

THE KNUD JANSEN LECTURE

Above-knee prosthetics

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Introduction

In any discussion of the development of technology as applied to the rehabilitation of the above-the-knee amputee, we must first acknowledge the many contributions of European pioneers in this field.

In Europe the needs resulting from World War I resulted in greatly accelerated research and development in the field of prosthetics technology during the next three decades. This research included scientific studies of the biomechanics of human walking, alignment principles, and methods of fitting for above-knee prostheses. Notable examples were the work of Braune and Fischer on human gait, and the publications of Dr. Schede and others on scientific principles of fitting and alignment. The list of German and Austrian contributors to prosthetic technology is long and distinguished. Their literature and professional meetings have always included a series of lively discussions between proponents of different construction systems and prosthetic component designs.

In 1946 a U.S. Army Commission under the direction of Col. L. T. Peterson made a study of European prosthetics technology. They were particularly impressed with the work of Haberman and Schneider in Germany. A major recommendation of this commission was for the introduction of the so-called "suction socket" for the above-knee amputee into American prosthetic practice. Responsibility for supervision of this programme was delegated to the University of California at Berkeley and a series of schools and clinical trials were successful in introducing this new fitting technique into the United States during the 1946-1948 period.

In 1949 a second group led by Dr. V. T. Inman and Professor H. D. Eberhart visited Europe

for another critical comparison of European and American practice, particularly in the area of lower limb prosthetic components and fitting and alignment principles. They were intrigued with the results obtained by Striede in Austria using the quadrilateral suction socket shape. They also described and catalogued a large number of physiological, linkage, and friction brake knee designs, alignment devices and other fitting aids. Among others, the condylar designs of Striede, Lang, Röck, and Haberman were studied and evaluated as well as the brake type designs of the Kleinkathöfer and Jüpa knees and the four-bar linkage design of Lammers. In addition, they observed the alignment principles and adjustable "walking machine" prosthesis of Schneider and other alignment systems such as those used by Haberman in Frankfurt, and others.

Upon the return of the Inman-Eberhart group, a series of clinical trials of devices and principles observed in Europe were undertaken at Berkeley in an effort to evaluate their potential for improvement of the comfort and function of the suction socket above-knee prosthesis as it was then being fitted in the United States. An extensive review of the German literature revealed a wealth of information but no universally recognized and accepted set of biomechanical principles which could guide the prosthetist in selecting components and methods for fitting the above-knee prosthesis. A number of different socket shapes were in use in Europe as well as considerable variety in the devices, materials and methods of fitting and alignment.

As a result of evaluations subsequent to the 1949 European trip, the following items were selected for more extensive study at Berkeley.

- (1) The quadrilateral ischial bearing socket shape as fitted by Striede in Austria.
- (2) The benefits of polycentric knee constructions; in particular, the Striede, Röck, and Lang knees from Germany.
- (3) The use of adjustable prostheses for walking

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trials and dynamic alignment adjustments as used by Schneider in Germany.

- (4) Biomechanical principles of fitting and alignment of the suction socket above-knee prosthesis.

In addition to the above, the Inman-Eberhart report noted the generally poor swing phase control in both American and European prostheses and proposed an additional project.

- (5) The development of improved swing control devices which would simulate quadriceps and hamstring muscle action about a normal knee and allow walking over an increased range of walking speeds.

During the period 1949-55, the research staff at the Biomechanics Laboratory continued their evaluation of the German technology and developed what was considered a rational biomechanical basis for fitting the carved willow wood suction socket. An alignment system using the "AK adjustable leg" and the "AK alignment duplication jig" was developed and tested with the co-operation of prominent American prosthetists, notably Trautman in Minneapolis, Haddon in Denver, and Thranhardt in Atlanta.

This information was organized in the form of a syllabus for the first "pilot school in above-knee prosthetics" held in August 1955 for the purpose of training future instructors for the new prosthetics education programmes then being developed at New York University and the University of California at Los Angeles. This material later became the basis for the *Manual of Above-Knee Prosthetics* edited by M. H. Anderson and R. E. Sollars and published January 1, 1957.

One of the major benefits of the 1955 technology was the introduction of a uniform and scientific terminology which enhanced the ability of members of the prosthetic clinic teams to communicate with each other in solving the various problems associated with amputee rehabilitation. The new education programmes in above-knee prosthetics were one step toward the eventual certification and improved professional education of American prosthetists.

The period 1955-60 saw the introduction of thermosetting laminating resins into lower limb prosthetics technology: first as a substitute for rawhide in the reinforcement of wooden prostheses, later as the basic material for plastic sockets moulded over plaster models of the stump. The remarkable success of the laminated

plastic socket used with the patellar-tendon-bearing below-knee prosthesis led to the use of similar techniques with above-knee sockets. The adjustable brims developed at Berkeley and the fitting stand developed by VAPC in New York City, both of which appeared in the early sixties, were designed to aid in the plaster wrap casting of the above-knee stump. These new tools made the "total contact" AK suction socket possible and helped to solve the problems of distal stump oedema often experienced with "open-end" carved willow wood sockets.

