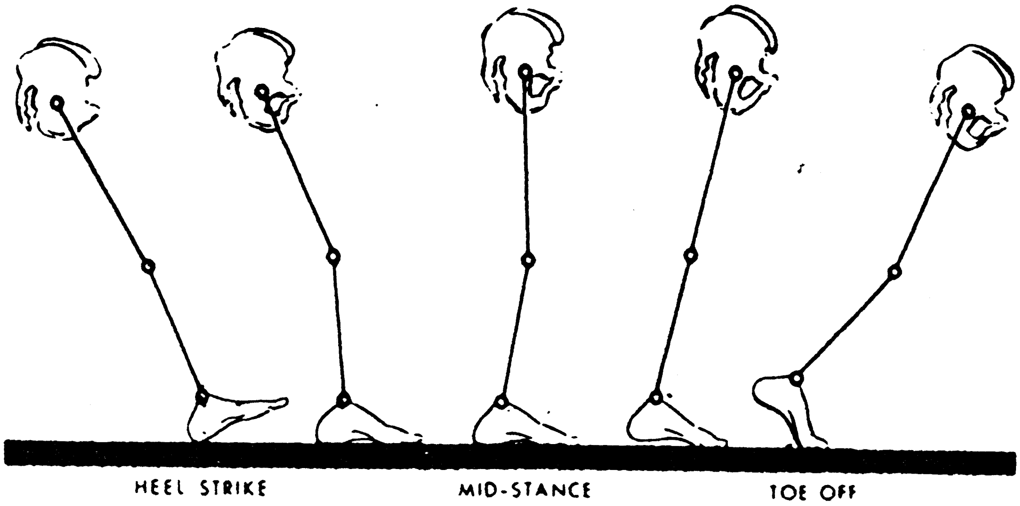
Figure 3.1 : Power requirements versus speed for human terrestrial locomotion [3]

refers to that half of the stride encompassed by two consecutive footstrikes (eg. ipsilateral footstrike to contralateral footstrike) [37]. In normal symmetric gait, a step would simply be half of a stride. It is not always valid, however, particularly with a pathological running gait such as that of Terry Fox or Steve Fonyo, to make such an assumption [53].

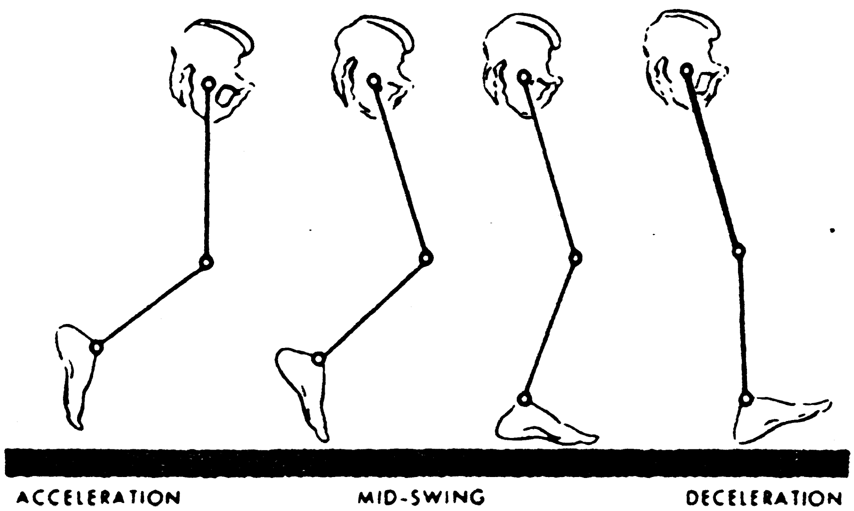
The walking stride of one limb can also be divided into two phases, the stance phase and the swing phase (Figure 3.2).

Stance phase begins at heel-strike, which is the instant the foot touches the ground. As the foot obtains a stable position on the ground, the body weight is swung directly over the supporting leg and continues to rotate over the foot. This is known as mid-stance. As the body rotates forward, it is propelled forward by powerful plantar flexing of the foot. The stance phase terminates as the foot is removed from the ground, known as toe-off [41].

Swing phase begins the instant the toe leaves the ground. Initially, the leg must accelerate forward in order to catch up to and get in front of the body in preparation for the next heel contact. When the leg has caught up to and passes directly beneath the body, it is referred to as being in mid-swing. Following mid-swing, there is a period where the leg is decelerated to correctly position the foot in preparation for heel contact and the completion of the



(a) Stance



(b) Swing

Figure 3.2 : The walking cycle [41]

walking cycle [41].

When alternating from swing to stance on each leg, there exists a period where both feet are in contact with the ground simultaneously, (pre-toe-off on one leg, post-heel-strike on the other). This is referred to as the period of double support, and its length of duration is inversely proportional to the walking speed [41].

During the walking gait cycle, at ordinary speeds, a single leg is in the stance phase for approximately sixty percent of the cycle, and in swing for approximately forty percent. The period of double support occupies about twenty-five percent of the gait time, and represents a portion of the stance phase time for each leg.

As mentioned previously, as walking speed increases, the period of double support decreases, and if speed is sufficiently increased, it is eliminated entirely (i.e. race-walking, [53]). In such a case, the overall cycle time will decrease, with the subject spending equal amounts of time in stance and swing.

At increasingly faster speeds, the subject will progress to running, which is characterized by an overall decrease in the cycle time, and, rather than periods of double support, there now exist periods of non-support, (float, flight), when both legs are in swing and the runner is airborne. The swing phase in running gait can now be further subdivided into two periods of flight separated by an interval of stance on the contralateral limb [37], (Figure 3.3). With increased running speed, the stance phase represents a smaller and smaller percentage of the overall cycle time [53].

In the case of lower limb amputees, the most generally accepted definition of running is when the period of double-support is reduced to zero [14]. Any increase in gait speed after that point will result in periods of flight.

It has long been assumed that it was not possible for lower limb amputees to run, particularly above-knee amputees. However, an increasing desire by amputees to participate in a wider range of activities involving running has resulted in the development of a number of "running" techniques.

There have been a number of studies concerning lower-limb amputee running patterns, (the majority by the Department of Kinesiology at the University of Washington, [13], [38]) but upon closer examination it becomes evident that it is only below-knee amputees that have been able to achieve a running gait in its truest sense (alternating periods of single support, followed by flight and swing phases).

The two most common running techniques for above-knee amputees are the unsymmetrical gait, as exemplified by Terry Fox and Steve Fonyo, and the crutch running technique [27]. As mentioned previously, the unsymmetrical running gait can be described as a "hop-hop-step", where one step is taken on

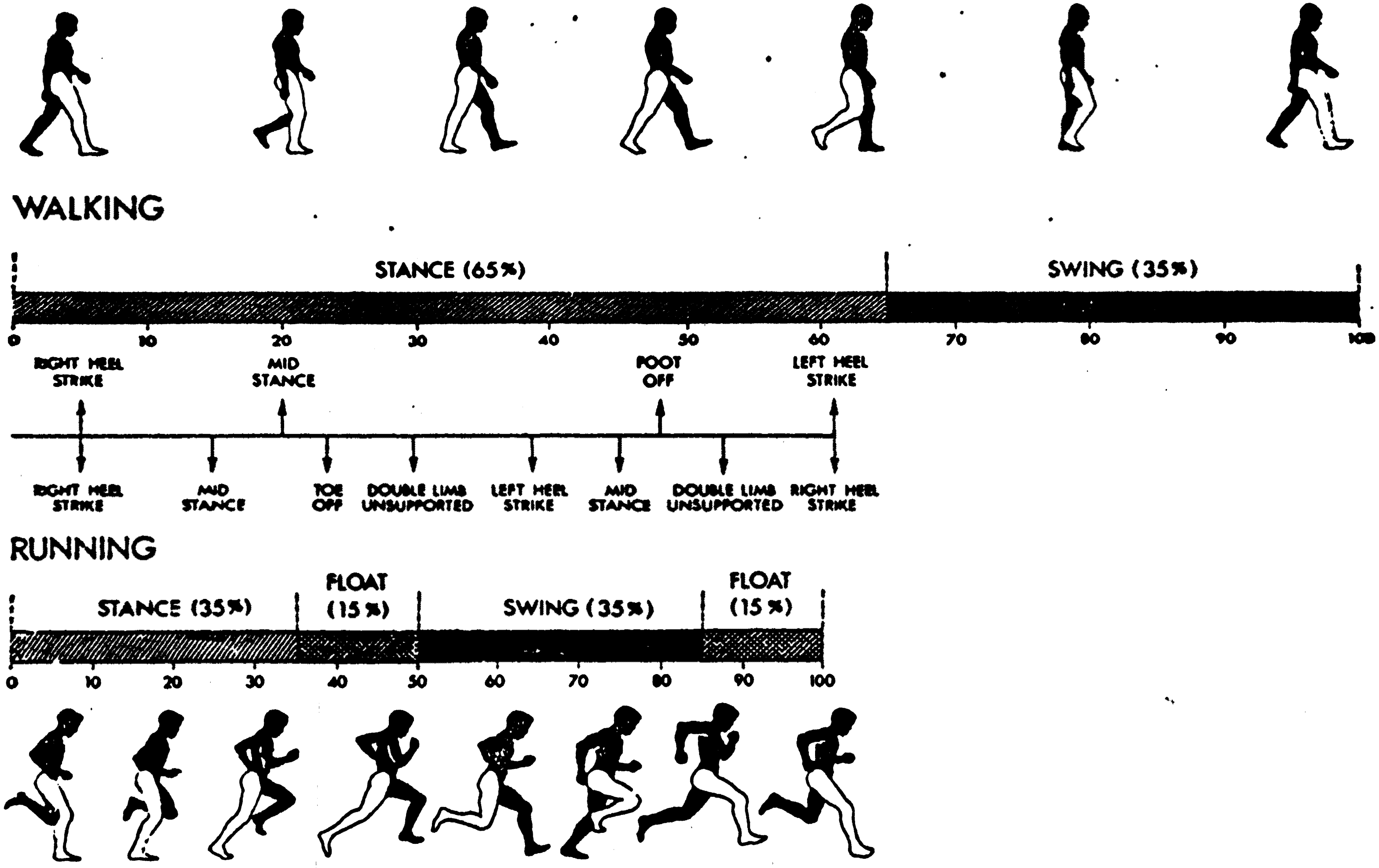


Figure 3.3 : Walking versus running cycle [1]

the prosthesis, followed by two "hops" on the sound limb. In crutch running, the amputee uses two crutches, but no prosthesis, The running technique is similar to the unsymmetrical gait; one step is taken on the crutches, followed by two steps on the sound limb.

With increased participation in running among above-knee amputees, the inherent inadequacies of conventional prostheses have been magnified, as well as introducing other difficulties.

One inherent difficulty which is emphasized during running is the lack of proprioception, or awareness of the orientation of the prosthesis in space by the amputee. The amputee often begins walking training using visual confirmation of prosthetic placement, until there eventually develops a "feel" for the device, and dependence on visual feedback decreases. The increased velocities of all aspects of the running gait, however, require complete retraining of the amputee in terms of prosthesis placement.

The anatomical knee's natural swing phase control and impact absorption abilities upon heel strike are a crucial component of a non-amputee's running gait. The function of the intact knee and associated musculature in running is even more difficult to duplicate than in the walking cycle. Similarly, the loss of the ankle musculature and its ability to provide the forward propulsion at toe-off is also increasingly emphasized as the above-knee amputee attempts to run. Fortunately, designs such as the Terry Fox Jogging Prosthesis, the Physiological Jogging Prosthesis and the Seattle Foot are attempting to simulate the energy absorption and energy generation characteristics of these joints.

Running can also introduce the need for additional socket suspension, such as silesian belts. Socket drop during swing, due to increased centrifugal forces from increased swing velocities, can result in inadequate clearance during swing. Also, amputee runners can experience a rotation of the socket around the stump, such that the prosthetic is excessively inverted or everted at heel strike [27].

An above-knee amputation invariably results in a decrease in available stump musculature and associated output as compared to the intact limb. Although the musculature may be more than sufficient for walking purposes, this decreased musculature can act as a limiting factor when the amputee attempts to run. It is possible, however, if the amputee is sufficiently motivated, for the stump strength to be increased through training.

Excessive perspiration within the socket due to the increased physical effort associated with running can result in skin irritation if proper care of the stump is not maintained, including frequent changes of stump socks.

Above-knee amputee running can also put excessive physical strains on both the amputee and his prosthesis.

Ground reaction forces generated during the amputee walking cycle rarely exceed body weight, and these are the conditions for which conventional prostheses are designed. Ground reaction forces generated during amputee running, however, can exceed three times body weight. Thus, the life of the prosthesis is significantly reduced when used for running, while the possibility of in-service catastrophic failure is increased. Prosthetic feet and knees are especially prone to premature breakage under such harsh loading conditions.

Such high ground reaction forces, when transferred through a fully extended knee, can place excessive physical strains on the amputee's stump, hip, and back. Once again, prostheses which incorporate impact absorption devices, such as the Terry Fox Jogging Prosthesis and the Physiological Jogging Prosthesis, are attempting to alleviate these problems.

In unsymmetrical running gaits, such as Terry Fox's, as running speed increases, the amputee begins to spend less and less time on the prosthesis and increasing amounts of time on the sound limb. Thus the major proponent of propulsion during the gait cycle is the sound limb, resulting in, as stated by Burgess et al. [4], a "high-energy consuming, uncomfortable, unstable, and unsightly gait pattern". Such a gait can result in excessive amputee fatigue.

To date, the only prosthesis that has allowed an above-knee amputee to achieve a symmetrical running gait has been the Terry Fox Jogging Prosthesis.

A bio-mechanical evaluation of the prosthesis was performed at the gait labs at the University of Waterloo [7]. The subject, a twenty-one year old male uni-lateral above-knee amputee was able to achieve a proper running stride consisting of alternate periods of single support and non-support.

The peak ground reaction forces were found to be considerably higher, (almost twice as high), as those of a normal jogger. The commercial components of the prosthesis were anticipated to have a considerably decreased life under such loading. Also, the exceptionally high ground reaction forces again indicated that the amputee was subjecting himself to significant physical strains.

As evidenced by the subject's performance, the swing phase control abilities provided by the pneumatic knee were adequate to allow a one-to-one running gait to be achieved. There was, however, insufficient kinematic study to determine whether "optimum" swing characteristics were being achieved, or rather that the subject was able to accommodate the deficiencies in the swing phase control provided.

The Terry Fox Jogging Prosthesis shin assembly incorporates a spring, to absorb and return impact forces, and an air damper, the main function of which is to prevent rapid collapsing of the unit during the end of stroke. There were, however, several apparent difficulties associated with this design.

The spring was found to be of inadequate stiffness to absorb the ground reaction forces. The spring collapsed completely under the impact loads, with the remainder of the impact forces being transmitted through the fully compressed unit, and passed on to the stump and socket. Thus, not only did this result in unnecessary stresses being placed on the stump, but the transferred energy was irretrievably lost and unavailable for use for propulsion at toe-off.

The air-damper dissipates energy not only upon impact, but also during the return stroke at toe-off, once again reducing the total amount of energy available. Using a spring of sufficient stiffness, the collapse of the unit would be self-limited by the forces generated, and thus an air-damper would not be necessary.

Due to the air-damper, and the losses due to the insufficient spring stiffness, the testing revealed that only about fifty percent of the energy absorbed by the shin assembly was returned to the jogger.

The aforementioned difficulties in amputee running can lead to only one conclusion, as stated by Burgess et al., [4]; that it had become evident that "...the state-of-the-art lower limb prostheses would have to be redesigned if real progress in amputee running was to be accomplished."

CHAPTER 4

MUSCULATURE IN RUNNING

The previous chapter has described the basic kinematic elements of a "normal" jogging or running gait, as well as the current inadequacies apparent in amputee running gait. If a prosthesis is to be developed that will allow amputees to simulate a natural gait, a true understanding of the biomechanics involved in the achievement of that gait by a non-amputee must be realized.

4.1 Definitions

A number of terms concerning various body movements must be defined before a discussion of lower limb musculature can begin.

Duvall, [9], defines the universally accepted terms describing various body motions in the following ways. Flexion is a motion which decreases the angle formed at the joint, as in the closing of a hinge. Extension is the opposite of flexion, and is a motion which increases the joint angle (Figure 4.1). Specific terms refer to the

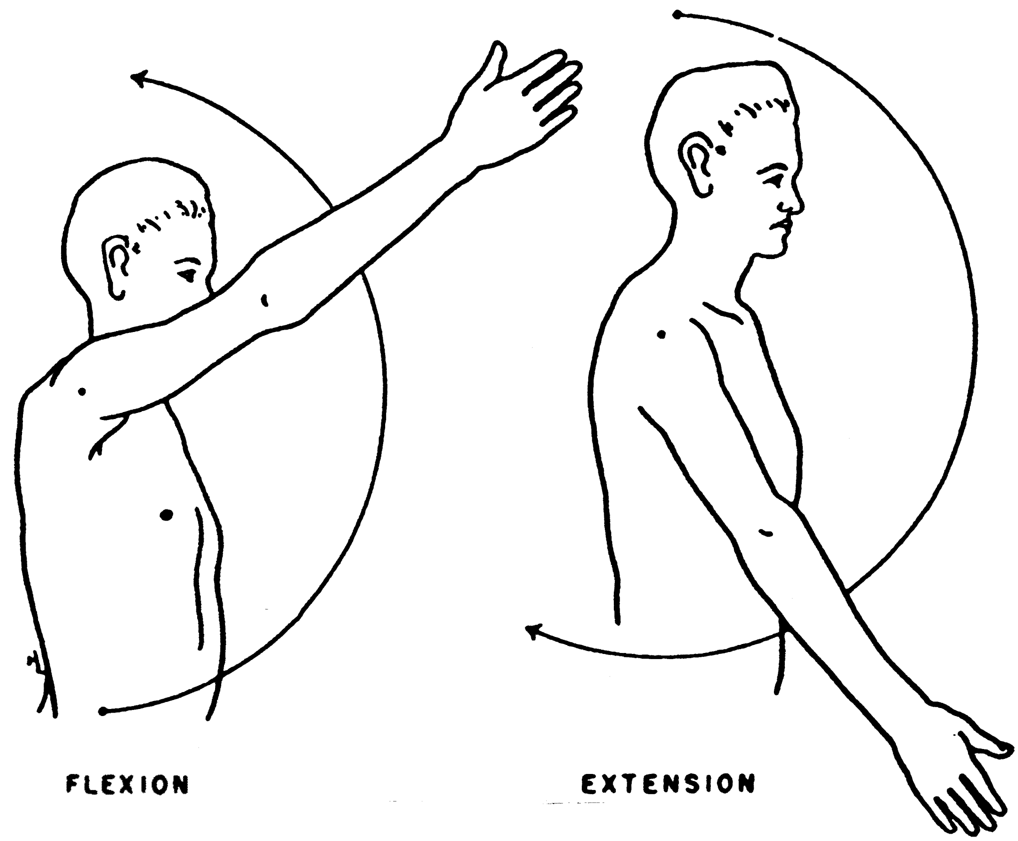


Figure 4.1 : Flexion and extension of the arm [9]

flexion and extension of the ankle joint. True ankle flexion is referred to as dorsi-flexion, whereas ankle extension is denoted as plantar flexion (Figure 4.2). Lateral rotation of the ankle, when the sole is turned inward with the weight bearing on the outer edge of the foot, is known as inversion. The opposite case, with the sole turned outward, is known as eversion (Figure 4.3). The movement of a body part away from the longitudinal axis of the body is known as abduction, while the movement of a body part toward that axis is known as adduction (Figure 4.4).