Studies of the kinematics of various European polycentric, physiological, and brake-type knee constructions indicated a general confusion as to the function, amputee benefits, and prescription criteria for such devices.

It was clear that a comprehensive theory which could relate the functional characteristics of such devices to their alignment and the needs of the amputee was required. It was hoped that studies of this type would show the relationship between the physical capabilities of the amputee, the functional characteristics of the device, and the manner in which the prosthesis was used in walking. It became clear that it was instructive to discuss above-knee prosthetics in a somewhat different manner than had been typical in the past. It was typical in the pre-1950 era to discuss fitting and alignment of the above-knee prosthesis as almost separate problems in three areas:

- (1) socket shape and fitting methods
- (2) alignment for stability and swing through
- (3) selection of components.

Eventually, a somewhat different and hopefully more logical approach evolved as given below. We will first discuss the functional characteristics of certain classes of prosthetic knee designs and relate their function to alignment principles. The use of a certain class of knee mechanism plus the need to balance the body weight above the prosthesis during the stance phase creates a varying pattern of biomechanical pressures between stump and socket. These pressure patterns can be anticipated and influence the basic socket design. The most difficult part of the entire process is the fitting of the socket to the individual anatomical characteristics of the amputee. Here the skillful prosthetist will "shape" the socket to achieve a comfortable accommodation of the required biomechanical forces and pressures. In the actual fitting, the casting of the stump and shaping and

adjustment of the socket are often the first steps in the procedure. However, the successful fitting also must include careful planning of the complete procedure including the influence of components and alignment on the type of socket fitting to be achieved.

Classification of prosthetic knee mechanisms

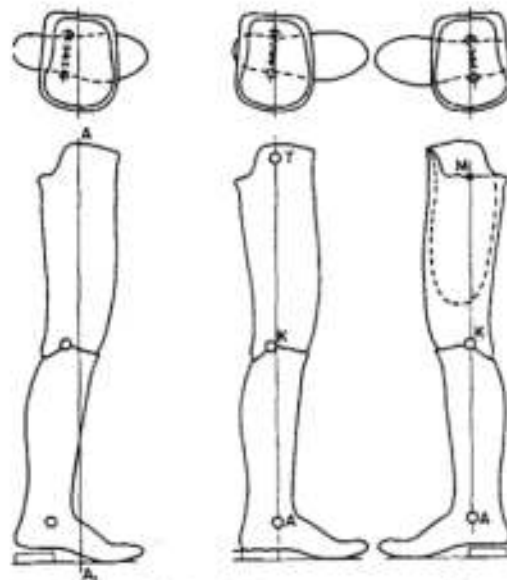
A large number of different knee mechanisms have either been invented or developed as research studies, and many are available commercially for fitting to the above-knee amputee. The potential user often has great difficulty in evaluating the claims of the developer as to the benefits to be derived from the special features of a particular device. When a particular mechanism is considered in terms of its basic functional elements it is often possible to catalogue the device as a member of a certain class. There will, of course, be differences in construction, weight, durability, and even the degree to which a particular device approaches the optimal functional behaviour expected of a device of that class. In this discussion we will classify knee mechanisms in terms of (1) their special features which are related to enhancing knee stability, i.e., safety against collapse of the knee during the stance phase, and (2) the type of mechanism employed to control the flexion of the knee during the swing phase of walking.

Bench alignment systems

The basic means employed in many above-knee prostheses to achieve knee stability is to align the knee axis in a location such that the load carried by the weight-bearing prostheses always passes ahead of the knee axis and forces the prosthetic knee against a mechanical stop in the fully extended position. This simple principle is employed in all above-knee prostheses regardless of other means used to achieve stability and will be referred to as *alignment stability* in the following discussion.

Alignment stability is often ensured by a process which has become known as *bench alignment*. Two somewhat different philosophies for bench alignment have been proposed. In Germany the prosthesis is often assembled in a fixture using a plumb line viewed from the lateral side which passes downward from the centre of the socket brim through a point on the bisector of the length of the foot as shown

in Figure 1. The trochanter is sometimes used as the upper reference point. To achieve what may be described as a rolling action of the hip over the foot the heel of the foot is elevated slightly in the reference position. The bench alignment diagram displays the prosthesis in a position corresponding to the point where the amputee would begin to roll over the ball of the foot in walking, i.e., in a position where the hip joint is close to the highest point in its trajectory. The clearance under the heel is often called a "safety factor". An increased clearance results in a more rapid transfer of weight to the ball of the foot and improves knee stability at heel contact.



Figs. 1, 2 and 3. Bench alignment systems. Fig. 1 (left) German, fig. 2 (centre) TKA reference, fig. 3 (right) MKA reference.

The position of the knee axis relative to the vertical reference line is dictated by the combination of knee kinematics and friction brake action inherent in a particular mechanism and very definite rules have been established for different combinations of knee and foot designs.

In the United States a similar plumb line reference system has been used which has the trochanter as the upper reference point and which locates the ankle joint directly under the trochanter as shown in Figure 2. Rules have been established for locating the knee joint on or behind this reference line which has been called the TKA line. In the American system the

heel is shown in contact with the floor in the reference position. This system has evolved because of the need to have a convenient way to check bench alignment prior to walking trials with adjustable devices without the need for special holding or limb assembly fixtures as are common to some European systems.