4.2 Muscular-Skeletal Biomechanics

The human skeleton consists of a number of bones articulating at joints. Muscles circumscribe these joints and upon contraction exert moments which can resist or produce motion. The muscular-skeletal system is quite simply a series of levers.

In most instances in the body, the force arm in the muscle-bone lever is much shorter than the resistance arm, and thus the force generated by the muscle to produce movement is much greater than that actually seen at the point of resistance. The short force arm, however, results in greater speed and range of motion of the resistance arm in return for the inherent mechanical disadvantage.

The actual arrangement of muscles, tendons, ligaments

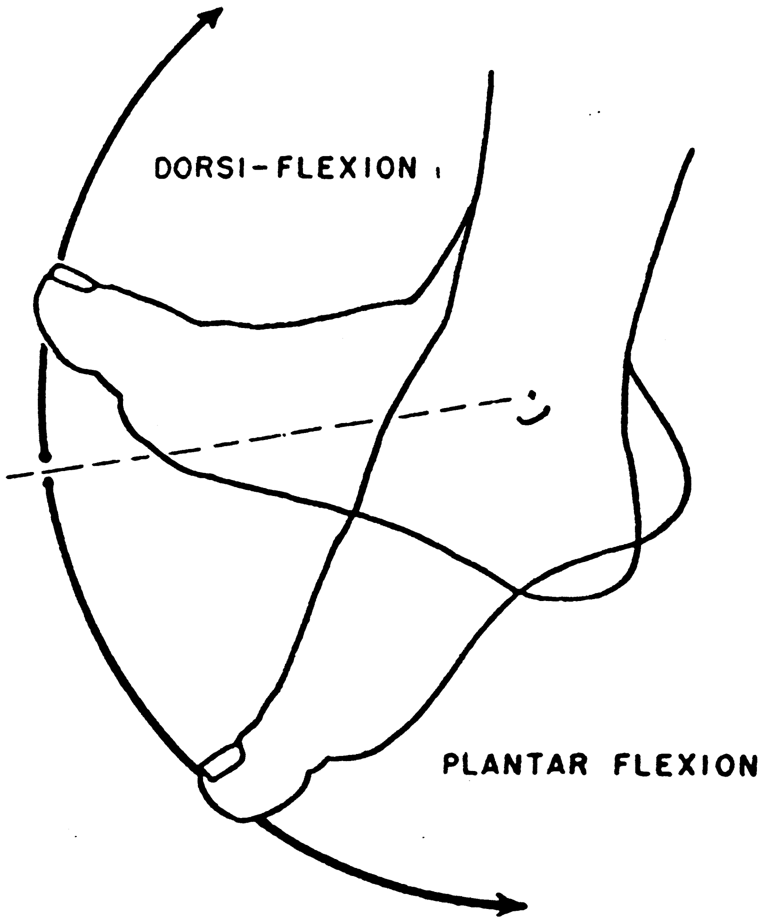


Figure 4.2 : Dorsi and plantar flexion of the foot [9]

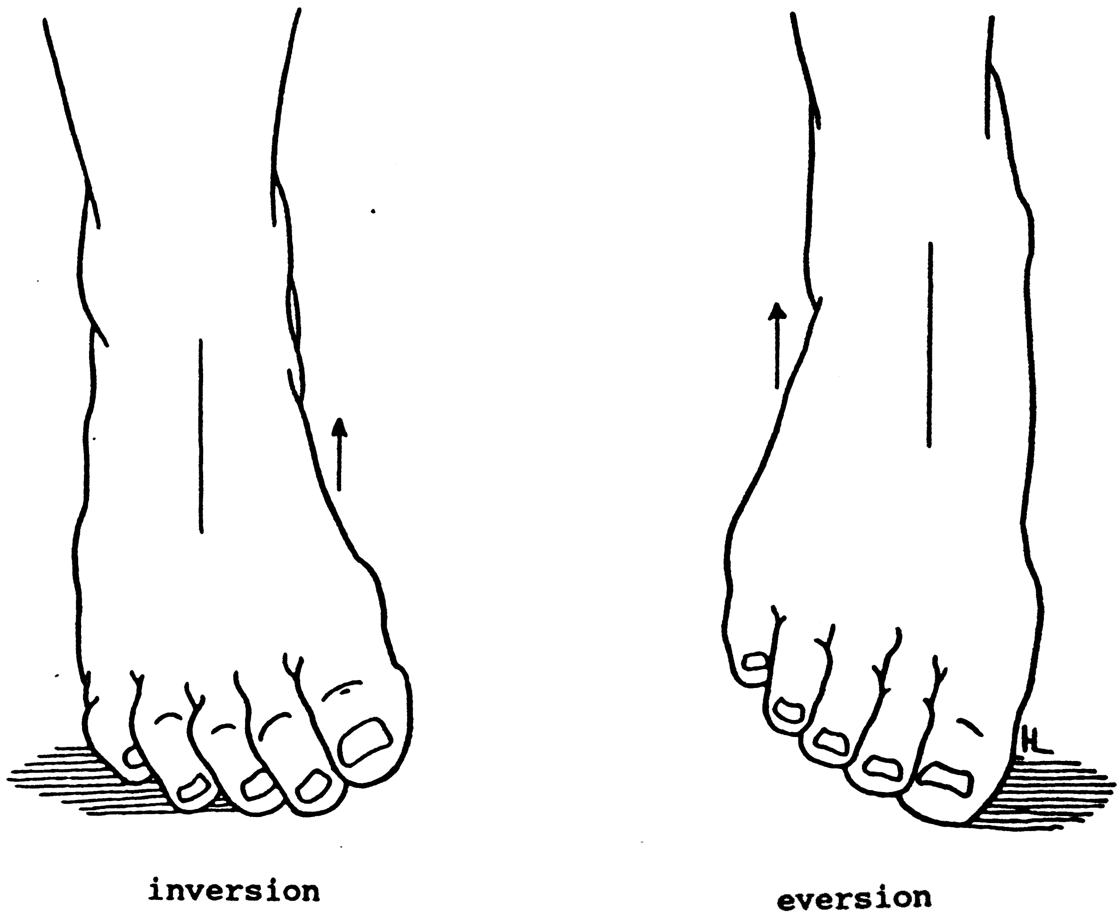


Figure 4.3 : Inversion and eversion of the foot [9]

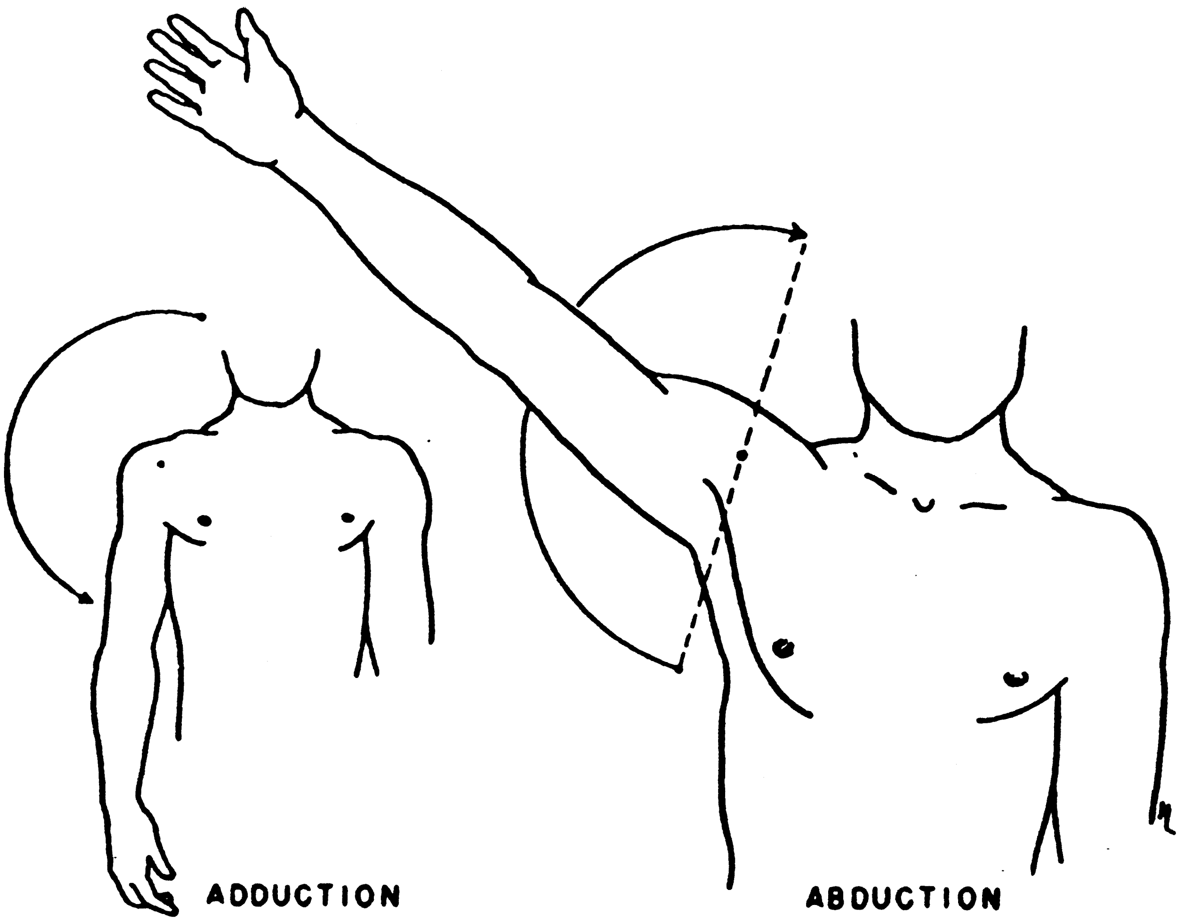


Figure 4.4 : Adduction and abduction of

the arm [9]

and bones throughout the body is quite complex, with many muscles transversing a joint at a variety of angles and locations. Muscle action about all joints, however, follows a few basic principles.

Muscles are capable of producing only a resistive, contractile force. Muscle action can be divided into two categories, isotonic and isometric contraction. Isotonic contraction is that action which is commonly associated with muscles, the generation of a force resulting in a movement of a load, (external work), due to a change in the length of the muscle. A person lifting a weight through the contraction of the bicep is an example of isotonic contraction. In isometric contraction, a force is generated, but no change in muscle length occurs, and no external work is done. All muscles which exert a force that stabilize the leg joints when a human stands erect and motionless are in isometric contraction [22].

Isotonic contractions can be further subdivided into concentric and eccentric contractions. Concentric contraction is the generation of muscle force by the shortening of the muscle, such as the shortening of the biceps in the lifting of a weight. The production of a resistive force while the muscle is lengthening is called eccentric contraction. Eccentric contraction occurs in the bicep in the subsequent lowering of the weight (Figure 4.5).

In some discussions of muscle activities, [56],

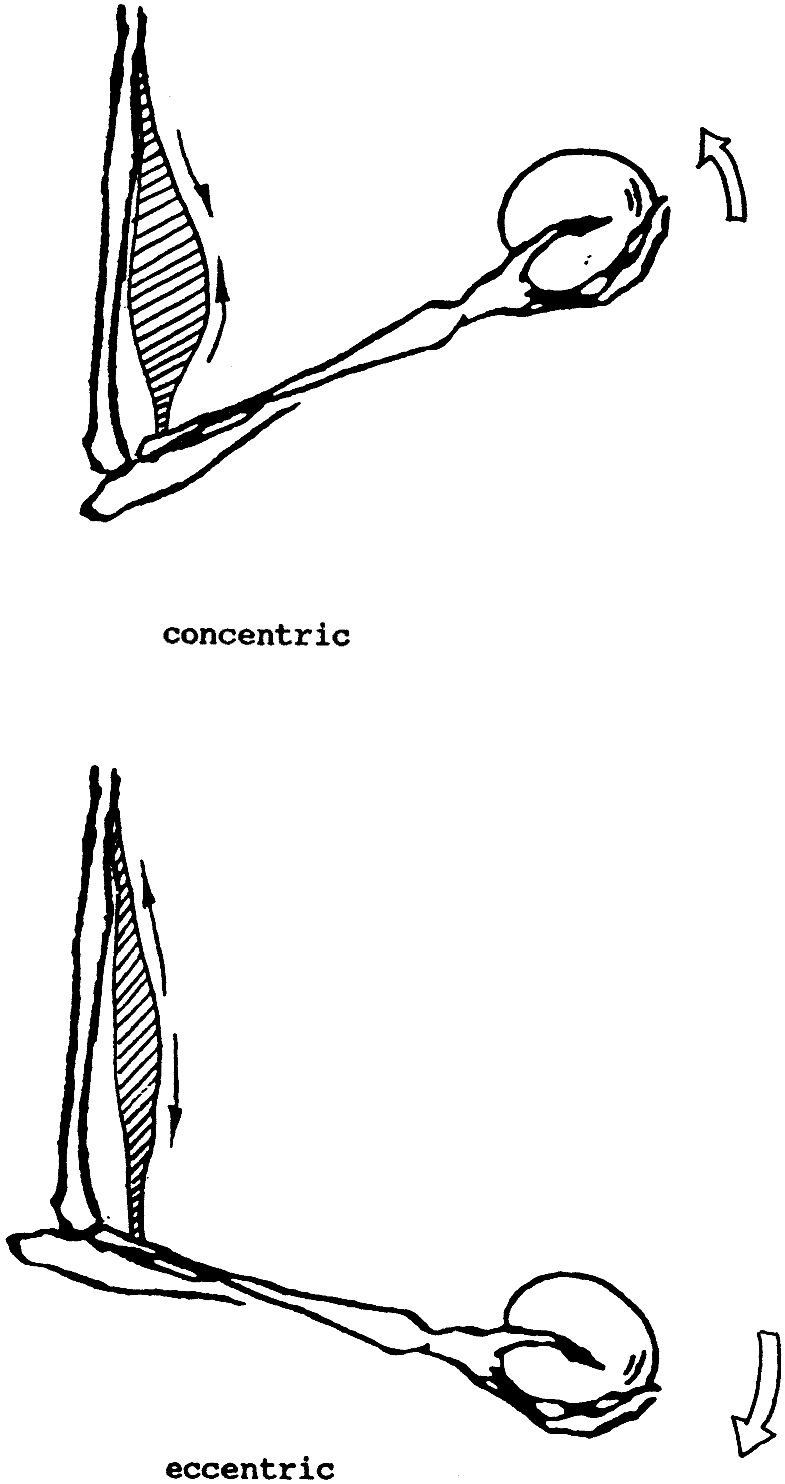


Figure 4.5 : Concentric and eccentric contraction of the bicep

including the bench-mark studies of Elftman, [10], movement associated with eccentric muscle contraction is referred to as negative muscle work, since the direction of motion is opposite to the direction of the applied force of the muscle. Motion in the direction of the applied force, (i.e. concentric contraction), is referred to as positive muscle work. This terminology shall be utilized throughout the remainder of this work.

As stated previously, skeletal muscles can only produce a contractile force; they are unable to "push", only to "pull". Therefore, in order to produce a full range of controlled motion at a given joint, the body provides a minimum of two sets of muscles. One set of muscles contracts concentrically to provide the desired motion, (the hamstrings to flex the knee, for example), while the opposing muscles contract eccentrically to resist and control the motion (the quadriceps, in this example). When the opposite motion is required, the roles are reversed. The muscles directly responsible for the motion are known as agonists, while those resisting the movement are called antagonists [9] (Figure 4.6). The preceding example is the simplest of all muscle arrangements. In joints where a wider variety of movements are possible, the muscle configurations are more complicated, however the same principles of agonist-antagonist muscle action apply. The arrangements can also become more complex since the muscles are often situated such that they are

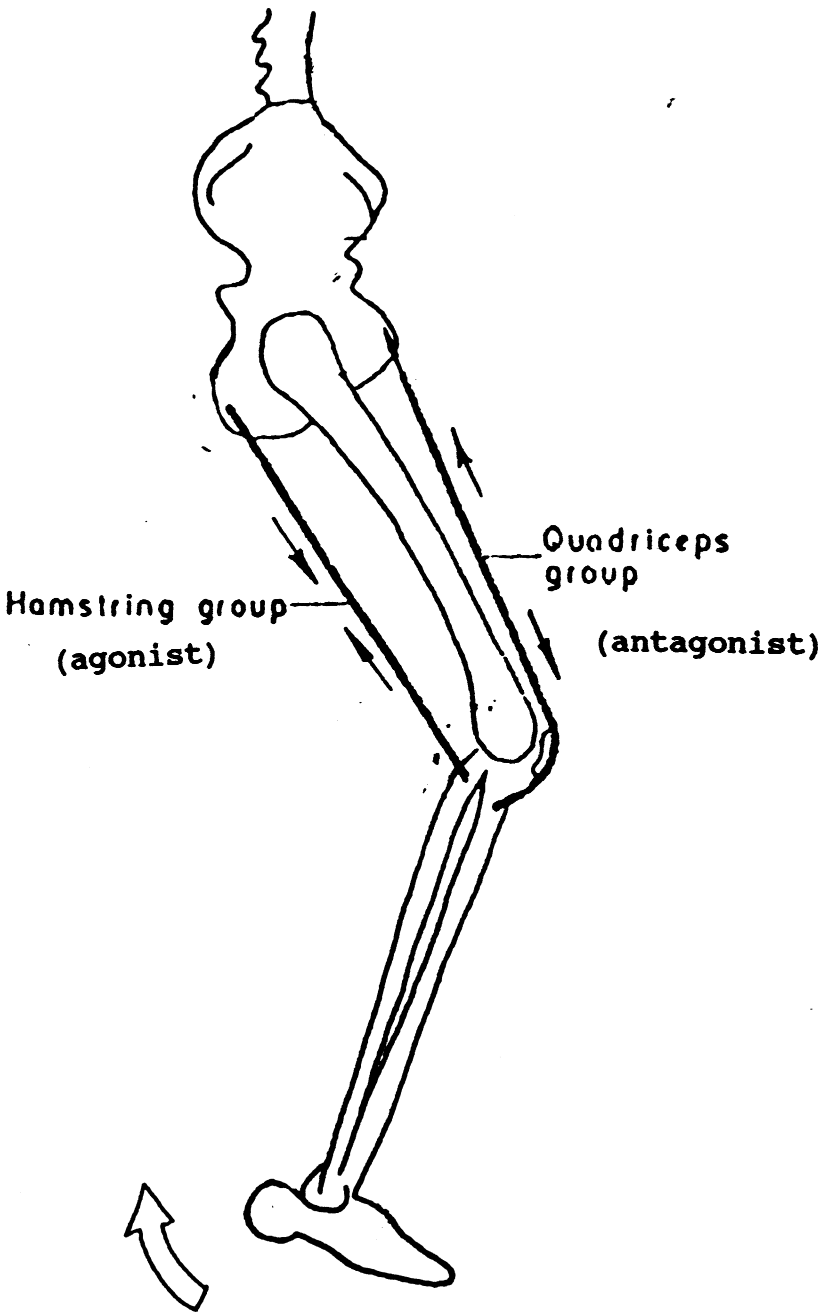


Figure 4.6 : Agonist-antagonist muscle action

responsible for more than one type of movement. A muscle can pass anterior to the ankle joint and attach to the foot in such a way that contraction can result in both dorsi-flexion and eversion of the foot, with the muscle not acting to full efficiency in either case. Also, there exist a number of multi-joint muscles, muscles which transverse more than one joint. In most cases, these muscles can be equated with uni-joint muscles since they rarely act on both joints simultaneously. Most commonly, one joint is stabilized while the muscle acts on the other joint [9]. Finally, it must be remembered that gravity is a force acting on the body and can act as either an agonist or a antagonist. Returning to the example of the bicep lifting and lowering the weight, as the bicep acts as the agonist to lift the weight, gravity acts as the antagonist to provide the resistive force. Conversely, as the weight is lowered, gravity becomes the agonist.