The American system may result in a prosthesis which is somewhat longer than a similar prosthesis if the pelvis level is checked with the limb in the alignment reference position.

As the American prosthesis rotates forward over the ball of the foot toward the German reference position, it is clear that the heel would rise from the floor and the hip joint would be elevated slightly. This explains in part the controversy which often occurs between the American prosthetist who prefers to construct the prosthesis shortened a few millimetres to make it easier for the amputee to walk and the physician who insists on a level pelvis with the prosthesis in the alignment reference position.

Errors in both systems can result when the trochanter is used as the upper reference point. It requires that the alignment be checked with the amputee wearing the prosthesis and precise location of the point of contact of the head of the trochanter against the lateral brim of the socket is often difficult.

The reference line shown in Figure 3 is suggested as an alternative which is particularly well suited to the fitting of the quadrilateral socket shape. The upper reference point is established at the bisector of the interior medial wall of the socket. In the modified American system the new reference line becomes the MKA line and the knee joint is located on the medial aspect of the prosthesis rather than the lateral side as with the use of the TKA line.

When using the TKA line, the knee axis is located on the lateral side of the prosthesis and for a typical prosthesis the knee axis is located approximately 6 mm ($\frac{1}{4}$ ") posterior to the TKA line.

The knee axis is typically aligned in 5 degrees external rotation, as necessary to prevent lateral movement (whip) of the foot in the swing phase. The medial end of the knee axis is forward of the lateral end by approximately 6 mm. Therefore, when using the MKA reference line along the medial aspect of the prosthesis, the knee axis can be located directly on the MKA line for a typical active above-knee amputee with medium to long

stump.

The usefulness of the German air space under the heel is also acknowledged and where greater knee security is desired, the use of increased heel rise in the prosthetic foot is encouraged.

A spacer under the heel 6 to 10 mm thick at the time of bench alignment will give the amputee greatly increased knee security at heel contact. The prosthesis will not have its length increased with this procedure since the distance to the ball of the foot remains the same.

Polycentric knee mechanisms

Alignment stability can be enhanced through the use of polycentric knee mechanisms. A polycentric knee mechanism is any device where the instantaneous centre of rotation of the knee changes its position as the knee flexion angle increases or decreases. Kinematically all polycentric devices, where stability is controlled by the location of the instantaneous centre of rotation (the instant centre), are of the same class. However, there is an infinite variety of kinematic arrangements possible, each with its own special functional characteristics. The following discussion is an attempt to explain the behaviour of such devices assuming, in the beginning, that there is no brake-type friction acting simultaneously as part of the function of a particular device.

There are several ways that polycentric motion between the thigh piece and shank can be created in a knee mechanism.

Figure 4 displays schematically most of the polycentric prosthetic knee mechanisms that have been used to date. In each diagram the point I represents the instantaneous centre of rotation of the thigh piece or knee block with respect to the fixed shank with the knee in full extension. In many cases, it has been possible to locate the point I in exactly the same position when the knee is fully extended.

Case A represents the double-skid mechanism often used for so-called "physiological" knee mechanisms. In this case the shape of the upper condylar surface determines the manner in which the instant centre changes its location with knee flexion. Point I is located at the intersection of the two lines which are perpendicular to the surfaces in contact at the two skid points.

Case B is a simple variation where one skid point has been replaced by a pivoted slider.

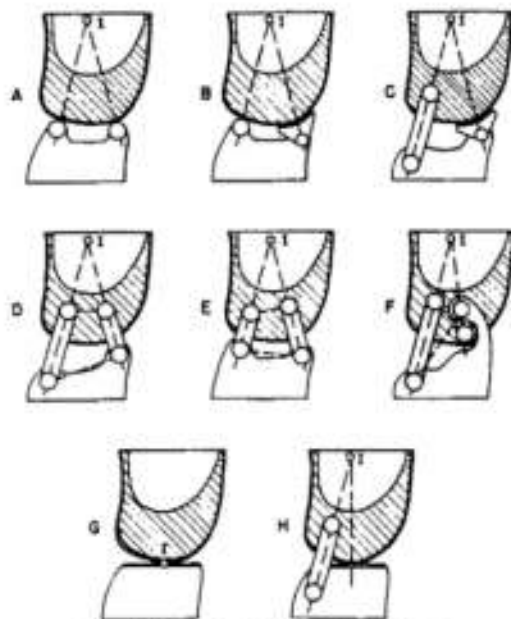


Fig. 4. Polycentric knee mechanisms.

Point I is again located by two lines, one of which passes through the slider pivot perpendicular to the upper condylar surface.

In case C the rear skid point has been replaced by a two-joint link and the centre line of the posterior link becomes one of the lines required to locate point I.

Cases D and E are four-bar linkages with the same instant centre I but obviously different flexion characteristics. The knee block and shank structure form the second pair of "bars" as shown by dotted lines.

Case F is also a four-bar linkage. The upper pivot of the anterior two-joint link is attached to the shank in this case.