4.3 Lower-Limb Musculature

The basic lower-limb muscle groups can be generally categorized as flexors and extensors of each of the major joints: hip, knee and ankle. Some muscles are multi-jointed and thus are named as activators in more than one joint motion. The muscles are listed as agonists for the motions indicated, and can be assumed to be antagonists for the contrary motion.

The major hip flexors are comprised of the iliopsoas, sartorius and rectus femoris muscles. The major hip extensors are comprised of the gluteus maximus and the "hamstring" muscle group, consisting of the biceps femoris, semitendinosus and the semimembranosus (Figure 4.7).

The major knee flexors are the gastrocnemius, sartorius and the hamstring musculature. The major knee extensors are the quadriceps group, made by the rectus femoris and the vastus muscles (lateralis, intermedius and medialis) (Figure 4.8).

The major plantar flexors of the ankle are the gastrocnemius and the soleus muscles. The major dorsi-flexor is the tibialis anterior, with minor contributions from a number of other pre-tibial muscles (Figure 4.9).

For purposes of study, these muscles can be further simplified into six main muscle groups, [35], responsible for flexion and extension of the three lower-limb joints in the sagittal, (anterior-posterior), plane. These groups are illustrated in Figure 4.10.

4.4 Muscle Action in Running

As previously mentioned, the running cycle can be subdivided into two phases, stance and swing. The stance phase begins the instant the foot contacts the ground and ends with the final forward drive at toe-off. The swing

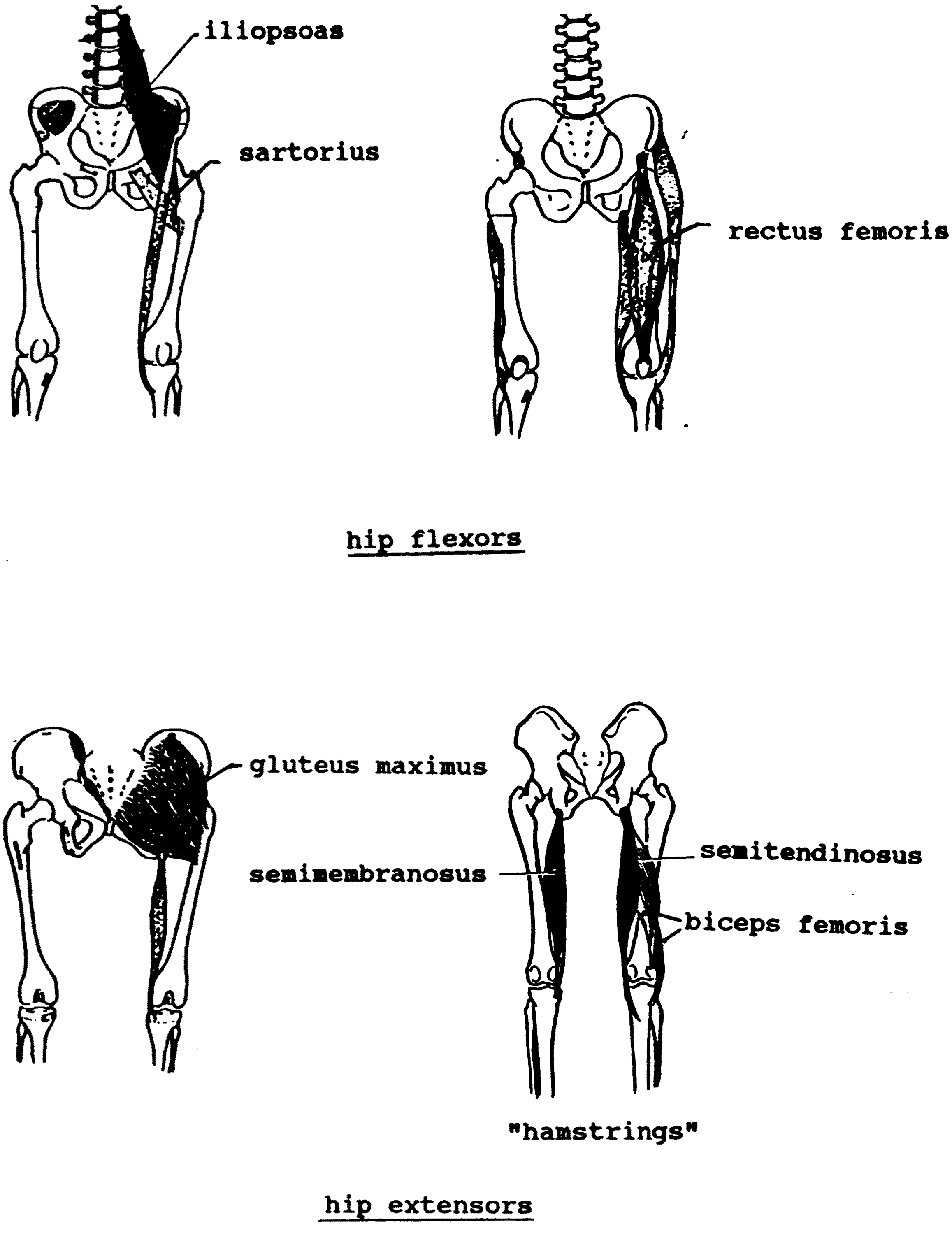


Figure 4.7 : Hip musculature

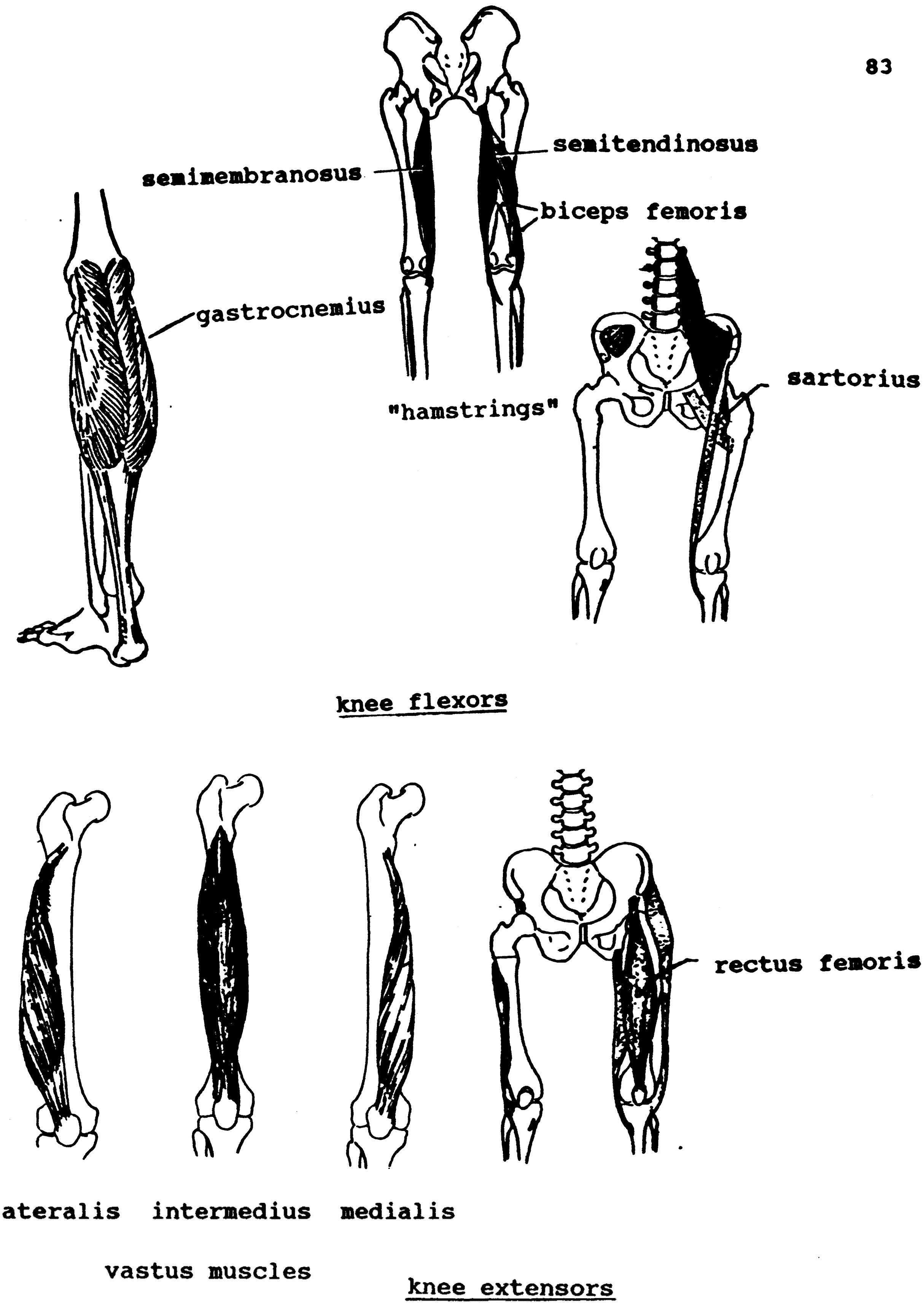
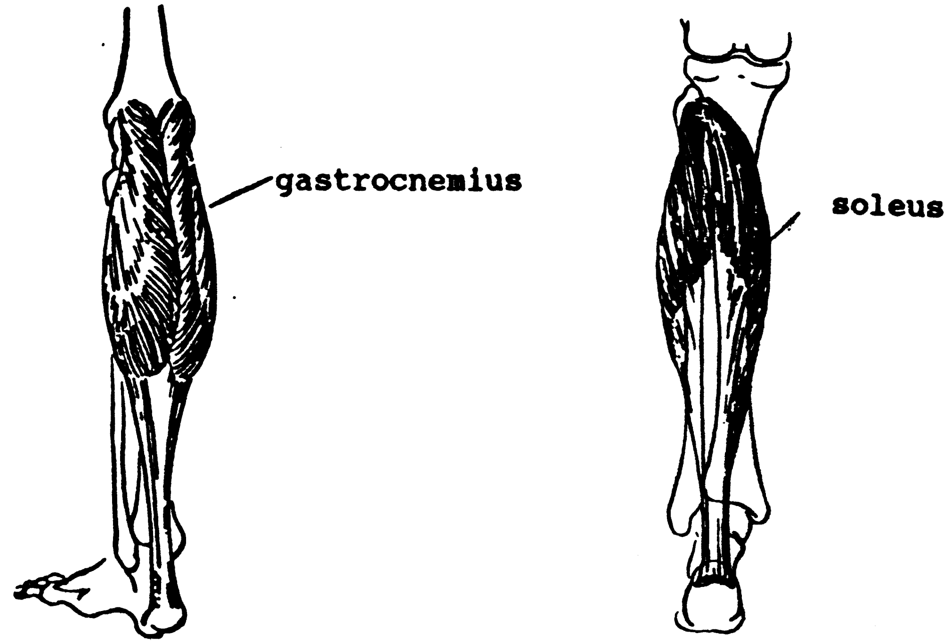
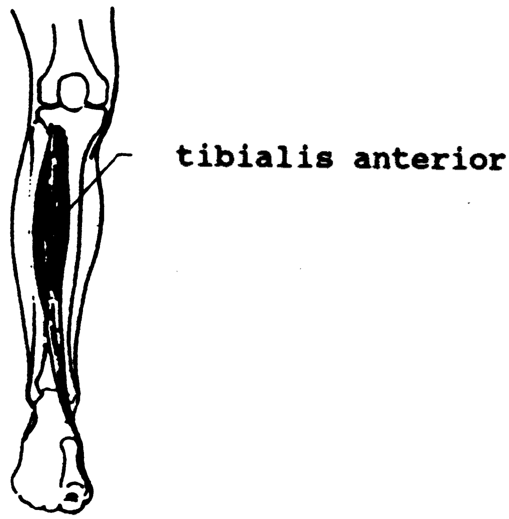


Figure 4.8 : Knee musculature



plantar flexors



dorsi flexors

Figure 4.9 : Ankle musculature

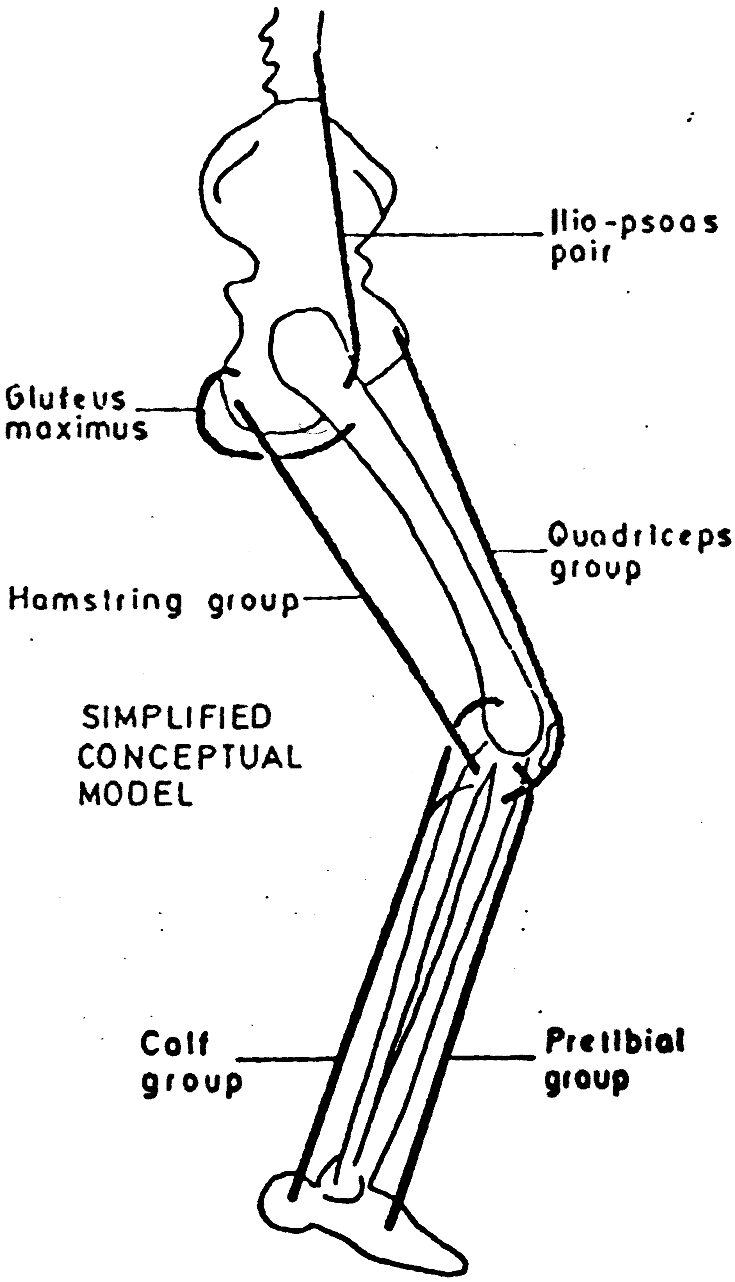


Figure 4.10 : Major lower-limb muscle groups [35]

phase is comprised of two non-support periods separated by a period of support on the contralateral limb, during which the leg is brought forward in preparation for heel strike.

As evidenced by the preceding discussion, the number of muscles involved in ambulation is quite significant. A number of studies using electromyography (EMG) to measure muscle action potential during the running cycle, [11], [12], have been performed, but only on the most major muscles. The inadequacies of the present technical level of EMG techniques has resulted in a lack of significant, comprehensive, qualitative results concerning muscle activity during running.

In addition, it is quite evident that actual physical duplication and replacement of the leg musculature with artificial devices is not a practical solution to the biomechanical problems posed by today's prostheses and their associated gait. The muscular structure and associated power generation are not mechanically reproducible within the confines of a cosmetically acceptable prothesis of reasonable weight. Thus the goal becomes the understanding of the sum of all muscle actions at each the major joints during the various stages of the running cycle, in hopes that the overall resultant actions of the leg musculature during running can be simulated.

To that end, the stance and swing phases will now be discussed in the broadest terms, referring only to the flexors and extensors of the various joints, and to whether positive or negative work is being performed.

4.4.1 Muscle Action in the Stance Phase

In studying muscle activity during the stance phase, it is important to be aware that the effect of gravity acts both as a major agonist and antagonist during this phase, as the joints resist the loading due to the upper body weight. Often only one muscle group, flexor or extensor, is active around a given joint at any one instant.

Several stages of the stance phase are shown in Figure 4.11. Extension and flexion actions of muscles are shown at the joint locations, represented by curved boxes placed at appropriate sides of the joint. A black box indicates positive work, or movement of the joint in the direction of that muscle groups applied force (i.e. knee flexors active and resulting in knee flexion). A white box indicates negative work, or joint movement opposite to the forces applied by the muscle (i.e. knee flexors actively resisting extension of the knee). The absence of a box indicates insignificant muscle activity by that muscle group at that instant in the gait cycle.

For proper understanding of muscle action during running, it is most conducive to study the actions at each joint individually throughout the cycle.

Figure 4.11 : Muscle action during the stance phase [1]

Depending on the running style and speed, the runner may make initial ground contact with only the heel. If this occurs, the foot will tend to plantar flex, with the dorsi-flexors resisting this movement. More commonly, however, the runner will land with weight bearing anterior to the ankle, at which point the plantar flexors will resist the inherent dorsi-flexion. As the body continues to pass over the foot until mid-stance, the foot continues to dorsi-flex, with the plantar flexors acting as the antagonists. Past the point of mid-stance, the plantar flexors become positively active until toe-off, where they contract powerfully to propel the runner forward.

In the knee, the line of action at heel strike passes behind the knee and would tend to collapse it. It is important to note this and also that the knee extensors absorb the substantial impact forces by resisting the knee flexion. The knee extensors continue to provide resistance to knee flexion throughout stance, and begin working positively past the point of mid-stance as they begin to extend the knee to prepare for toe-off. At toe-off, the knee extensors attempt to stabilize the knee in an extended position to realize the maximum benefit from the drive from the ankle, while the knee flexors attempt to initiate knee flexion in preparation to enter the swing phase. The net result is a slight flexion movement of the knee from an almost extended position.

The hip moves in a continuous extension motion from heel strike to toe-off. The hip extensors provide impact absorption at heel strike, but both flexors and extensors work throughout stance to stabilize the upper trunk and position it properly over the legs. The extensors are the primary support group prior to mid-stance, while the hip flexors provide the majority of support in latter stance, and become increasingly active at toe-off as they prepare to bring the leg forward into swing.

4.4.2 Muscle Action in the Swing Phase

Stages of the swing phase and the associated muscle actions are shown in Figure 4.12.

It is important in the study of the swing phase to understand that the magnitude of the muscle forces generated in swing are much smaller than those of stance. The limb is free-swinging, with very little friction existing within the joints.

Very little muscle activity occurs at the ankle during swing. From toe-off to mid-swing, minimal dorsi flexion action occurs to resist the natural tendency of the foot to planter flex due to its inertia. Approaching mid-swing, the foot actively dorsi flexes to aid in ground clearance and shorten the over-all moment arm of the leg, and the foot remains flexed until heel strike.

toe-off mid-swing heel-strike

Figure 4.12 : Muscle action during swing phase [1]

There is also very little muscle activity at the knee during the swing phase. Following toe-off, the knee flexes naturally as the hip flexors accelerate the thigh forward. This knee flexion is initially controlled by the knee extensors. Throughout mid-swing, however, there is little muscular activity of any kind as the knee is allowed to swing through naturally. Towards the end of swing, as the thigh is decelerated by the hip extensors, the shank begins to swing into full extension due to its inertia; this movement is resisted and controlled by the knee flexors. Immediately prior to heel strike, the knee extensors act momentarily to quickly bring the knee into the correct position for the impact.

Following toe-off, the hip flexors contract powerfully to accelerate the thigh forward. Through mid-swing, the thigh is allowed to swing through with minimal muscle activity. In the final stages of swing, the hip extensors contract to decelerate the thigh, bring the knee into extension, and position the leg for heel contact.

CHAPTER 5

PROSTHESIS DESIGN CRITERIA

Having established the basic kinematic patterns of the lower limbs during the running cycle (Chapter 3) and the mechanics associated with each of the lower limb joints necessary to produce that pattern (Chapter 4), the actual design of the prosthesis was considered. Before the design process could begin, however, it was necessary to develop a more replete set of design criteria, both specific to this design, and relevant to prosthesis design in general.

Radcliffe, [43], summarizes the goals of any prosthetic design in the following manner:

> "In the fitting of any artificial limb, the goal of the prosthetist is simply to restore to the amputee the ability to perform everyday activities in an easy, natural and comfortable manner. The basic requirements are therefore three in number-comfort, function and appearance, the latter embracing both cosmetic appearance and appearance in use. Unless a prosthesis is reasonably comfortable, the amputee will be unable to wear it. Unless it performs the necessary functions with reasonable ease and dexterity, the amputee is not apt to find the device very useful. Unless it is reasonably acceptable cosmetically, and unless it can be operated in a natural manner, the limb is likely to be disagreeable both to the wearer and to his

friends and associates."

Although such a set of requirements appears rather simplistic, they are very much interrelated; in any prosthetic design, a number of compromises must be reached between these three priorities. The most common prosthetic design conflict arises between the need for function and the desire for cosmesis. A number of authors, [5], [29], [30], [43], have discussed the importance of aesthetics in prosthesis design while acknowledging the primary necessity for functionality. It is recognized that the need for basic cosmetic standards limits the flexibility in the design process, however, a functional prosthesis which is cosmetically unacceptable by the amputee will not be worn. On the other hand, the designer can utilize the fact that until a minimum cosmetic level is reached, the amputee will accept a decreased level of cosmesis in exchange for increased functionality.

In the design of a running prosthesis, the initial priority is functionality. Amputee running is a relatively new area of study, and it is of primary importance to produce functioning prototypes of running prostheses in order to establish more extensive design criteria to act as the basis for the evolution of an improved generation of running prostheses. In addition, since running is a specialized amputee activity, it is not unreasonable to design a specialized running prosthesis primarily for function, with the assumption that the amputee possesses a more cosmetic limb for everyday use.

There are many general criteria that must be considered during the design of a prothesis and its subsequent evaluation. The foremost priority is the maximization of function. In the past, prosthetists have defined functionality as the prosthesis' ability to mechanically simulate the function, (and functional appearance), of the lost limb. More recently, however, it has been recognized that the human body does not necessarily perform all functions in the most efficient or optimal manner, and that the most functional prosthesis may result from a design which does not directly simulate the biomechanical action of the lost appendage. Therefore, the most accurate method of determining the functionality of the prosthesis can be derived from the amputee himself. In fact, the physical appearance of the amputee's gait to an external observer becomes regarded as a cosmetic factor; the true determination of a prothesis' functionality lies in whether the amputee "feels" like he is walking or running properly, and is doing so in comfort.

Comfort, in terms of prosthetic design, encompasses a number of factors, most obvious of which is the physical comfort of the stump within the socket during use. Other factors, however, include the minimization of induced stresses on the remainder of the body, and the overall maximum efficiency, both cardio-pulmonary and muscular, associated with use of the leg. Also, due to the external, inanimate nature of the prosthesis, the amputee perceives the prosthesis as being heavy, although it is often only a fraction of the weight of the missing limb. Thus, overall prosthetic weight must always be minimized to maximize comfort, without, of course, compromising the strength and durability of the device.

It is also desirable to minimize other distracting factors such as excessive operating noise.

A number of practical mechanical aspects of design must also be considered. Lower limb prostheses are often in continued, constant service, and must be designed with sufficient strength and durability to allow the limb to perform for a number of years with a minimum of maintenance. A major complaint of present wearers of above-knee prostheses is inadequate durability of prosthetic joints and feet [40].

As previously mentioned, new prostheses should be designed such that they are compatible with the existing modular prosthetic standards, to further broaden the current flexibility in prosthesis prescription. The design should be as simplistic as possible to aid in manufacture and assembly, allow the greatest ease in component replacement and service, and, most importantly, to minimize the cost. Lower limb prostheses must also endure harsh operating conditions, (i.e.

water, salt, extremes of temperature), and must be designed for such, including good corrosion resistance.

When considering the design of a running prothesis in particular, an additional set of more specific design criteria are generated.

As stated, the primary design goal of an above-knee running prothesis is to allow the amputee to achieve a comfortable, one-to-one running gait, and thus afford the amputee participation in all activities where running ability is a basic criteria.

Ideally, the amputee should be able to utilize the same prosthesis throughout the day for all ambulation, including walking and running, with little or no prosthetic adjustment. It is unlikely, however that this would be a possibility in the initial development of a running prosthesis.

Amputee running also introduces significant impact loads at heel-strike, which were previously unexperienced during walking. For an average bi-lateral runner, the peak force reached at heel-strike is between two and three times the runner's body weight [3], [37]. It can be assumed that the impact loads for an above-knee amputee will be as great, or higher as evidenced by the Terry Fox Jogging Prosthesis [7]. An above-knee running prosthesis must be designed to aid in the absorption of these impact forces, to minimize the induced stresses on the stump and upper torso. The action of the prosthesis upon impact must be smooth, with no inherent jarring.

Hip, or pelvic, oscillation in the sagittal plane is an intrinsic characteristic of both running and walking. This oscillation is sinusoidal in nature, when studied in conjuncture with forward movement. It is important that a running prosthesis incorporate a simulation of hip oscillation for the amputee, to again simulate the "feel", or perception, of running.

Having consolidated the aforementioned general prosthetic design criteria, specific design work on a functional running prosthesis was initiated.

CHAPTER 6

PRELIMINARY DESIGN

A fundamental summary of the resultant actions of the major muscle groups surrounding the three primary lower-limb joints during the running cycle was presented in Chapter 4. The objective for the prosthesis design was designated as a bio-mechanical simulation of these joint actions within an above-knee prosthesis.

Upon initial inspection of these joint actions by a casual observer, a mechanical simulation may appear trivial; the actual movement of the leg, when taken as a series of linkages, is uncomplicated. When this motion is coupled with the necessary support and swing characteristics of the joints during the various stages of gait, however, the mechanical requirements become more complex. Meeting these requirements within design constraints such as cosmesis and minimal weight, the running prosthesis becomes a challenging design task.

It was necessary to make an initial design decision concerning the energy characteristics of the prosthesis to be designed. Prosthetic limb designs can be divided into three major groups: generative, passive and conservative. These terms describe only the power characteristics, or lack thereof, of the prosthesis itself. In all cases, there is some power transferred to the prosthesis from the upper body and hip.

As previously discussed, the muscles in an intact limb can act both as power absorbers and power generators. Theoretically, the generative aspect of muscle activity can be simulated in a prosthesis using electrical, hydraulic or pneumatic devices, with internal or external power sources. Advances in EMG technologies are also being made, raising the possibility of coupling such generative devices with a bio-feedback system. In such a scenario, sensors would be placed in the sound limb to continuously monitor the magnitude and timing of the actions of the major muscle groups during the running cycle, and this information would be analyzed in real time in order to act as the immediate controlling criteria for the prosthesis. In this way, the actions of the sound limb would, after an appropriate delay, be "echoed" by the prosthesis. Such a situation is not practical within the scope of today's technology, however. The most accurate form of EMG sensor is intramuscular, and thus obviously not practical for everyday use. It is also difficult to predict the final bulk of the necessary microprocessing hardware to analyze the EMG data. The greatest restriction, however, would most probably arise from required energy sources. A prosthesis with external power sources is inadequate for use in other than limited, laboratory exercises. Presently available power sources contained within a prosthesis would far exceed acceptable limits on prosthetic bulk, weight and cosmesis. Therefore, although a controlled, generative prosthesis such as the one described is a design objective for the future, it remains impractical within the limits of present technologies.

Having determined the impracticality of a truly generative prosthesis, the remaining alternatives were examined. The prosthesis could be designed to be passive, similar to the majority of presently available conventional prosthesis for walking. These devices are generally quite inflexible, however, with minimal impact absorption abilities and no generative qualities. Conventional prostheses have already proved inadequate for above-knee amputee running, and thus the passive prosthesis design approach was immediately rejected.

The final alternative, which was ultimately pursued, was the concept of a conservative above-knee prosthetic system. In the intact limb, a majority of the body's kinetic and potential energies , (such as the impact energies at heel-strike), are absorbed and dissipated by the muscles, only to necessitate further muscular power generation later in the cycle to perpetuate the forward motion, (such as at toe-off). In a conservative prosthesis, there is an attempt to conserve and store energies such as those generated at heel-strike so that they may be released as necessary throughout the remainder of the gait cycle. In such a prosthesis, care must be taken to minimize operational losses, both in absorption and generation, in order to maximize the levels of power available.

The initial design process consisted of a study of each of the major lower limb joints throughout both stance and swing to determine the necessary prosthesis characteristics and the corresponding mechanical requirements.

In the majority of above-knee amputations, both the hip flexors and hip extensors remain intact and functional, although to varying degrees. In the design of an above-knee running prosthesis, it must be assumed that these muscle groups are capable of producing flexor and extensor moments about the hip comparable in magnitude to those produced in the unaffected limb. This assumption may seem unrealistic upon initial inspection due to the muscular atrophy which inevitably occurs following such surgery, and is often propagated by the permanent, marked decrease in activity level of the amputee from that point onward. If necessary, however, it is not unreasonable to expect that specific strength training for the hip musculature could be prescribed to prepare the amputee for running. As the amputee becomes more active through utilization of the running prosthesis, hip musculature strength and endurance should increase further. In addition, advances in socket design, such as the CAT-CAM socket, should allow the amputee to use his hip musculature more effectively than was possible with previous conventional sockets. Therefore, for design purposes, it is not unreasonable to assume that the hip musculature is fully functional.

Knee function was then studied through both the stance , (see Figure 4.11), and swing, (see Figure 4.12), portions of the running cycle. At heel-strike, the knee absorbs the majority of the impact forces through a controlled flexion of the knee. The knee then remains slightly flexed and stable throughout the remainder of stance. This action can be simulated mechanically quite simply by utilizing a conservative, resistive device of sufficient strength either anterior to the knee, through the use of an extension spring, or posterior to the knee with a compression spring. Alternatively, a resistive torsion spring could be incorporated into the knee. As often occurs in such design scenarios, a myriad of possible solutions exist, with varying degrees of functionality and practicality. The Physiological Jogging Prosthesis, [6], utilizes a compression spring posterior to the knee to simulate the knee's impact absorption characteristics, with results that have not yet been properly determined.

A knee design difficulty arises when one examines the actions at toe-off, however. Until the instant when the body weight is fully removed from the leg, the knee must resist flexion, remain stable and provide full support. The moment the weight is removed, however, the knee must flex freely as the thigh accelerates into the swing phase, providing adequate clearance for the shank. The Physiological Jogging Prosthesis, (Figure 2.25), attempts to solve these contrary requirements by using a latching system which "senses" the instant of toe-off and unlocks the knee flexion resistance, allowing the shank to swing freely.

Once the knee is swinging freely, it is assumed that the inertial characteristics of the prosthesis will allow it to swing similarly to the natural limb, with little or no active control being necessary. The incorporation of a swing phase control into the prosthetic knee, however, would allow optimal adjustment of the swing phase characteristics of the knee.

The ankle, similar to the knee, may remain quite inactive during the swing phase, (see Figure 4.12), provided that the knee flexion is sufficient to provide adequate ground clearance for the foot. Assuming that such clearance exists, due to an adequate prosthetic knee design, it is the ankle's stance phase characteristics that become of priority. Immediately following heel-strike in the stance phase, (see Figure 4.11), the ankle provides support by resisting the naturally occurring dorsi flexion of roll-over. At toe-off, the ankle provides powerful plantar flexion to propel the amputee forward.

As previously mentioned, the introduction of the Seattle Foot has been innovative as a energy conserving alternative to conventional prosthetic feet. Using a leaf spring incorporated into a prosthetic foot, it absorbs the roll-over energies in the form of the dorsi flexion deflections during roll-over, and then releases them in the form of plantar flexion at toe-off. Once again, although there exists a myriad of ways to mechanically imitate the action of the Seattle Foot, such as the incorporation of a torsion spring into a prosthetic ankle, it is unlikely that any alternate approach could match the functional simplicity of the present design.

Roll-over energies alone, however, represent only a minor fraction of the energies available for conservation and release during the gait cycle. The major source of available energy is the impact forces generated at heel-strike, and the Seattle foot makes little or no use of these energies. Based on the previous discussion, and assuming adequate swing characteristics, the two primary design criteria for an above-knee running prosthesis become evident. First, impact energies at heel-strike must be absorbed and conserved. Secondly, powerful plantar flexion is necessary at toe-off. Finally then, when viewing the running prosthesis as a conservative system, these two design criteria become interdependent; the primary goal of a running prosthesis design becomes the conservation of impact energies and the utilization of those stored energies to provide forward propulsion at toe-off.

Upon inspection, it became evident that there were two elementary methods to achieve impact force absorption. As mentioned previously, the first would be physiologically correct, as in the Physiological Jogging Prosthesis. The knee would be allowed to flex upon impact, with the impact energies being absorbed by a spring actively resisting the knee flexion. The foremost inherent difficulty with this design scenario is the fact that at toe-off, the drive provided by the impact energies will be in the form of knee extension rather than plantar flexion. In a normal, intact limb, knee extension provides only a fraction of the total forward drive at toe-off. The Physiological Jogging Prosthesis was designed such that the resultant forward drive provided by the knee extension at toe-off would be equivalent to that normally provided by plantar flexion, however, the actual performance characteristics of the prosthesis have yet to be determined.

A modification of this design concept could involve absorbing the impact energies in the form of knee flexion, but then to transfer that absorbed energy to the ankle for release in the form of plantar flexion. The mechanical impracticalities of such a design placed it beyond the scope of this work.

The second alternative for impact energy absorption would be a linear absorption device such as that incorporated into the shank of the Terry Fox Jogging Prosthesis. In such a design, the prosthetic shank is free to shorten, or "telescope", upon heel-strike, while a compression spring actively resists the shank collapse and thus absorbs the impact energies. In such a design, care must be taken to ensure that the decrease in the overall length of the prosthesis during stance does not result in an excessive pelvic drop for the amputee on the amputated limb.

Another design constraint must be considered regardless of the manner of impact absorption. In all cases, the absorbed impact energies must be controlled and stored until the proper moment of release at toe-off. If the impact energies are not stored, premature release will occur, as evidenced by clinical trials of the Terry Fox Jogging Prosthesis. Upon heel-strike, the telescoping shank assembly absorbs the impact forces, reaching a maximum of approximately three times body weight. Immediately following heel-strike, however, the only resistance to that stored energy is the amputee's body weight as he passes over the prosthesis during mid-stance, and thus the stored energy begins to be released. In addition, this release can actually occur before the point of mid-stance, and have the effect of driving the amputee upwards and backwards. During testing of the Terry Fox Jogging Prosthesis, it was found that the amputee compensated for this fact by tending to "vault" over the prosthesis, utilizing an excessive drive from his intact limb, such that heel-strike actually occurred with the prosthesis in a vertical, mid-stance position. Therefore, even though the impact energies began to be released immediately after heel-strike, they still contributed to a vertical and forward propulsion of the amputee. If the impact energies could be stored, however, it would allow the amputee to enter normally into heel strike, while delaying the energy release until the optimal instant at toe-off.

Having designated the design goal of active prosthetic plantar flexion at toe-off, it was decided that a linear displacement impact absorption mechanism should be pursued. A sketch of the initial design concept is shown in Figure 6.1. The lower assembly is configured such that both the shank and the spring's internal sleeve "telescope" to allow the spring to absorb the impact energies, and then subsequently allow absorption of the deflectional energies generated by the dorsi-flexion of the foot during roll-over. At toe-off, these energies would be released by the spring in the form of dorsi flexion.