Case G represents a design involving pure rolling rather than a skidding action. In devices of this type, the instant centre I is always at the point of rolling contact.

Case H is also somewhat different. The posterior two-joint link acts in the same manner as in the four-bar linkage. The anterior guidance is provided by the skidding of the surface of knee block along a flat surface attached to the shank. Since the shank surface is flat and horizontal, the line which locates I is always perpendicular to the shank surface at the point of contact.

Brake knee mechanisms

The second basic way of enhancing knee

stability is through the incorporation of devices which resist the tendency of the knee to flex or collapse under load by the creation of a braking action which either locks or resists a tendency of the knee joint to flex during the stance phase. There have been many designs proposed over the years which provide for braking action of this type. Figure 5 illustrates a few types which have been produced commercially. In Figure 5A the braking effect is due to friction developed between mating cylindrical surfaces on the knee block and shank. The surfaces are separated by a spring which is compressed as load is transferred to the prosthesis and allows the friction surfaces to contact. The magnitude of the friction is influenced by the surface materials, spring stiffness, and other geometrical parameters. It should be noted that friction can also be present in the mechanisms shown in Figures 4A, 4B, 4C and 4H and often is a major functional feature in such arrangements.

Figure 5B illustrates a friction clamp which is actuated by a posterior rotation of the knee block about point P. The resulting pressure on

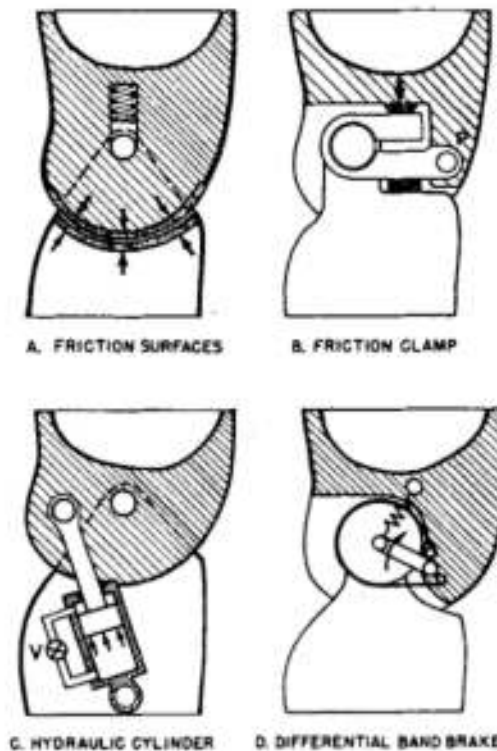


Fig. 5. Brake type knee mechanisms.

the clamp creates friction on the knee bolt which is rigidly attached to the shank.

Figure 5C shows a hydraulic cylinder with a passage for fluid to flow between upper and lower chambers. The valve V can be closed by a variety of mechanisms not shown which prevents the knee from flexing under load. A controlled rate of flexion under load is possible if a small amount of fluid is allowed to flow through the closed valve.

Figure 5D shows a friction brake of the differential band brake type. Load on the prosthesis causes the pivoted arm to rotate upward against the compression spring. This relative motion increases the tension in the band wrapped around a cylinder attached to the shank. The sensitivity of the device in generating a friction brake moment is determined by the setting of the spring.

Voluntary control of knee stability

The brake knee mechanisms shown in Figure 5 provide what might be called automatic control of knee stability. When operating properly, the brake friction is created automatically each time the prosthesis assumes its weight bearing function.

The brake action is generally beneficial in level walking, particularly on rough surfaces. In some cases there may be difficulty in flexing the knee at the end of stance phase when walking on slopes and stairs.

Biomechanics of polycentric knee mechanisms*

A linkage knee offers improved function for the amputee because it is *polycentric*. As the knee flexes, the apparent or instantaneous centre of rotation between the thigh and shank changes in a manner which improves the stability characteristics of the knee and also allows an increased range of knee flexion. Swing control devices can be added to the basic mechanism.

The characteristics of any prosthetic knee are best described in terms of the relationship between four factors: (1) the instant centre of thigh-shank rotation, (2) the load line, (3) the brake moment or torque generated by the prosthetic knee, and (4) the hip moment which can be supplied voluntarily by the amputee.

Figure 6 illustrates a simple but important principle of mechanics. The combination of a force P plus moment M as shown in Figure 6a

is equivalent to the single force Q offset a distance d as shown in Figure 6b. We see that Q is equal to P and that the distance d is calculated by noting that $Qd = M$. Hence

$$d = \frac{M}{Q} = \frac{M}{P} \quad (1)$$

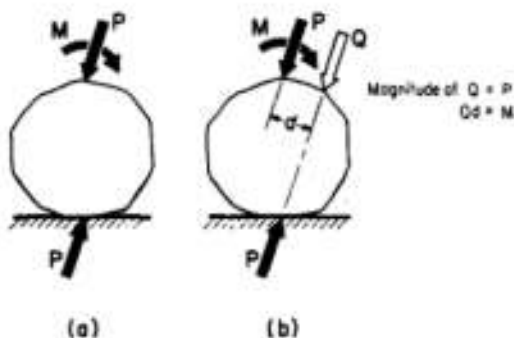


Fig. 6. Single force Q equivalent to combined force P and moment M .