The design presents several inherent problems, however. First, assuming that the impact forces stored in the spring at heel strike will be of the order of three times

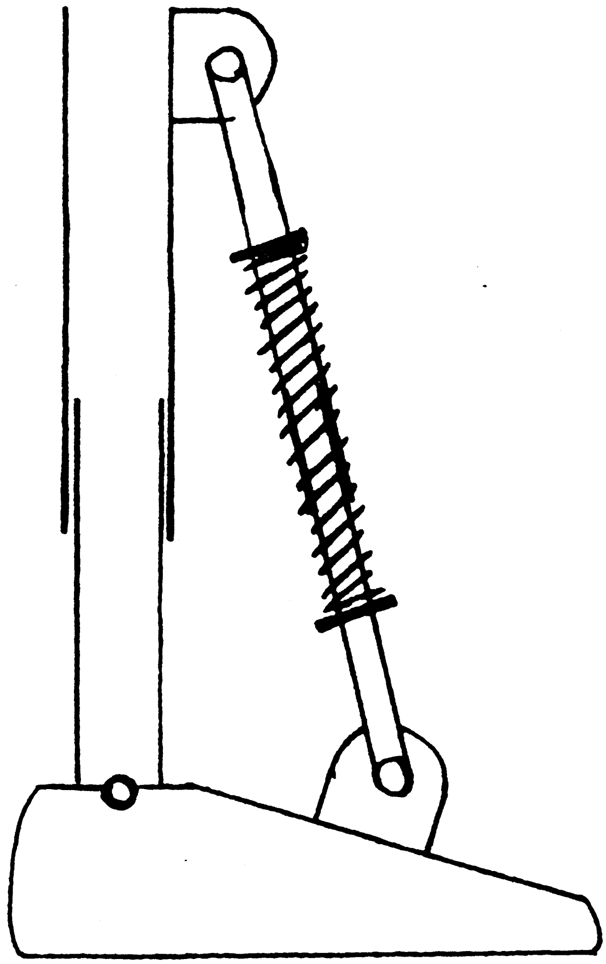


Figure 6.1 : Original design concept

body weight, then the forces generated during roll-over, (approximately one body weight), will be insufficient to result in any further deflection of the spring. The spring, therefore, will actively prevent dorsi-flexion during roll-over and result in a maladaptive gait.

Secondly, in order for the foot to extend into a truly plantar flexed position at toe-off, the unloaded foot must begin in a plantar flexed position at heel-strike. This could result in severe clearance difficulties during swing.

In addition, this design requires that the ankle be locked upon heel-strike, since if it was able to rotate upon impact, the shank would telescope but the spring would resist the impact forces, and, instead of telescoping, would tend to drive the foot into plantar flexion. The ankle would have to be subsequently unlocked, obviously, to allow for the plantar flexion required at toe-off.

It is intuitively evident that the spring will be most efficient in its absorption of the impact energies if it is situated parallel to the shank at heel-strike. Also, the stored energies will be most efficiently used if the point of release of the impact energies at toe-off lies as close as possible to the base of the prosthetic shank. It was this fact that led to the initial concept of the following running prosthesis design.

The initial design proposal is shown in Figure 6.2. This design incorporates a telescoping shank and spring

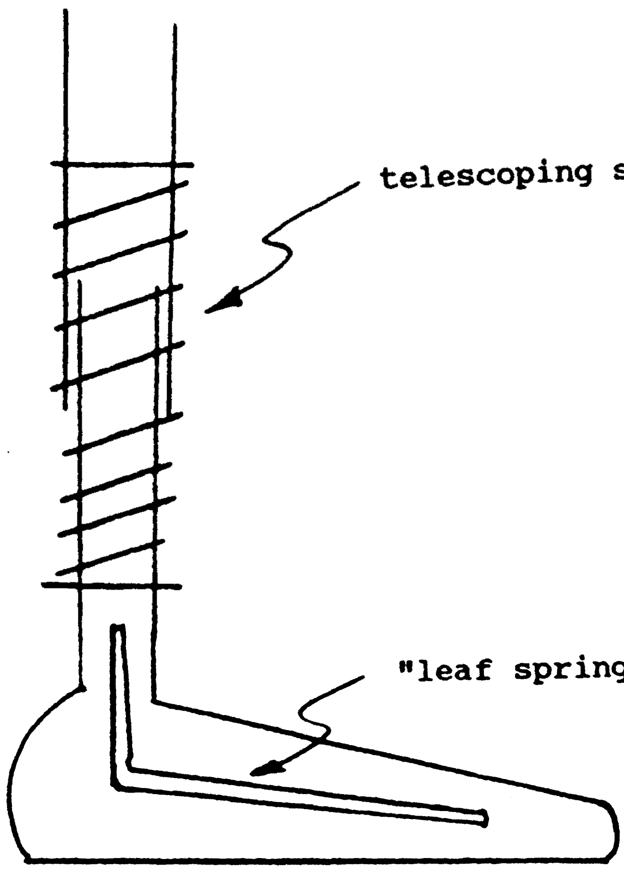
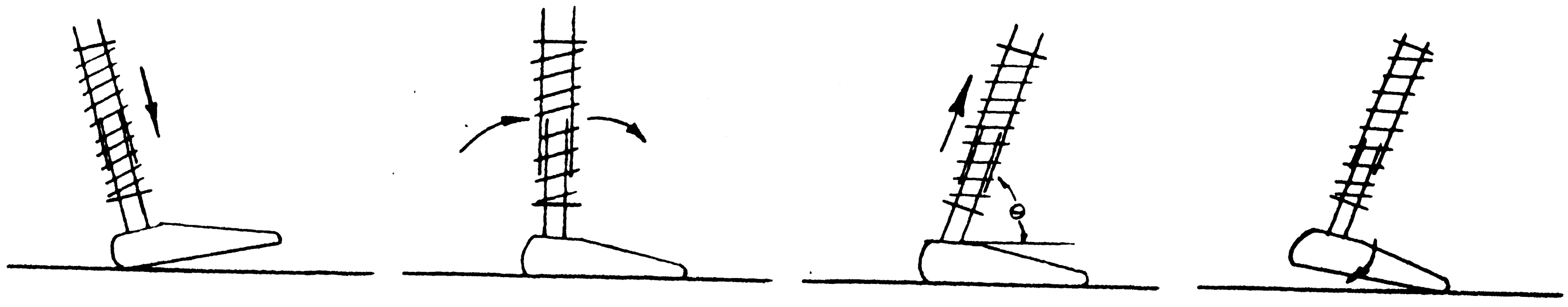


Figure 6.2 : Initial concept of shank-foot assembly

assembly, including a mechanism to provide energy storage, in conjunction with a "leaf-spring" type prosthetic foot.

The theoretical operation of the system is illustrated in Figure 6.3. At heel-strike, the impact forces are absorbed by the shank assembly, and this compression is locked, and the energy stored, at the point of maximum compression. Through mid-stance, the prosthetic foot absorbs the roll-over energies as it would if it was incorporated into a conventional prosthesis. In this way, both impact and roll-over energies are collected and stored in the most efficient manner. When the prosthetic foot has dorsi flexed to a prescribed angle, the impact forces stored in the shank would be released, driving the amputee forward and simulating plantar flexion. The magnitude of these released forces would be sufficient to maintain the prosthetic foot in a flat, stable position on the ground. As the drive from the shank is exhausted, and the weight is being removed from the prosthesis, the roll-over energies stored in the foot would be released, resulting in a final secondary forward propulsion of the amputee.

There are many inherent advantages to this design. It is relatively simple in mechanical nature in comparison to many prosthetic designs, and is capable of incorporating commercially available components such as the Seattle Foot. Also, although the kinematic action of the limb is not physiologically correct, the amputee's perception should be



heel-strike

At heel-strike, the impact forces are absorbed by the compression of the shank assembly. At the point of maximum compression, the assembly is locked in position and the energy is stored.

roll-over

The energy potential resulting from the dorsi flexion of the foot during roll-over is stored in a leaf spring incorporated into the prosthetic foot.

late stance

When the foot has dorsi flexed a prescribed amount, the shin assembly lock is released, propelling the amputee upward and forward. The force resulting from this release is sufficient to keep the foot stable and flexed on the ground, thus retaining the roll-over energies.

toe-off

As the drive from the shin assembly nears completion and weight is being removed from the prosthesis, a secondary propulsion of the amputee will result from the release of the roll-over energies stoted in the foot.

Figure 6.3 : Theoretical operation of shank assembly

that he is running with a physiologically correct gait, including a natural pelvic drop at heel-strike, (due to the spring's compression), and an effective drive from the ankle at toe-off.

Assuming the release angle remains adjustable, this design is also quite effective from the perspective of training the amputee for its use. In this way, during training, the energy storage mechanism can initially be disconnected, resulting in the prosthesis operating similar to the Terry Fox Jogging Prosthesis. As the training, and ability, advances, impact energy storage can be introduced, and, as proficiency with the prosthesis increases, the angle of dorsi-flexion necessary to trigger the energy release can be increased until optimal gait characteristics are obtained. It is also possible that the prosthesis could be adjusted such that it would be a preferential walking prosthesis.

Having specified the design concept parameters, detailed design work was initiated.

CHAPTER 7

RUNNING PROSTHESIS DESIGN

Having determined the conceptual and functional basis for the design of the running prosthesis, detailed design work was initiated. The development of prosthesis was an iterative process, beginning with the determination and imposition of the least flexible design constraints and the adaptation of subsequent design components within the increasingly restrictive constraints.

The following subsections describe the specific nature of the design constraints, in the order of their introduction into the design, and illustrates their effect on the development of the final prosthesis configuration.

7.1 Over-all Dimensions

For practical testing purposes, this prosthesis was developed for a specific, individual amputee, Mr. Grant Darby. Although the design process was conducted such that it could be generalized to accommodate any amputee, the primary thrust of the research was to develop a prototype to allow testing of the functionality of the proposed design through actual field testing. Mr. Darby was involved in the testing of the Terry Fox Jogging Prosthesis from its inception, and had direct experience running with a prosthesis which incorporated a collapsible, shock-absorbing shank. For these reasons, Mr. Darby was considered an excellent candidate for the field testing of this prototype.

In addition, it was felt that the imposition of actual physical constraints of a specific amputee would add increased realism to the over-all design process.

Measurements taken from the Terry Fox Jogging Prosthesis worn by Mr. Darby indicated an over-all length of twenty and one-quarter inches from the distal edge of the suction socket to the upper edge of the SACH foot (see Figure 7.1).

For the prosthesis being developed, a SACH foot and a suction socket similar to that used on the Terry Fox Jogging Prosthesis were designated for use, and thus this knee-shank length was imposed as the first design constraint on the prosthesis design.

7.2 Knee Selection

The second configural restraint to be placed on the design was the selection of a suitable prosthetic knee.

The necessary performance criteria for the knee was

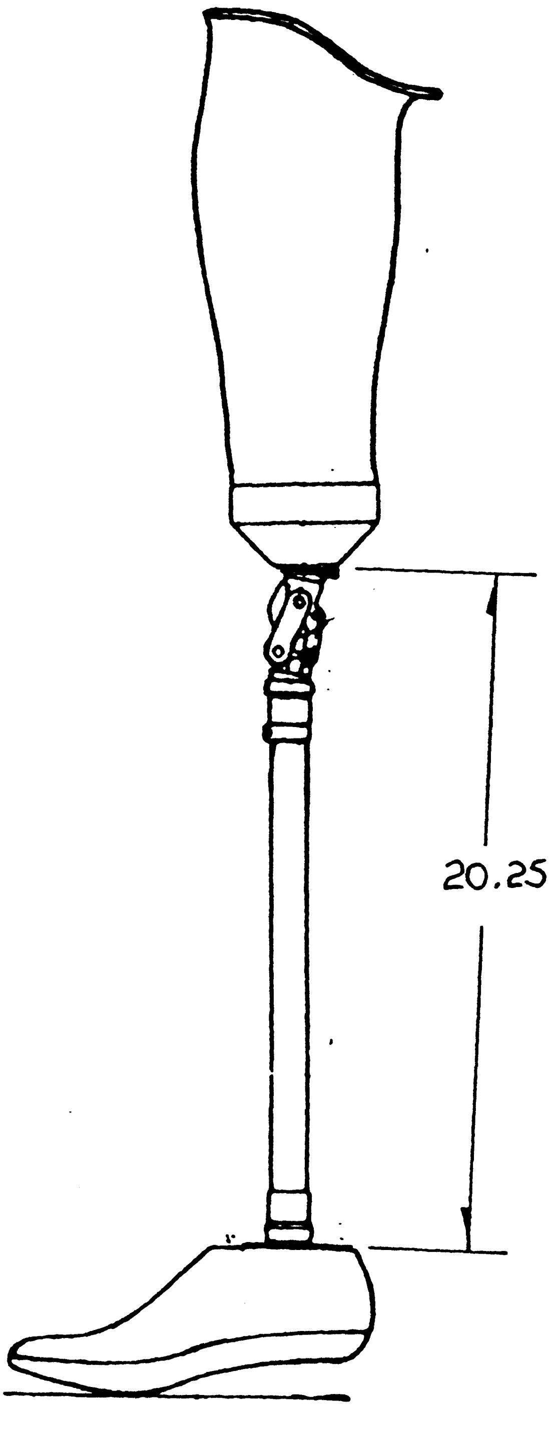


Figure 7.1 : Knee-shank Dimensions of Terry Fox Jogging Prosthesis

based on that demonstrated by the knee used in the Terry Fox Jogging Prosthesis, since the task to be performed by the knee in the new design would be functionally similar to that of the Terry Fox limb. The first selection criteria was the necessary provision of secure stance phase stability throughout the gait cycle. The second was the provision of adequate swing phase control to allow proper extension of the knee for stability upon heel strike.

The knee used in the Terry Fox Jogging Prosthesis was a Teh-Lin four-bar polycentric knee with pneumatic swing phase control. This knee, although not designed for use in a running prosthesis, was found to provide adequate stance phase stability and acceptable swing phase control.

Despite its performance in the Terry Fox Jogging Prosthesis, however, the Teh-Lin unit was found to be unacceptable for use in the new prosthesis design due to its substantial size. As stated, the entire working length of the new knee-shank unit to be designed was constrained to twenty inches, and the length of the Teh-Lin, (in excess of nine inches), was deemed excessive and unacceptable for the proposed design.

An investigation of prosthetic knee alternatives indicated that due to the physical requirements necessary for their operation, both pneumatic and hydraulic swing phase control prosthetic knees were, in general, too excessive in size to allow their inclusion in the proposed design.

The knee selected for use in the proposed design was the 3R36 Otto Bock-Habermann Polycentric Modular Knee Joint (see Figure 7.2). This knee was chosen based on its incorporation of the maximum number of desired knee characteristics, while remaining exceptionally compact in size.

The over-all functional length of the knee is approximately three and one-half inches. As stated, this is a polycentric knee and thus affords the amputee the greatest degree of voluntary knee stability. In addition, the knee also incorporates an adjustable extension stop which allows even further adjustment of the degree of stance phase stability upon heel strike.

With regards to swing phase control, although the Otto Bock-Habermann knee does not include a sophisticated swing phase control mechanism, it does incorporate an adjustable mechanical extension assistance device and adjustable knee axis friction. Proper use of these devices should provide adequate swing phase control and adjustment for testing purposes.

Although no commercially available prosthetic knee has been designed for the purpose of running and the forces generated during such activities, the Otto Bock-Habermann knee does incorporate titanium components for increased strength and durability. Ideally, utilizing results of field tests of the Terry Fox Jogging Prosthesis and, eventually,

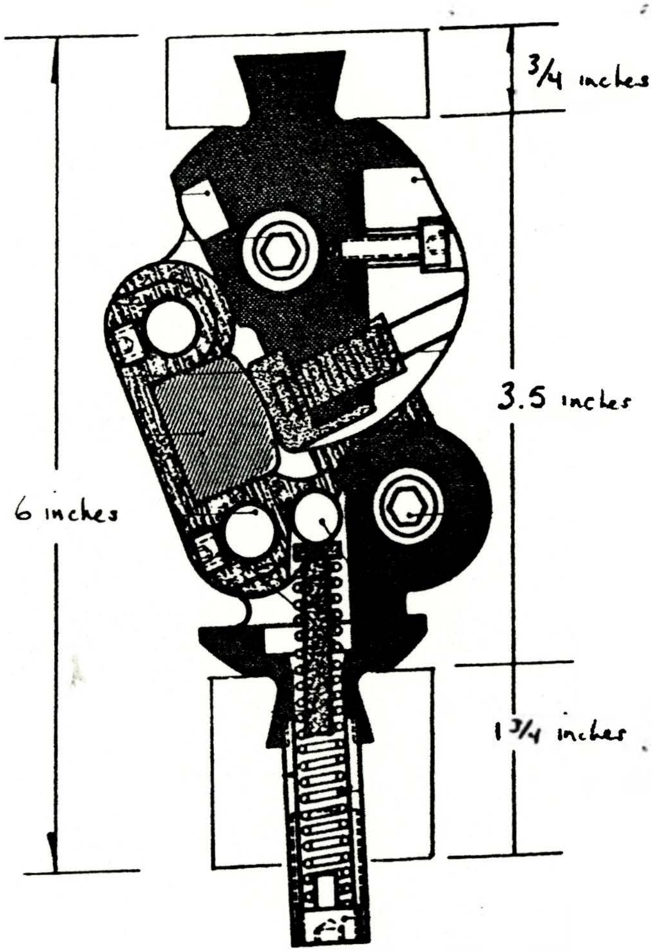


Figure 7.2 : 3R36 OTTO BOCK-HABERMANN modular knee joint [57]

this design, sufficient specific design criteria should be generated in terms of desired swing phase characteristics, stance phase stability and impact forces to allow for the design of an above-knee prosthetic running knee.

As shown in Figure 7.2, the functional length of the Otto Bock-Habermann Knee is approximately three and one-half inches. The upper and lower modular coupling fixtures add approximately two and one-half inches to the length of the knee unit, giving the knee an over-all functional length of approximately six inches. Thus, the remaining design length for the knee-shank unit is reduced to, conservatively, fourteen inches.

7.3 Spring Design

With the selection of the most adequate and compact knee available for incorporation into the knee-shank unit design, the next design constraint imposed was the determination of the characteristics and dimensions of the spring necessary to provide the desired impact absorption characteristics.

It was first necessary to determine the required spring rate to provide the desired hip deflection upon impact, and, having determined that spring rate, a computer package was developed to determine the optimum dimensions of the spring required.

7.3.1 Determination of the Required Spring Rate

The prediction of the magnitude of the maximum impact forces generated by a runner upon and immediately following heel-strike is quite subjective since the magnitude of these forces is directly related to the gait pattern adopted by the runner. It can be shown, utilizing data from the kinematic studies of slow joggers done at the University of Waterloo's Gait Laboratory [56], that during the stance phase the minimum height of the hip coincides with the instance of maximum impact force and is an average of one and one half inches below the subject's stationary hip height with the leg fully extended. A deflection of two inches, however, is within an aesthetically acceptable range.

The maximum force generated during stance was quite consistent between subjects at an average maximum impact force of approximately 2.2 times the subject's body weight [56]. Relating these impact forces to the design of a prosthesis incorporating impact absorption and assuming linear axial compression of the fully extended prosthesis, the spring rate necessary to simulate the hip displacement of a non-amputee jogger under maximum impact forces can be shown to be :

$$K = \frac{2.2 * B.W.}{H.D.}$$

where K - required spring rate
B.W. - subject' body weight (in pounds)
H.D. - desired hip displacement (in inches)

Using a desired hip displacement of two inches, the desired spring rate becomes a function of body weight :

$$K = 1.1 * B.W. \quad (\text{lbs./in.})$$

Assuming the amputee can approximate the gait of a non-amputee runner, a spring rate in the range of one to one and one-half "body weights" per inch should be adequate.

The incorporation of such a spring would also result in an adequate prosthesis for walking purposes. The hip displacement in an average walking, adult non-amputee is most often modelled as a sinusoid with an amplitude of approximately one inch, under maximum impact forces slightly in excess of the subject's body weight. By incorporating a spring with a spring rate of approximately 1.1 times the subject's body weight into the proposed prosthesis, during walking the spring would tend to absorb the impact forces upon heel strike, while resulting in a physiologically correct hip displacement. This hypothesis is supported by Mr. Grant Darby's continued use of his Terry Fox Jogging Prosthesis for all daily activities.

Grant Darby's Terry Fox Jogging Prosthesis is fitted with a spring with a rate of 200 lbs./in. (approximately 1.25 body weights per inch). During initial jogging trials, however, even with an initial precompression of the spring of 160 pounds, at heel strike the spring was found to compress to its solid height, indicating a generated force in excess of 560 pounds. Subsequent testing at the University of Waterloo's Gait Laboratories indicated that average maximum forces of 688 pounds, or 4.3 times body weight were being generated upon heel strike [56]. The magnitude of these substantial impact forces can be accounted for due to the gait style adopted by Mr. Darby while using the prosthesis. While running, Mr. Darby tended to "vault over" the prosthesis prior to heel-strike in such a way that the prosthesis was approximately co-linear with his upper trunk upon impact. Therefore rather than running "through" the leg during stance as would a non-amputee, Mr. Darby tended to vault, raising his upper trunk to abnormal heights and land vertically on the limb in order to avoid the release of the impact energies prior to mid-stance, which would result in an effect counter-active to the forward drive of the amputee. This vaulting action resulted in the generation of the excessive impact forces. As discussed previously, this gait style is characteristic of the Terry Fox Jogging Prosthesis and is one of its greatest shortcomings.

With regards to the selection of an appropriate spring rate for the proposed design, the inherent difficulty lies in the fact that it is difficult to anticipate the magnitude of the impact forces that will be generated until an initial testing has occurred and it has been determined to what extent a natural running gait is being simulated. In ideal testing results, a normal running gait would be closely approximated and a spring rate similar to that originally calculated would be adequate. It is unlikely, however, that a prosthetic limb would approximate the natural gait to such a degree, and it is reasonable to assume that some evidence of the Terry Fox Jogging Prosthesis gait phenomena would be exhibited in any new designs.

Assuming then that the proposed design would result in a closer approximation of the non-amputee running pattern by the amputee than exhibited by the Terry Fox Jogging Prosthesis, it is also reasonable to assume that the impact forces, and corresponding spring rate, would lie between the two extreme values exhibited by a non-amputee and an amputee using a Terry Fox Jogging Prosthesis. This is supported by the fact that since the new design incorporates impact energy storage, there should be no need for the amputee to vault over the prosthesis prior to heel-strike, and thus the magnitude of the impact forces should decrease.

In selecting the desired spring rate, one consideration was that a "conservative" spring rate estimate would emphasize a larger spring rate, or a "stiffer" spring. Although a spring with excessive stiffness might provide a less than adequate hip deflection upon heel strike, this is more desirable than the possibility of a spring of inadequate stiffness compressing to its solid height, requiring the stump and socket to absorb the remainder of the impact forces and irrecoverably dissipating energies which otherwise could be utilized for the forward drive at toe-off.

The final selection of the magnitude of the desired spring rate rests with a subjective judgement of whether the proposed unit's performance will more closely approximate the impact characteristics of the Terry Fox Jogging Prosthesis or that of a normal jogger. Until initial testing has occurred, this is a difficult estimation. For design purposes, a direct average of the spring rates required for the two extreme cases was taken as the desired rate. Thus the initial design spring rate was taken as 1.6 body weights per inch.

7.3.2 Optimization of the Spring Configuration

A helical coil compression spring can be defined by three basic design variables: the mean coil diameter, the wire diameter, and the number of active coils. In many spring design situations, one or more of these variables are set, or constrained, and the required values for the other variables are calculated. In the case where no values are tightly constrained, the process becomes an iterative one with incremental adjustments being made to the variables until an acceptable design results.

For the purposes of this work, however, it was decided to utilize computerized numerical optimization methods to determine the optimum variable values.

In very simplistic terms, in a numerical optimization routine, one aspect of the design is chosen to be maximized or minimized within imposed design constraints.

The program that was developed, SPRING, was based on a static spring design program by J.N. Siddall [51]. The program was significantly modified and extended, however, for the purposes of this prosthesis design. The program is written in FORTRAN and was run on a VAX 8600 computer system, and uses the numerical optimization methods and routines contained in the package OPTIVAR [51].

The software package SPRING, (see Appendix A), was developed to standardize the design of springs to be incorporated into the knee-shank unit of the prosthesis. It is an interactive program that has been designed in such a way as to allow a prosthetist with no fundamental knowledge of mechanical design principles to input data, in real time, concerning the physical characteristics of the patient/amputee and in return be provided with the optimum configuration of the appropriate spring.

The program receives input from the user concerning the subject's initials, weight, and the knee-shank length and also provides the option of having the results written to a data file. The numerical optimization is then initiated.

The spring to be designed is a helical coil compression spring with the ends closed and ground. As previously stated, the basic configuration of such a spring can be defined through three design variables (see Figure 7.3):

d - wire diameter
D - mean coil diameter
N - number of active coils

In the optimization process, it is these variables that are iteratively varied until optimum values are determined.

As previously stated in this work, the minimization of weight is of fundamental importance in any prosthesis design. Due to the passive nature of prosthetic limbs, a prosthesis which in fact weighs only a fraction of the weight of the missing limb can, when worn, actually appear to be much heavier to the amputee. It was therefore decided that the package should be developed with the minimization of weight as the optimization criteria.

The weight of a helical coil compression spring with the ends closed and ground can be calculated using the formula:

$$\text{Spring Weight} = (\pi/4)^2\ (D*d^2*(N+2))*p$$

where p - material density

The material designated for use was chrome-silicon steel wire

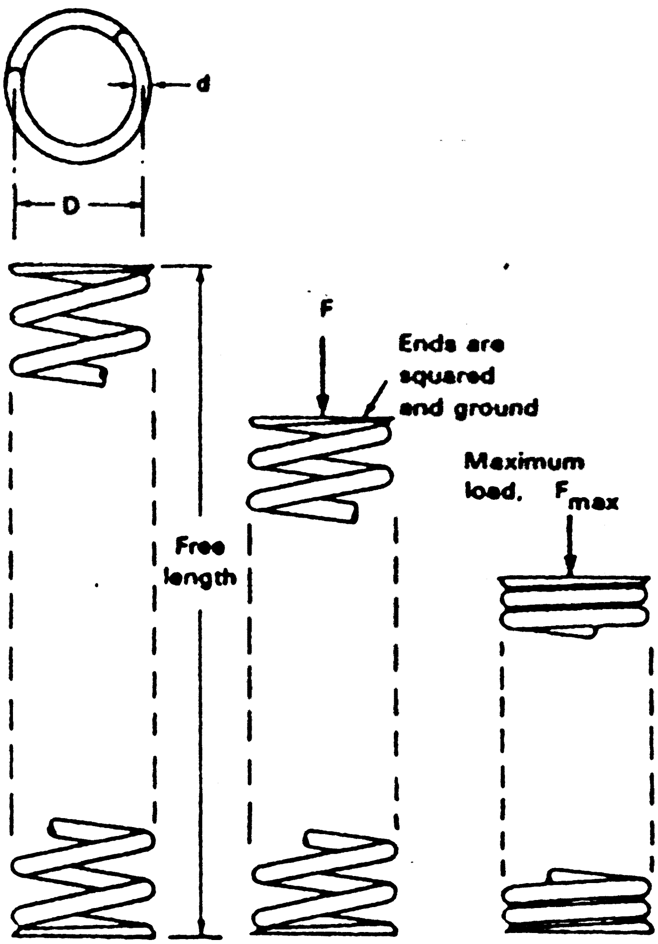


Figure 7.3 : Helical coil spring design variables [51]

due to its common availability and use in spring manufacture as well as its superior strength, fatigue and corrosion characteristics. The values of the steel's material characteristics are:

Shear Modulus, G - 11.5 Mpsi.

Modulus of Elasticity, E - 30 Mpsi.

Density, p - 0.284 lbs./in.

With the implementation of the numerical optimization, the first constraints imposed specify an acceptable range for the final spring rate. As discussed previously, it had been determined that the required spring rate would be 1.6 "body weights" per inch. In terms of the three design variables, however, the spring rate can be expressed as:

$$K = \frac{G*D^4}{8*d^3*N}$$

However, by its nature, the OPTIVAR routines require a constraint "range" within which to vary the design variables. Initially the total allowable range was set at twenty percent of that desired, resulting in values for the maximum and minimum allowable spring rates:

$$KMAX = K + 0.1*K$$

KMIN = K 0.1*K

with the actual final value to be determined by the program to be constrained between these values.

The allowable wire diameter, d, was constrained within the range of commonly available wire sizes.

$$d > 0.032 \text{ inches}$$

$$d < 0.4 \text{ inches}$$

It was also necessary to constrain the allowable size range for the mean coil diameter. For reasons of aesthetics and practicality of design, the maximum allowable mean coil diameter was limited to three inches. The minimum allowable mean coil diameter is constrained by the minimum allowable inside diameter of the spring, and thus is also a function of the wire diameter. The absolute minimum allowable inside diameter of the spring was taken as the minimum allowable outside diameter of the assembly which passes within the spring. Although logistical problems would result if this was found to be a tight constraint in the optimization, it was included in that form for initial runs:

$$D < 3.0 \text{ inches}$$

$$(D-d) > 1.125 \text{ inches}$$

To ensure that the spring is not wound too tightly, Shigley [50] advises that the spring index, "C", the ratio of the mean coil diameter to the wire diameter should be greater than three, thus resulting in the seventh constraint:

$$D/d > 3.0$$

The free length of the spring must also be constrained within the remaining available working length of the knee-shank assembly. Realistically, however, the remaining space could not be entirely devoted to the spring and thus the available space constraint was reduced by another two inches. In terms of the design variables, a spring's free length can be calculated as:

$$\text{Free Length} = 1.05*(N+2)*d + ((4.0*\text{Body weight})/K)$$

and thus the eighth constraint becomes:

$$\text{Free length} > \text{Unit length } 2.0 \text{ inches}$$

The two final constraints limit the amount of stress that may be generated in the spring.

The shear stress must be constrained below the maximum allowable. For chrome-silicon steel wire, the ultimate strength can be calculated as:

$$S_{ut} = A/d^m$$

where A = 202,000 psi. [50]

m = 0.112

but the yield strength:

$$S_y = 0.75*S_{ut}$$

and using the distortion energy theory:

$$S_{sy} = 0.577*S_y$$

and thus the torsional yield strength can be calculated by:

$$S_{sy} = 0.4328*S_{ut}$$

$$= 0.4328*A*d^{-m}/S.F.$$

where S.F. - safety factor (initially = 1.5)

The maximum stress generated in the spring can be calculated by:

$$\text{Generated stress} = \frac{K.S.*8*F*D}{\pi*d^3}$$

where K.S - shear stress multiplication factor

$$= \frac{4*C - 1}{4*C - 4} + \frac{0.615}{C}$$

C - spring index

and F - maximum force experienced

$$F = 2.0*K$$
$$= 3.2*\text{Body weight}$$

and thus the ninth constraint imposed was that the generated stress must always be less than the torsional yield strength.

It was also necessary to design the spring to resist fatigue. The loading characteristics are illustrated in Figure 7.4.

Zimmerli, [50], has reported that the endurance limit for wire sizes under ten millimeters are constant for peened springs at :

$$\text{Endurance Limit} = S_{se} = 67{,}500 \text{ psi.}$$

where this value has been corrected for surface finish and size. Fatigue failure will occur when:

$$_a = S_{se}$$
$$\text{where} \quad _a = \text{Generated stress}/2.0$$

and S_{se} is the endurance limit corrected for reliability, temperature and stress concentration:

$$S_{se} = K_c*K_d*K_e*S_{se}$$

where for this temperature range $K_c = 1.0$

and for 99.0% reliability $K_d = 0.814$

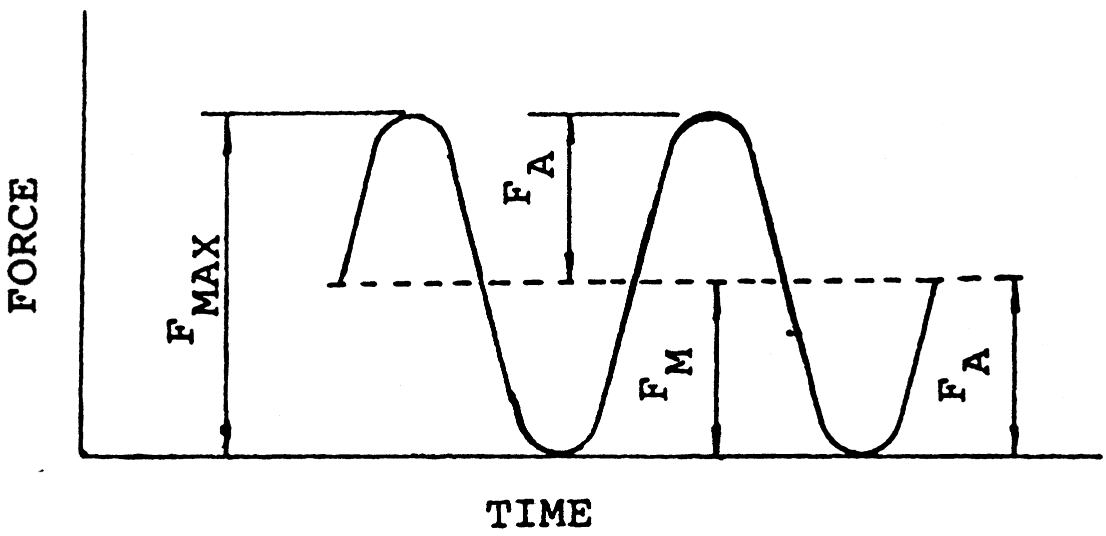


Figure 7.4 : Fatigue loading characteristics

To calculate the stress concentration factor, a spring index value of six was assumed, giving:

$$K = 1.125$$

$$K_s = 1.08$$

$$\text{thus the curvature effect, } K_f = 1.125/1.08$$
$$= 1.1574$$

$$\text{and } K_e = 1.0/K$$
$$= 0.864$$

and thus the fatigue endurance limit became:

$$S_{se} = (0.814)(1.0)(0.864)*S$$
$$= 47,472 \text{ psi.}$$

and thus the final constraint limited half the value of the generated stress to below the value of the fatigue endurance limit.

In running the program, although there are a number of numerical optimization alternatives available in the OPTIVAR software package, this program was found to give the best results when used with the SEEK optimization routine in conjunction with the penalty function OPTIM 1 [51].

After the optimization routine is completed, the routine outputs the final values of the design variables and the associated spring characteristics. An example of the program's output is shown in Figure 7.5.

The program was run using input values for Mr. Grant Darby (a body weight of one hundred and sixty pounds and a knee-shank length of fourteen inches), giving the results shown in Figure 7.5.

It was felt that both the wire diameter and the final unit weight were excessive and thus the constraints were examined in an attempt to improve the optimization results. There were only two tight constraints, (i.e. constraints that actually limited the springs design), and those were the constraints governing the maximum allowable spring rate and the maximum allowable generated stress. It was decided to "loosen" these tight constraints in attempt to realize a preferable spring design.

It was initially considered odd that the program was driven to the maximum allowable spring rate constraint during the optimization, since it was initially assumed that to achieve a minimum spring weight, the programme would attempt to achieve the lowest allowable spring rate. To investigate this characteristic of the optimization, an informal trade-off study was conducted.

In an optimization trade-off study, a tight design constraint is incrementally relaxed to study the subsequent effect on the optimum values of the design values and the objective function. To this end, the maximum allowable spring rate constraint was relaxed, (allowed to increase), to

OPTIMUM SOLUTION FOUND

MINIMUM U = 0.10675161E+02


```
               X( 1) =  0.17751997E+01
               X( 2) =  0.35981268E+00
               X( 3) =  0.16824987E+02
INEQUALITY CONSTRAINTS
             PHI( 1) =  0.13336182E-01
             PHI( 2) =  0.25586670E+02
             PHI( 3) =  0.32781267E+00
             PHI( 4) =  0.40187329E-01
             PHI( 5) =  0.12248003E+01
             PHI( 6) =  0.27519965E+00
             PHI( 7) =  0.19336772E+01
             PHI( 8) =  0.88775396E+00
             PHI( 9) =  0.36210938E+01
             PHI(10) =  0.15125211E+05
```

OPTIMUM SPRING DIMENSIONS FOR SUBJECT GD
****

THE SUBJECTS WEIGHT IS.........160.0 LBS

THE TOTAL UNIT LENGTH IS....... 12.00 IN.

THE MEAN COIL DIAMETER IS......1.78 IN.

THE WIRE DIAMETER IS...........0.360 IN.

THE NUMBER OF ACTIVE COILS IS..16.82

RESULTING SPRING CHARACTERISTICS
****

THE OPTIMUM SPRING RATE IS.....255.99 LBS/IN

THE WEIGHT OF THE SPRING IS....3.03 LBS

THE VOLUME OF THE SPRING IS....10.68 IN.**3

THE SPRINGS NATURAL FREQ. IS...26.07 CYC/SEC

THE SPRINGS FREE LENGTH IS..... 9.11 IN.

Figure 7.5 : Original optimization results

study the effect. It was determined that with every increase in the constraint, the optimum values of the wire diameter and the mean coil diameter remained unchanged while the number of active coils decreased. Upon examination, a possible cause for this unique trend became evident.

The maximum allowable generated shear stress constraint is a function of mean coil diameter and wire diameter but not of the number of active coils. The spring rate calculation, however, is not only a function of mean coil diameter and wire diameter, but also varies directly as an inverse of the number of active coils. What the optimization routine seems to do is find optimum values for the mean coil diameter and wire diameter with regard to the generated stress constraint, and then having set these values, the program minimizes the spring's weight by reducing the number of active coils until the maximum allowable spring rate constraint is reached.