Using this principle we see that the position of the load line at heel contact or at toe off may be related to the load P carried through the hip joint and the hip moment M_H exerted by the amputee, either in flexion or extension. An extension moment moves the load line ahead of the hip, a flexion moment moves it behind. Figure 7 shows a hypothetical situation where in Figure 7a the amputee is exerting a hip extension moment to maintain knee stability at heel contact. Figure 7c shows the situation where the amputee is exerting a hip flexion moment in an attempt to flex the prosthetic knee while the prosthesis is continuing to support the body weight. These two diagrams are combined in Figure 7b and drawn relative to a fixed reference position of the limb.

In Figure 7a we note that the prosthetic knee will be stable at the critical instant of heel contact whenever the knee centre is aligned behind the line PQ . In Figure 7c it is apparent that the knee can be flexed voluntarily under load if the knee centre is located ahead of the line P^1Q^1 . In Figure 7b the two zones E and F when superimposed define a common region S, the *zone of voluntary stability*. For the hypothetical amputee of Figure 7 the zone S would

*Excerpted from "Polycentric Linkages as Prosthetic Knee Mechanisms for the Through-Knee Amputee", ISPO, Montreux, October 1974.

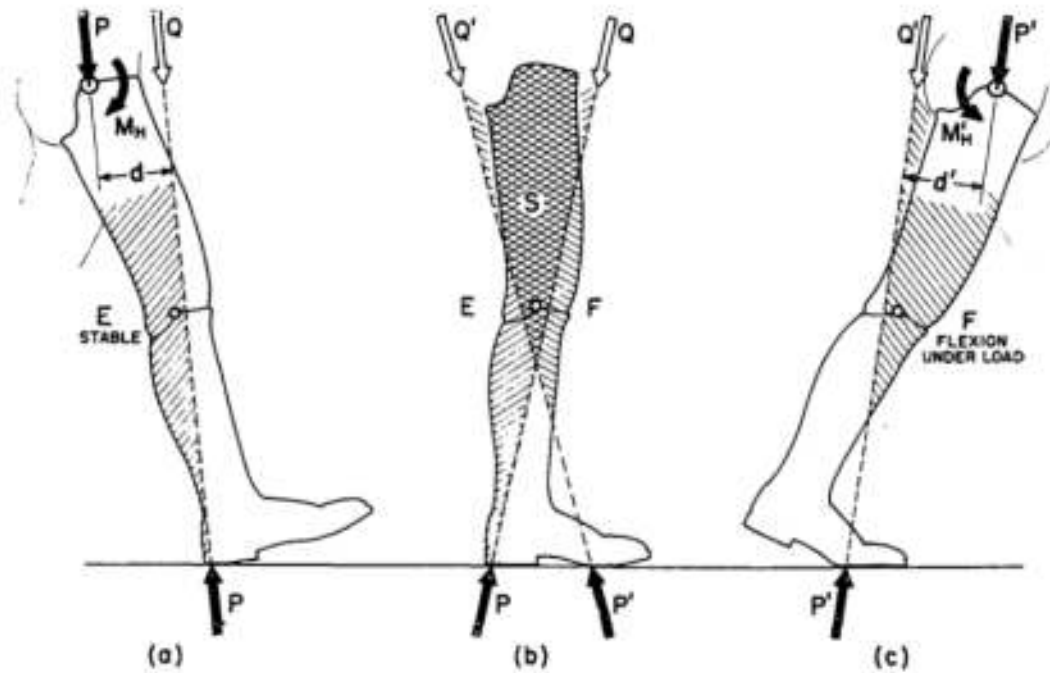


Fig. 7. The zone of voluntary stability, S.

allow considerable variation in the alignment of the prosthetic knee joint while maintaining stability at heel contact and ease of knee flexion during push-off.

Figure 8 illustrates a more typical situation where the amputee either has reduced hip moment capabilities or prefers to use his hip musculature at less than maximum strength under ordinary conditions. In this case the zone of voluntary stability is dramatically reduced. Under these conditions the prosthetist fitting a single axis prosthetic knee joint must align the knee joint with its centre behind line PQ in order to ensure knee stability at heel contact. Such a location is also well behind line P'Q' and outside of the desirable region F. The result is a stable knee at heel contact but a knee which is difficult to flex under load at push-off. Voluntary knee flexion is important in achieving an aesthetic and energy saving gait.

As Figures 7b and 8 indicate, the ability of the amputee to control knee stability is influenced by the height of the instant centre of knee rotation above the floor. A high knee centre provides improved leverage for voluntary control of knee stability. Single axis knees provide little or no opportunity to make practical use of this fact because any significant change in the

vertical position of the knee joint is cosmetically unacceptable. The polycentric knee can be designed with the initial instantaneous centre of rotation located above the usual knee joint and well within the zone of voluntary stability, S. The prosthetist then has the option of emphasizing either stability at heel contact or ease of knee flexion by appropriate alignment of the knee mechanism within the prosthesis.