With these results it became evident that the optimum spring's weight could be reduced by increasing the maximum allowable spring rate or increasing the maximum allowable generated stress.

It is possible that testing of the proposed prosthesis could indicate the need for a stiffer spring. However, at this stage of design it was felt that that assumption could not be made, and that the proposed spring rate should remain unchanged. In fact, the first design constraint in the program was altered, making the maximum allowable spring rate exactly equal to the desired rate, in the knowledge that the optimization routine would drive the final spring rate value to this upper limit.

Possible modifications to the maximum allowable generated stress were then examined. As an experimental measure, the safety factor of 1.5 was removed from the stress calculation and the optimization initiated. The resultant spring was quite short in length, and less than half the weight of the originally optimized spring. In this optimization, the fatigue calculation became the tight constraint.

With this information, several subjective decisions were made. This prosthesis was being designed as a prototype to test the viability of the theoretical function of the design. As such, it would not be in service as a daily prosthesis and therefore it was felt that safety factors in the stress calculations could be relaxed. However, if the safety factor was removed entirely, then the spring would experience its theoretical yield stress if the maximum anticipated force was reached. Thus, the safety factor was not removed entirely, but was reduced from 1.5 to 1.1. Also, since the originally input maximum anticipated force was not particularly conservative, an extra measure of safety was incorporated by including an estimate of the maximum anticipated generated force which was closer to that seen by the Terry Fox Jogging Prosthesis, approximately four times body weight. Also with this increase in the maximum force, the linear deflection due to that force would increase to two and one half inches, and thus the free length calculation was adjusted accordingly. With the inclusion of these modifications into the program, it was again run, returning the results shown in Figure 7.6.

It was then decided that the optimum values for the spring should be altered to conform to industry standards for spring manufacture. The resultant optimum wire size was not one standardly available. Also, the optimum number of active coils returned was 14.79, whereas the smallest practical division for manufacturing purposes is one quarter of a turn. Therefore, for practical purposes of manufacture, the results of the optimization are not usable.

The program was then modified in order that it would output a optimal spring design within the abovementioned manufacturing constraints.

The program was altered such that after the program had completed an initial optimization, it would then examine the resultant wire size and then determine the closest standard wire size. That wire size was again input into the optimization routine to determine the new values for the mean coil diameter and the number of active coils.

With this new optimum determined, the program was then modified such that the value of the number of active

```
                    OPTIMUM SOLUTION FOUND

                  MINIMUM  U =  0.25221062E+01

                       X( 1) =  0.17718005E+01
                       X( 2) =  0.34787461E+00
                       X( 3) =  0.14785918E+02
INEQUALITY CONSTRAINTS
                     PHI( 1) =  0.20019531E-01
                     PHI( 2) =  0.25579987E+02
                     PHI( 3) =  0.31587461E+00
                     PHI( 4) =  0.52125394E-01
                     PHI( 5) =  0.12281995E+01
                     PHI( 6) =  0.29892588E+00
                     PHI( 7) =  0.20932159E+01
                     PHI( 8) =  0.38684793E+01
                     PHI( 9) =  0.17328125E+02
                     PHI(10) =  0.72248672E+04




.OPTIMUM SPRING DIMENSIONS FOR SUBJECT GD
*********************************************


THE SUBJECTS WEIGHT IS.........160.0  LBS

THE TOTAL UNIT LENGTH IS....... 14.00  IN.


THE MEAN COIL DIAMETER IS......1.77  IN.

THE WIRE DIAMETER IS...........0.348  IN.

THE NUMBER OF ACTIVE COILS IS..14.79


RESULTING SPRING CHARACTERISTICS
******************************


THE OPTIMUM SPRING RATE IS.....255.98  LBS/IN

THE WEIGHT OF THE SPRING IS....2.52  LBS

THE VOLUME OF THE SPRING IS.... 8.88  IN.**3

THE SPRINGS NATURAL FREQ. IS...28.58  CYC/SEC

THE SPRINGS FREE LENGTH IS..... 8.63  IN.
```

Figure 7.6 : True optimum values

coils is then examined and adjusted to the nearest quarter of a turn. Since this adjustment makes such a slight difference in the springs over-all characteristics, the other design values are not re-optimized, rather the output sequence is initialized and the spring characteristics are calculated using the previously determined values for the wire diameter and mean coil diameter, and this new value for the number of active coils.

With these modifications completed, the program was initialized for Mr. Darby's input characteristics, with the final spring design results shown in Figure 7.7.

7.4 Final Unit Design

Having specified both the spring and knee to be incorporated into the prosthesis design, the design of the final unit configuration was initiated.

The design process was an arduous and iterative one, due to the unit's size constraints as well as other common manufacturing and assembly constraints associated with prototype productions.

A comprehensive summary of the iterative design process is beyond the scope of this work, however it is important to briefly discuss the major design considerations leading to the final unit configuration shown in Figure 7.8.

As previously discussed, the unit was configured in

```
                         OPTIMUM SOLUTION FOUND

                       MINIMUM  U =  0.25427125E+01

                          X( 1) =  0.16818005E+01
                          X( 2) =  0.34299999E+00
                          X( 3) =  0.16339043E+02
INEQUALITY CONSTRAINTS
                        PHI( 1) =  0.33721924E-02
                        PHI( 2) =  0.25596634E+02
                        PHI( 3) =  0.31099999E+00
                        PHI( 4) =  0.57000011E-01
                        PHI( 5) =  0.13181995E+01
                        PHI( 6) =  0.21380055E+00
                        PHI( 7) =  0.19032083E+01
                        PHI( 8) =  0.33951674E+01
                        PHI( 9) =  0.10707813E+03
                        PHI(10) =  0.72015820E+04


 OPTIMUM SPRING DIMENSIONS FOR SUBJECT GD
*******************************************************


THE SUBJECTS WEIGHT IS.........160.0  LBS

THE TOTAL UNIT LENGTH IS....... 14.00  IN.


THE MEAN COIL DIAMETER IS......1.68  IN.

THE WIRE DIAMETER IS...........0.343  IN.

THE NUMBER OF ACTIVE COILS IS..16.25


RESULTING SPRING CHARACTERISTICS
***********************************


THE OPTIMUM SPRING RATE IS.....257.40  LBS/IN

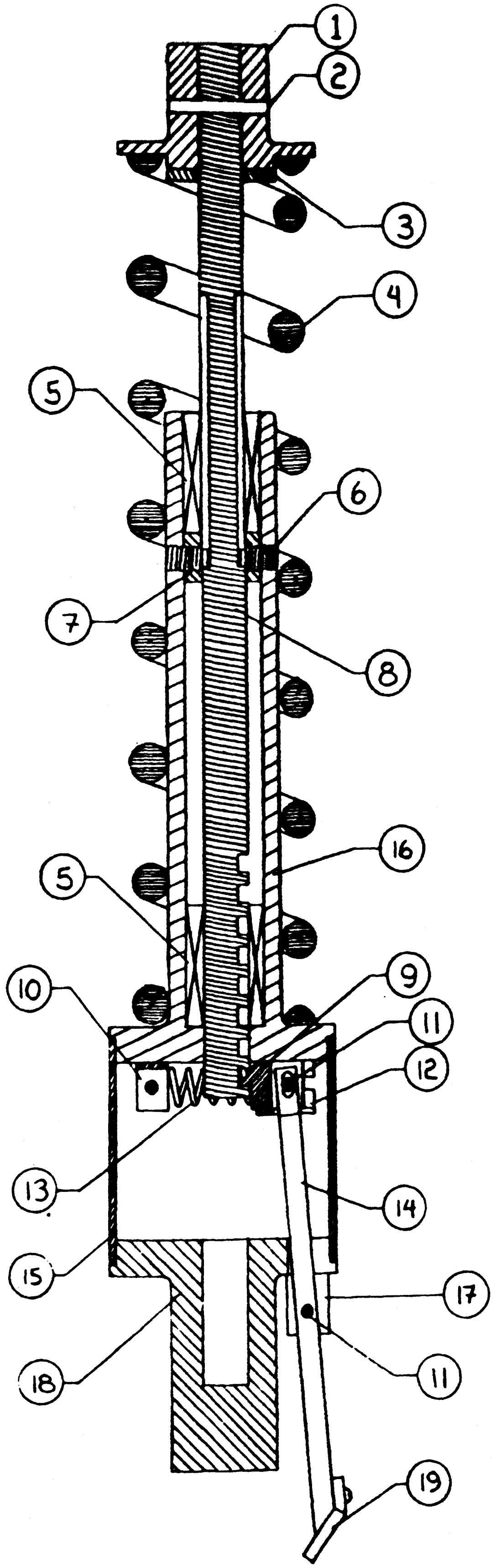
THE WEIGHT OF THE SPRING IS....2.54  LBS

THE VOLUME OF THE SPRING IS.... 8.95  IN.**3

THE SPRINGS NATURAL FREQ. IS...28.55  CYC/SEC

THE SPRINGS FREE LENGTH IS..... 9.06  IN.
```

Figure 7.7 : Final spring dimensions



ITEM	PARTS LIST
1	UPPER SPRING SEAT
2	SHAFT/SEAT PIN
3	RUBBER WASHER
4	MAIN SPRING
5	LINEAR BALL BUSHING
6	GUIDE PINS
7	BEARING SEAT/GUIDE PIN COLLAR
8	RATCHET SHAFT
9	RATCHET TOOTH
10	SPRING ADJUSTMENT PIN AND BRACKET
11	AXIS PIN
12	RATCHET TOOTH BRACKET
13	RATCHET SPRING
14	RATCHET RELEASE ARM
15	RATCHET HOUSING WALL
16	LOWER SPRING SEAT/SHAFT GUIDE
17	RELEASE ARM BRACKET
18	LOWER HOUSING SEAT
19	CONTACT PLATE

McMaster University
Department of Mechanical Engineering

PROSTHESIS FOR A/K
AMPUTEE RUNNERS

DRAWN BY: B. W. IRWIN WITNESS.
DATE: JULY 26, 1988
SCALE: FULL
© B. W. IRWIN HAMILTON, ONTARIO JULY 25, 1988

Figure 7.8 : Design of above-knee running prosthesis
(reduced to 53% of original)

such a way as to position the spring, by far the heaviest component in the unit, as close to the knee as possible to optimize the inertial characteristics of the prosthesis as a whole. The mounting provided for the spring in the unit's design was slightly shorter than the spring's actual free length to allow for inaccuracies in the spring's production, and to provide a slight pre-compression of the spring upon the unit's assembly.

Whenever possible, unit components were designed to be manufactured from 6061-T6 high strength aluminum for its superior strength and obvious weight advantages. The manufacture of such a unit with materials with poorer strength to weight ratios would be impractical in a device where the minimization of weight is of such priority. A disadvantage of using the aluminum for design components, however, is that all components must be machined from solid aluminum stock since the welding of the aluminum decreases its strength by a factor of two-thirds. Therefore all aluminum components were designed such that they could be manufactured from commercial stock sizes.

The upper spring seat and lower ratchet housing seat were also designed so as to be compatible with commercially available modular prosthetic knees and feet.

The lower spring seat incorporates two linear ball bushings to guide and support the one half inch ratchet shaft and allow for a smooth, linear compression of the unit with no lateral instability.

The ball bushings are normally intended for use with a Rockwell 60 C case hardened shaft, however it would be quite difficult to machine the necessary ratchet teeth into such a shaft and thus a one half inch diameter tool steel shaft was substituted for testing purposes.

The ratchet and release portions of the unit assumed many intermediate forms before the evolution of the final unit configuration.

An early ratchet design consisted of a tube with a serrated inner wall and the end of the shaft mounted with two small rotating gears which mated with the wall serrations but whose rotation could be locked by a latch at the end of the unit's stroke.

The ratchet was then modified to consist of a double-toothed shaft with two spring-tensioned, rotationally disengaging ratchet teeth.

The final configuration consists of the single toothed shaft and linearly-disengaging, spring-tensioned ratchet tooth shown in Figure 7.8. The design allows for the adjustment of the ratchet spring length to vary the necessary ratchet disengagement force. The ratchet tooth was also manufactured from tool steel, and both the tooth and ratchet shaft were heat treated to increase the surface hardness and reduce wear.

The method of disengaging the ratchet tooth was also considered in several configurations.

It was first necessary to specify which aspect of the prosthesis, and which aspect of its spacial orientation, would determine the ratchet's release. The three major aspects considered were the shank's angular position relative to the ground, the degree of dorsi flexion of the prosthetic foot during late stance, or, if incorporating a single axis ankle, the degree of rotation of the ankle axis.

Each of the methods considered posed general design difficulties in terms of functional consistency and accuracy in the point of release, as well as other difficulties specific to each method.

It was felt that the rotating ankle would introduce difficulties in the prosthesis' roll-over characteristics such that during roll-over, the foot, which would normally flex under the forces being generated, would remain passive, with only the ankle in rotation. It was also felt that a rotating ankle would introduce an undesirable perception of instability for the amputee during running.

A device which would measure the angular position of the shank relative to the ground might possibly be the most accurate and consistent method of determining the ratchet release, however such a device would necessitate that the amputee run only on smooth, hard, flat surfaces, and such limitations would preclude a majority of recreational activities which involve running. Also, such a device would most likely operate external to the amputee's prosthetic foot and shoe, and thus appear awkward and aesthetically undesirable.

In the unit's final configuration, the point of release of the ratchet is determined by the dorsi flexion of the prosthetic foot during late stance. Initially, the release was to be initiated by a series of cables running below and posterior to the foot, however the design was discarded in favour of the two bar lever release system illustrated in Figure 7.9.

The two bars are not rigidly attached at the contact point for two reasons. First, if the bars were rigidly attached, the natural plantar flexion of the prosthetic foot at heel-strike would be resisted. Second, by adjusting the length of the rod and/or the contact screw, the degree of foot flexion necessary to result in the release of the ratchet can easily be varied.

A stress analysis was performed for all critical areas of the final unit design, based on loading data obtained from the gait analysis of the Terry Fox Jogging Prosthesis [7]. The analysis indicated a maximum horizontal shear of one hundred pounds and a maximum torque about the shank axis of two hundred and fifty inch-pounds (based on a five inch moment arm), in addition to the previously discussed axial impact load of approximately six hundred and forty pounds. It must be recognized, however, that although

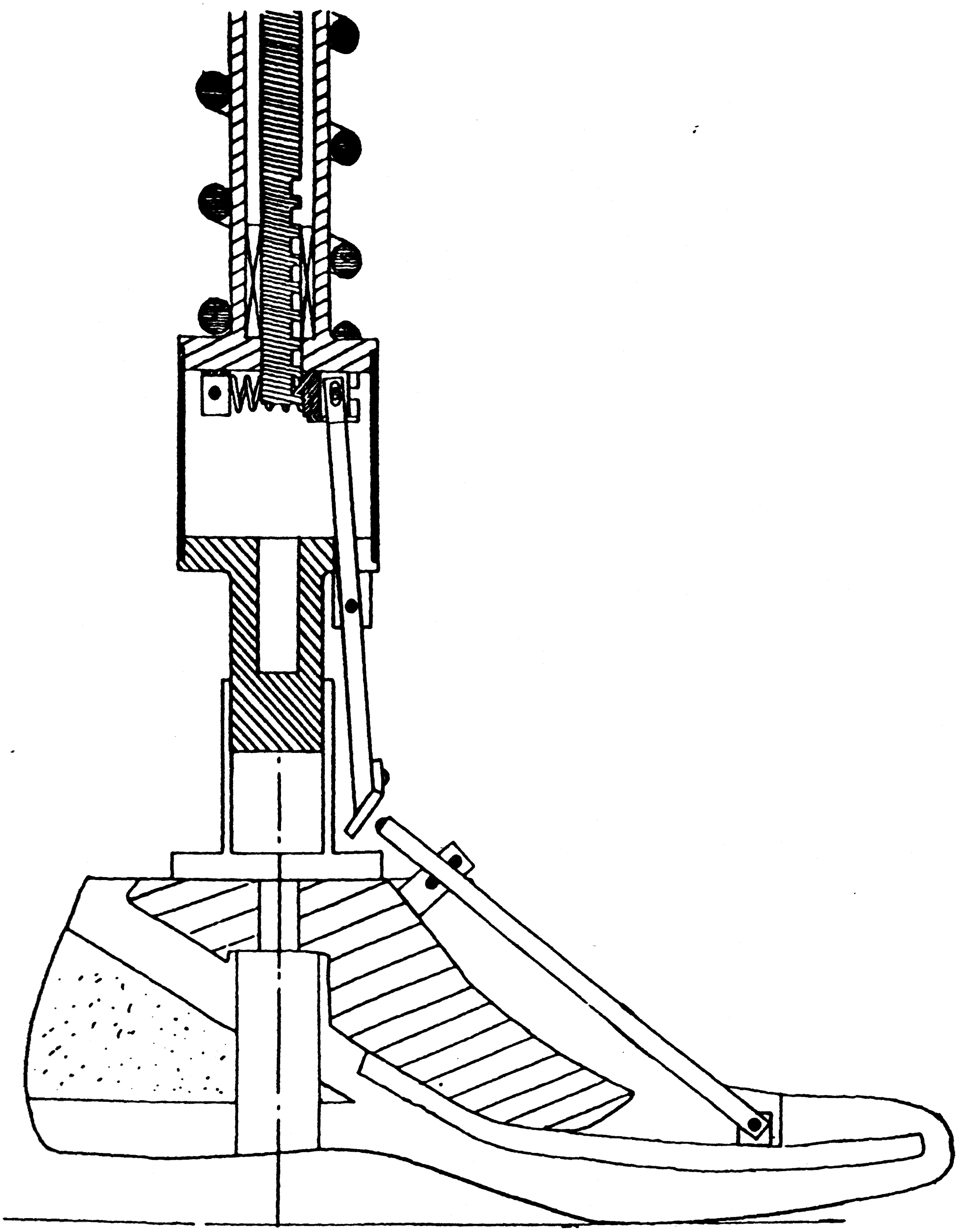


Figure 7.9 : Foot-ratchet release assembly

all the loads act through the foot, the transmission of the loading through the unit is such that the majority of components experience only one or two of the three loadings, and often not in combination.

The stress analysis indicated a reasonable factor of safety on all components for the purposes of testing and limited use.

7.5 Prosthesis Configuration

The final configuration of the total running prosthesis consists of the unit as illustrated in Figure 7.8 coupled with a SACH foot, modified as shown in Figure 7.9, and a quadrilateral, open-ended suction socket with silesian belt. The quadrilateral socket was chosen for use due to its ready availability and its successful use in the testing of the Terry Fox Jogging Prosthesis. Due to the modular nature of the prosthesis, however, either the ISNY or CAT-CAM socket could be easily substituted at a later time.

The SACH foot was chosen for use both for its desirable flexion characteristics in late stance and the ease of modifying the foot for incorporation of the necessary ratchet release hardware. Also, the use of conservative prosthetic feet such as the Seattle foot was avoided for the purposes of initial testing to allow for the isolation of the conservative effects of the operation of the shank unit. It is intended, however, that the Seattle foot or similar prosthetic foot eventually be incorporated into the prosthesis following the initial testing.