The interrelation of all factors, including the effect of a friction brake moment M_K developed by the knee joint, is given by the knee stability equation

$$M_H = \frac{L}{h} (Pd - M_K) \quad (2)$$

where, as shown in Figure 9,

M_H = the muscle moment about the hip joint on the amputated side required to maintain the weight-bearing knee in a stable flexed position.

M_K = the knee moment created by mechanical or hydraulic friction in a "brake" type design ($M_K = 0$ in the UC-BL four-bar knee).

L = the total length of the prosthesis from the hip to the bottom of the heel.

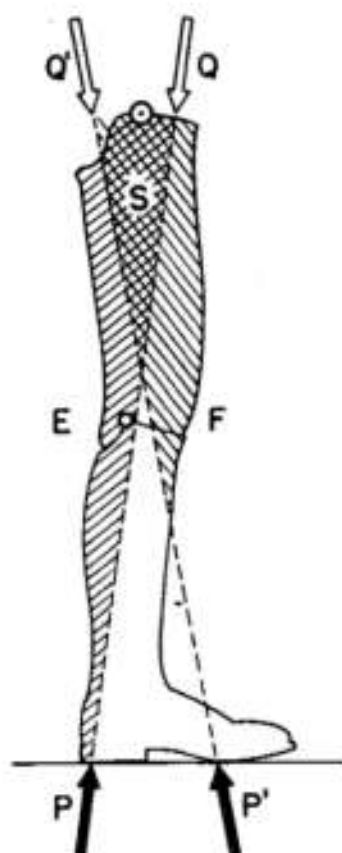


Fig. 8. Typical stability diagram for an above-knee amputee.

- P = the load carried along the long axis of the leg.
- h = the vertical height of the instantaneous centre of knee rotation measured from the bottom of the heel.
- d = the distance forward from the hip/heel line to the instantaneous centre of knee rotation.

Knee stability is influenced by three independently variable parameters: M_K , d , and h in equation (2). For a given kinematic arrangement it is obvious that the required M_H can be reduced to zero if the mechanism is capable of developing a brake moment:

$$M_K = Pd \quad (3)$$

This principle has been exploited successfully in a wide variety of *friction-stabilized* knee brake mechanisms. It should be recognized that the friction which resists flexion to provide knee stability at the beginning of stance phase will interfere with knee flexion near the end of the

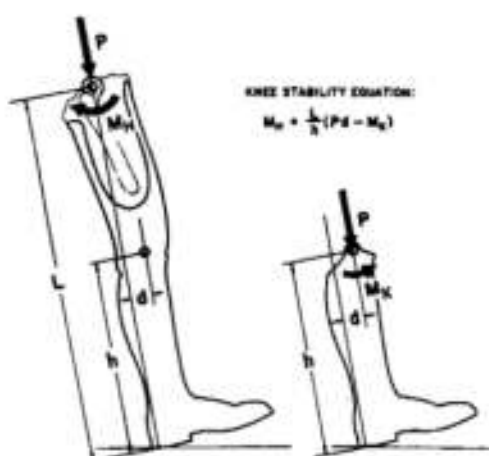


Fig. 9. Dimensions used in the knee stability equation.

stance phase. As the amputee attempts to flex the knee, the negative sign in equation (2) becomes positive because the friction moment is acting in the opposite direction, and the hip moment required to initiate knee flexion is *increased* as compared with a *non-brake* mechanism.

The parameter d in equation (2) governs the so-called *alignment stability*.

It is possible to reduce the variable d to zero by aligning the prosthesis in such a manner that the knee centre lies on the hip/heel line. With such an alignment, a muscle moment about the hip would not be required to maintain knee stability at heel contact. Unfortunately, however, such an alignment is not satisfactory because it provides too much stability later in the stance phase when weight is borne on the forefoot, thereby interfering with desirable flexing of the knee just prior to toe off.

The third parameter h in equation (2), in combination with dimension d , provides what may be described as *kinematic stability*, as discussed below.

The kinematic characteristics of a polycentric linkage are determined by the length and arrangement of its links. An infinite variety of arrangements is possible.

One measure of the functional characteristics attained with a particular linkage is a plot of the successive locations of the relative centre of rotation of the socket with respect to the shank as a function of the knee angle, as shown in Figure 10 for the University of California four-

bar linkage above-knee prosthesis. A larger value for h and/or a smaller value for d at a particular angle of knee flexion will result in a smaller hip moment required to maintain knee security. However, during knee flexion a large h also results in a cosmetically undesirable forward translation of the knee block in the region of the anatomic knee centre. The knee centre must eventually move downward with knee flexion or the knee will not bend in the proper place during sitting. It is desirable to maintain a large value

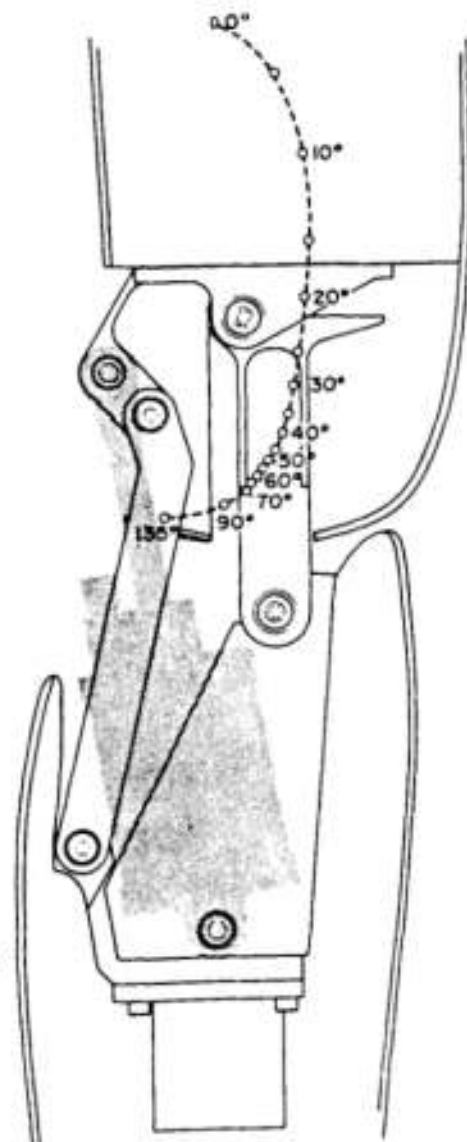


Fig. 10. Linkage arrangement and path of the instant centre for the UC-BL four-bar linkage polycentric knee.

of h and thereby keep the knee centre high over the first 10 degrees of knee flexion in order to help the amputee recover from a stumble or other situation when knee stability is accidentally disturbed. The linkage shown in Figure 10 has been selected as a compromise which tends to maximize the desired functional benefits of the linkage while minimizing the cosmetic problems due to translation of the knee block. The kinematics of the UC-BL four-bar linkage has been found to be useful to a wide variety of young male amputees, including one bilateral above-knee amputee, over periods of use ranging up to 25 years.

Figure 11 indicates the manner in which the rotation centre (plotted on the moving socket) changes position during a full walking cycle. Note that it is possible to have a relatively large dimension d in full extension, which is beneficial to the amputee near the end of the stance phase. The elevated position of the centre of knee rotation compensates for the large value of d by allowing the amputee to make better use of available muscle moments at the hip, and therefore provides improved *voluntary control* of knee stability. It is interesting to note that voluntary control is equally important in both flexion and extension, particularly to the younger, more active amputee.

Swing phase control

A swing control is a device which can be incorporated into the design of a prosthetic knee mechanism such that it simulates the action of the quadriceps and hamstring musculature which act about the normal knee joint during the swing phase of walking. Such a device has three functions: (1) to limit the maximum knee flexion angle and cause the shank-foot to swing forward smoothly in a manner similar to quadriceps action about a normal knee, (2) allow the knee to extend smoothly into full extension without impact, (3) provide automatic changes in the level of the resistance patterns to allow walking over an extended range of walking speeds.

One means of establishing design criteria for swing control mechanisms is to calculate the knee moment pattern which must be generated by the device in order to cause the shank-foot to swing through space with a motion pattern which approximates that of an average normal person. Figure 12 illustrates qualitatively the

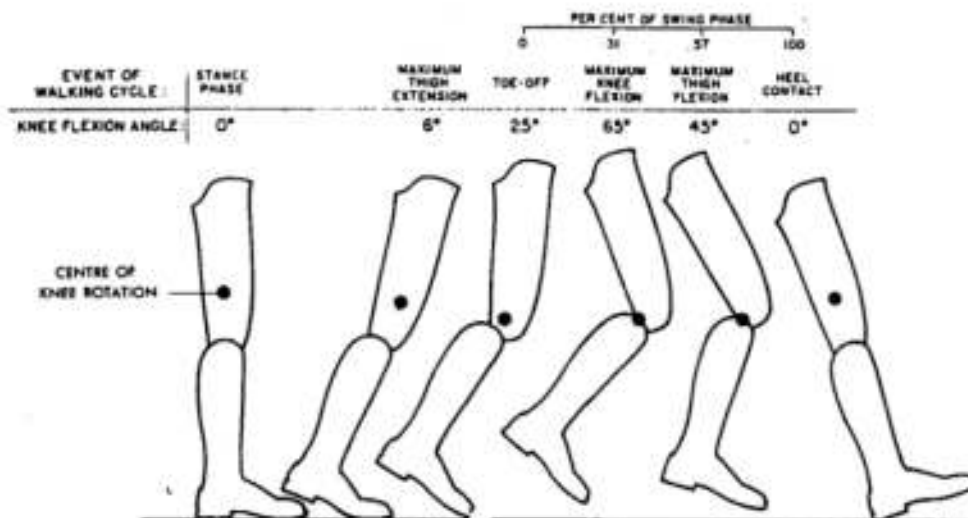


Fig. 11. Change in knee centre during walking.

results of such calculations. It is seen that the device should provide a resistance pattern with four phases.

In normal walking the knee begins to flex before the foot loses contact with the floor. The rate of flexion is controlled by the quadriceps group acting over the anterior aspect of the knee joint. Such action at present is difficult to duplicate in prosthetic knee mechanisms although the gliding action of certain polycentric knee designs is an approximation.