7.6 Prosthesis Operation

At the instance of heel-strike, the prosthetic knee is locked in the fully extended position and the full impact is absorbed by the linear compression of the shank unit. During this compression, the ratchet tooth is linearly displaced by the toothed shaft during its down-stroke, with this displacement being resisted by the ratchet spring. As the shaft reaches its point of maximum stroke, the ratchet tooth engages the toothed portion of the shaft, locking it in the maximum stroke position.

Since the impact energies are "locked in", and not immediately released as in the Terry Fox Jogging Prosthesis, the amputee will feel no need to "vault" over the prosthesis and thus can experience a more natural positioning of the prosthesis at heel-strike.

Although the spring was designed such that it would provide a compression stroke of approximately two inches under the anticipated impact loads, the unit has been designed such that if greater impact loads are experienced, the unit can compress up to two and one half inches to allow for conservation of those energies, rather than necessitate the amputee absorbing and dissipating the excessive impact energies through his stump and upper body.

During roll-over, the prosthesis functions similar to a conventional prosthesis, although since it is locked in a compressed position it provides the amputee with the sensation of the natural hip deflection associated with running.

In late stance, the dorsi flexion of the prosthetic foot drives the release rod into the contact plate, Figure 7.9, which, through the ratchet lever, then disengages the ratchet tooth from the shaft, releasing the stored impact energies and propelling the amputee upwards and forwards.

The ratchet tooth will not re-engage the shaft during the upstroke accompanying the energy release since the dynamics and magnitudes of the forces released will force the prosthetic foot to remain in a fully flexed position during the release.

With the release of the final energies, the prosthesis will leave the ground at toe-off and enter the swing phase.

During the swing phase, the prosthesis will exhibit swing characteristics similar to those exhibited by the Terry Fox Jogging Prosthesis, enabling the amputee to position the limb for heel strike and the continuance of the running cycle.

Designed on the basis of the aforementioned gait characteristics, this prosthesis will allow an above-knee amputee to achieve an efficient, comfortable and natural feeling one-to-one running gait.

CHAPTER 8

CONCLUSIONS

Many active, above-knee amputees have expressed a desire to run and to participate in recreational activities which involve running. They have also expressed dissatisfaction with the gait adaptations that are necessitated by the use of currently available above-knee prostheses are used for running. The preceding chapters have been intended as a comprehensive summary of the design and development of a prosthesis that will allow amputee runners to achieve a more normal, one-to-one running gait.

In addition to a review and functional description of standard prosthesis hardware currently available, more recent hardware developments were examined. An overview of the current state of above-knee running prosthesis technology was also provided.

The biomechanics of non-amputee running were examined to establish ideal design goals for the function of the prosthesis. The biomechanics of non-amputee runners were contrasted with the gait patterns associated with current above-knee running techniques. A critical comparison emphasized the inadequacies of currently available prostheses to allow above-knee amputees to achieve efficient, comfortable, one-to-one running patterns.

The action of the lower limb muscles, about each of the lower limb joints during the various stages of the running cycle, (i.e. heel-strike, toe-off), were analyzed and simplified to produce workable performance criteria for the proposed prosthesis. The foremost design criteria for the prosthesis was established as functionality, rather than as an accurate reproduction of the kinematics of the intact limb during running. The goal of the design is for the amputee to "feel" as if the prosthesis is functioning similar to the intact limb during running, although to an external observer the function of the prosthesis will appear kinematically different (i.e. no knee flexion upon heel-strike).

The three performance criteria which formed the focus of the design were the absorption of the impact energies at heel-strike, normally provided by the knee musculature, the forward drive provided at toe-off by the plantar flexion of the ankle, and the swing phase positioning on the lower limb by the knee musculature.

The final prosthesis, which is conservative in nature, meets these performance criteria by utilizing a linearly compressing shank unit and ratchet mechanism which allows the storage of the impact energies for their later use to propel the amputee forward at toe-off.

The prosthesis functionally simulates the action of the intact knee at heel-strike by absorbing the impact energies through the linear compression of the shank unit, with these energies being stored in the prosthesis by means of a ratchet device which locks the shank unit in position at the point of maximum stroke. The decreased prosthesis length which results from the storage of this energy provides the amputee with a hip deflection during mid-stance similar to that of the intact limb. Also, since the impact energies are stored and not immediately released, it allows the amputee to enter the stance phase with the prosthesis in a more natural position.

The energies remain stored until late stance when the natural flexion of the prosthetic foot results in the linear release of the stored energies, propelling the amputee upwards and forwards, functionally simulating the powerful plantar flexion of the ankle in the intact limb.

The swing phase characteristics are provided by a four-bar, polycentric prosthetic knee with mechanical swing phase control, similar in function to the knee used in the Terry Fox Jogging Prosthesis, which provided adequate swing phase characteristics to allow achievement of a one-to-one running pattern by an above-knee amputee.

In conclusion, this above-knee prosthesis has been designed to functionally imitate the performance of the intact limb during running, and will allow above-knee amputees to achieve a more natural, efficient one-to-one running gait in recreational activities.

CHAPTER 9

RECOMMENDATIONS

The present configuration of the prosthesis was developed to verify the function and performance of such a conservative design utilizing linear impact absorption and adjustable, delayed release of those energies at toe-off.

A comprehensive biomechanical gait analysis similar to that performed on the Terry Fox Jogging Prosthesis should be initiated to identify any major pathological gait characteristics, if any, associated with the prosthesis and indicate any functional modifications to the design which may be desirable.

The analysis should be used to more accurately measure the impact loads associated with the use of the prosthesis, for the purposes of improving the design of the main spring and to provide more accurate data for an exhaustive stress analysis of the present design and any future modifications.

The analysis should also be used to identify the optimum positioning of the prosthesis for the release of the impact energies at toe-off. A biomechanical evaluation might also indicate a need for a more controlled release of the stored energies at toe-off.

The shank unit was designed to be modularly compatible, and thus the prosthesis can and should be tested using a number of different prosthetic components such as the ISNY or CAT-CAM sockets, the Seattle foot, or any number of prosthetic knees, to determine which components result in the optimum gait characteristics and unit performance. If no currently available commercial knee is found to perform adequately, the design and development of a specialized knee for use in this prosthesis may be justified.

Although this prosthesis was designed, in general, to minimize its weight, significant decreases in the unit's weight are still possible. The weight of all the components could be further optimized given more accurate data concerning the forces generated during stance, provided by a biomechanical gait analysis. The heaviest of the unit's components, the main steel spring, could be significantly reduced if replaced by one of carbon-graphite composite construction.

With regards to aesthetics, it is possible that the noise associated with the operation of the prosthesis may be objectionable to some amputees, and thus future modifications to the design of the prosthesis may address this issue. Also, due to the self-contained, relatively compact nature of the shank unit, the prosthesis could easily be equipped with a foam cosmetic cover.

Future modifications to the prosthesis design should also address the simplification of the unit's manufacture and assembly, as well as the minimization of the prosthesis' cost. Many aspects of the unit's assembly could be simplified with an increase in the number being manufactured. For example, a number of the more complex aluminum components which are presently machined from stock could be cast if the volume of production could justify it.

The prosthesis has been designed to meet the physical requirements and constraints of an individual amputee. Obviously, in future the design must be generalized to allow prosthetists to prescribe and manufacture it for a wide number of amputees. Therefore, future work must pursue the development of a number of standard prosthesis configurations and shank unit sizes for a range of amputee heights and weights. Possibly the spring design program could be modified to output to the prosthetist the most suitable prosthesis configuration, including shank unit specifications, for a given amputee's physical characteristics.

While in service, running prostheses should be closely monitored for wear of the components due to the harsh operating conditions. Special attention should be paid to components such as commercial knees and feet, which have not been designed for the excessive loading associated with running.

Also, it is important to remember that an amputee's successful use of any prosthesis is dependent upon adequate training with the prosthesis under professional guidance. The amputee must learn the proper operation of the prosthesis, especially when it is radically different to any prosthesis to which he/she has previously used. Such training also increases the amputee's proprioception with regards to the prosthesis, and increases the amputee's confidence.

The majority of amputees will probably require some degree of strengthening and conditioning of their stump musculature to fully realize the full range of the prosthesis' function.

Finally, although this prosthesis has been designed primarily for the purposes of running, it is possible for it to be successfully used as a comfortable walking prosthesis, due to its impact absorbing characteristics. Future work could investigate the biomechanical characteristics of this prosthesis when used for walking.

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APPENDIX A - PROGRAM "SPRING"

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      PROGRAM SPRING
      REAL K,UNIT,SM,E,KOPT,VOL,NATFR,INSIDE,FL,SOLID,SUBWT,DIFF,
     +     AVE
      INTEGER NCOIL
      CHARACTER RESP*1,SUBNAM*8
      DIMENSION X(3),PHI(10),PSI(1),XSTRT(3),RMAX(3),RMIN(3),W(43)
     +          ,STAND(17),VAL(5)
      COMMON/INPUT/K,SM,UNIT
      COMMON/SEEK/IDATA,IPRINT,NSHOT,NTEST,MAXM,F,G,TOL,ZERO,R,REDUCE
      IPRINT=-1
      IDATA=0
      DATA STAND/0.218,0.225,0.234,0.243,0.250,0.262,0.281,0.306,0.312
     +           ,0.331,0.343,0.362,0.375,0.406,0.437,0.468,0.5/
      DATA VAL/0.0,0.25,0.5,0.75,1.0/

      TYPE*,'THIS PROGRAM IS DESIGNED TO GENERATE THE DIMENSIONS OF A
     + SPRING OF MINIMUM     WEIGHT  FOR USE IN THE 'MACH 2' JOGGING
     + PROSTHESIS.  THE USER WILL BE PROMPTED TO INPUT NECESSARY DATA
     + CONCERNING THE SUBJECT BEING FITTED, AND THE PROGRAM   WILL THEN
     + RETURN THE OPTIMUM DIMENSIONS OF THE SPRING.'

      TYPE*,' '
      TYPE*,'WHAT ARE THE SUBJECTS INITIALS ? ( TWO LETTERS )'
      READ(*,'(A)')SUBNAM
      TYPE*,' '
      TYPE*,'WHAT IS THE SUBJECTS WEIGHT? (IN POUNDS)'
      ACCEPT*,SUBWT
      TYPE*,' '
      TYPE*,'WHAT IS THE TOTAL LENGTH OF THE UNIT? (IN INCHES)'
      ACCEPT*,UNIT
      TYPE*,' '
      TYPE*,'SHOULD THE RESULTS BE WRITTEN TO A DATA FILE ?  (Y/N)'
      READ(*,'(A)')RESP
      TYPE*,' '

      IF ((RESP .EQ. 'Y') .OR. (RESP .EQ. 'y')) THEN
          OPEN(6,FILE=SUBNAM//'.RES',STATUS='NEW')
          TYPE*,'THE RESULTS WILL BE STORED IN A RESULTS FILE'
          TYPE*,'IDENTIFIED BY THE SUBJECTS INITIALS'
      END IF

      Z=0.0

C     SM=SHEAR MODULUS
      SM=11.5E6

C     E=MODULUS OF ELASTICITY
      E=3.0E7

C     THE REQUIRED SPRING RATE IS 1.6 TIMES THE SUBJECT'S BODY WEIGHT
C     PER INCH

      K= SUBWT*1.6


      N=3
      NCONS=10
      NEQUS=0
      NPENAL=1

      DATA RMAX/3.0,0.4,20.0/
      DATA RMIN/1.5,0.03,0./
```

```
C       THE ARRAY XSTRT CONTAINS THE INITIAL VALUES FOR THE
C       OPTIMIZATION SEARCH

        DATA XSTRT/2.2,.25,6./

C       THIS PROGRAM UTILIZES THE OPTIMIZATION ROUTINE SEEK TO
C       DETERMINE THE OPTIMUM VALUES OF THE SPRING PARAMETERS

  7     CALL SEEK(N,NCONS,NEQUS,NPENAL,RMAX,RMIN,XSTRT,X,U,PHI,PSI,
     +  NVIOL,W)

        CALL ANSWER(U,X,PHI,PSI,N,NCONS,NEQUS)

C       OUTPUT

C       WITH THE OPTIMAL VALUES DETERMINED, THE RESULTANT SPRING
C       CHARACTERISTICS ARE OUTPUT

  8     WRITE(6,20)SUBNAM
  20    FORMAT('1',1X,'OPTIMUM SPRING DIMENSIONS FOR SUBJECT ',A2)
        WRITE(6,50)
  50    FORMAT(1H,'***************************************************',
     +  ///)

        Z=Z+1.0

        WRITE(6,21)SUBWT
  21    FORMAT(1H,'THE SUBJECTS WEIGHT IS.........',F5.1,2X,
     +  'LBS',/)

        WRITE(6,22)UNIT
  22    FORMAT(1H,'THE TOTAL UNIT LENGTH IS.......',F6.2,2X,
     +  'IN.',///)

        WRITE(6,23)X(1)
  23    FORMAT(1H,'THE MEAN COIL DIAMETER IS......',F4.2,2X,
     +  'IN.',/)

        WRITE(6,24)X(2)
  24    FORMAT(1H,'THE WIRE DIAMETER IS...........',F5.3,2X,
     +  'IN.',/)

        WRITE(6,25)X(3)
  25    FORMAT(1H,'THE NUMBER OF ACTIVE COILS IS..',F5.2,2X,///)

        WRITE(6,26)
  26    FORMAT(1H,'RESULTING SPRING CHARACTERISTICS')
        WRITE(6,27)
  27    FORMAT(1H,'**********************************',///)


C       ACTUAL SPRING CONSTANT VALUE
        KOPT=((X(2)**4.0)*SM)/(8.0*(X(1)**3.0)*X(3))
        WRITE(6,12) KOPT
  12    FORMAT(1H,'THE OPTIMUM SPRING RATE IS.....',F6.2,2X,
     +  'LBS/IN')


C       DETERMINING THE FINAL WEIGHT OF THE SPRING
        WRITE(6,13) U
  13    FORMAT(/,1H,'THE WEIGHT OF THE SPRING IS....',F4.2,2X,
     +  'LBS',/)
```

```
      VOL= U/0.284
      WRITE(6,30)VOL
 30   FORMAT(1H,'THE VOLUME OF THE SPRING IS....',F5.2,2X,
     +'IN.**3')

C     FIND THE SPRINGS NATURAL FREQUENCY
      INSIDE=(KOPT*32.2)/U
      NATFR=0.5*SQRT(INSIDE)
      WRITE(6,31)NATFR
 31   FORMAT(/,1H,'THE SPRINGS NATURAL FREQ. IS...',F5.2,2X,
     +'CYC/SEC',/)

C     THE SPRINGS FREE LENGTH
      SOLID= 1.05*(X(3)+2.0)*X(2)
      FL=(2.5*K)/KOPT + SOLID
      WRITE(6,32)FL
 32   FORMAT(1H,'THE SPRINGS FREE LENGTH IS.....',F5.2,2X,
     +'IN.',/)

      IF (X(2).LT. 0.2) X(2)=0.2
      DO 1,I=1,17
      IF (X(2) .LT. STAND(I)) GO TO 2
 1    CONTINUE
 2    TEMP=(STAND(I)+STAND(I-1))/2.0
      IF(X(2) .LT. TEMP) THEN
      X(2)=STAND(I-1)
      ELSE
      X(2)=STAND(I)
      END IF
      XSTRT(2)=X(2)
      RMIN(2)=X(2)
      RMAX(2)=X(2)
      R=1
      XSTRT(1)=X(1)
      XSTRT(3)=X(3)
      IF (Z .EQ. 1.0) THEN
      GO TO 7
      END IF

      NCOIL=X(3)
      DIFF=X(3)-NCOIL
      DO 3,J=1,5
      IF (DIFF .LT. VAL(J)) GO TO 4
 3    CONTINUE
 4    AVE=(VAL(J)+VAL(J-1))/2
      IF (DIFF .LT. AVE) THEN
      X(3)=NCOIL + VAL(J-1)
      ELSE
      X(3)=NCOIL+ VAL(J)
      END IF
      IF (Z .EQ. 2.0) THEN
      GO TO 8
      END IF

      STOP
      END


      SUBROUTINE UREAL(X,U)
      DIMENSION X(1)

      X(1)= ABS(X(1))
      X(2)= ABS(X(2))
```

```
      X(3)= ABS(X(3))
C     X(1)= MEAN COIL DIAMETER
C     X(2)= WIRE DIAMETER
C     X(3)= NUMBER OF ACTIVE COILS
      U=2.4674*X(1)*(X(2)**2.0)*(X(3)+2.0)*0.284
      RETURN
      END


      SUBROUTINE EQUAL(X,PSI,NEQUS)
      DIMENSION X(*),PSI(*)
      RETURN
      END


      SUBROUTINE CONST(X,NCONS,PHI)
      COMMON/INPUT/K,SM,UNIT
      REAL K,UNIT,MXSHR,A,EXP,KMX,KMN,KOPT,KS,TA,TM,RATE
      DIMENSION X(*),PHI(*)
      X(1)= ABS(X(1))
      X(2)= ABS(X(2))
      X(3)= ABS(X(3))
      C=X(1)/X(2)
      IF (C .LE. 1.) C=1.1
      KS=(((4.*C)-1.)/((4.*C)-4.)+(0.615/C))
C     UPPER AND LOWER ALLOWABLE SPRING RATE VALUES
      KMX=K
      KMN=K - 0.1*K
      RATE=((X(2)**4.0)*SM)/(8.0*(X(1)**3.0)*X(3))
C     SPRING RATE CONSTRAINTS
      PHI(1)=KMX - RATE
      PHI(2)= RATE - KMN
C     WIRE DIAMETER CONSTRAINTS
      PHI(3)=X(2)-0.032
      PHI(4)=0.4 - X(2)
C     MEAN COIL DIAMETER CONSTRAINTS
      PHI(5)=3.0 - X(1)
      PHI(6)= (X(1)-X(2))- 1.125
C     CONFIGURATION CONSTAINT
      PHI(7)= C - 3.0
C     MXSHR=MAXIMUM ALLOWABLE SHEAR STRESS
      A=202000.0
      EXP=-0.112
      MXSHR=(0.4328*A*(X(2)**EXP))/1.1
C     FREE LENGTH CONSTRAINT
      PHI(8)=(UNIT-2.0)-(2.0*K)/RATE  -1.05*(X(3)+2.0)*X(2)
C     SHEAR STRESS CONSTRAINT
      PHI(9)=MXSHR-(KS*20.0*K*X(1))/(3.1417*(X(2)**3.0))


C     FATIGUE CONSTRAINTS
      TA= (KS*8.0*1.125*K*X(1))/(3.1417*(X(2)**3))
      PHI(10)=47472 - TA
      RETURN
      END
```