After the foot leaves the ground the quadriceps continue to act to limit the knee flexion to a maximum of approximately 65 degrees. As extension of the knee begins, there is a second burst of quadriceps activity associated with forward acceleration of the shank-foot. During the forward swing the quadriceps action decreases abruptly and the shank-foot swings as a pendulum. Almost immediately the hamstring muscles begin to act behind the knee to decelerate the rate of extension and allow the foot to move smoothly into contact with the floor.

This sequence is difficult to simulate accurately with the typical mechanical friction swing control mechanism. Figure 13 illustrates the resistance characteristic of a constant friction device with elastic extension bias. When properly adjusted, the amputee is able to walk reasonably well at the speed for which adjusted but the device lacks automatic adjustment for changes in walking speed.

Hydraulic devices of the type shown in Figure 14 offer much better possibilities for meeting the design objectives shown in Figure 12. A typical design incorporates multiple holes in the cylinder wall which are progressively covered by the piston to reduce the available flow area and create a highly non-linear resistance characteristic. Hydraulic devices can match the desired curves very closely. The peak moments of the characteristic curve are increased automatically with increasing walking speeds with a resistance level approximately proportional to the square of the walking speed.

Another method of achieving a variable moment pattern is with a pneumatic device of the type shown in Figure 13. The pneumatic devices rely upon a combination of air compression and controlled leak rate to achieve an approximation to the desired moment characteristic. As the walking speed is increased, the time available for air to leak from one side of the piston to another is decreased hence higher peak pressures are maintained at fast walking speeds.

Biomechanics of socket fitting

The following discussion assumes a socket in which the major support of the body weight is through the ischial tuberosity with additional contributions from the pressures acting upward against the gluteal musculature and provided by the flare of the anterior socket brim. Any pressure against the distal stump tissues is

assumed to be of low magnitude and of primary benefit in preventing the development of oedema in the distal stump.

As an example, consider the pressure patterns shown in Figure 16 acting between stump and socket as viewed from the side. The pressure

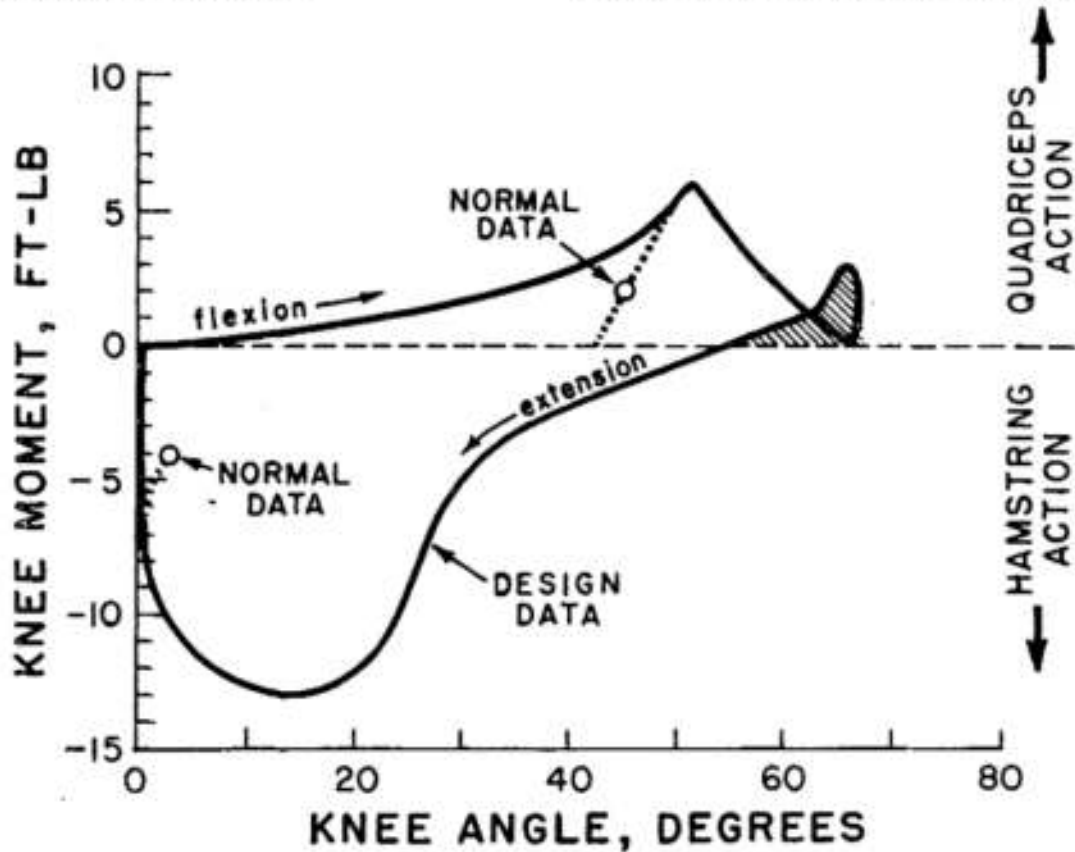


Fig. 12. Prosthetic knee moment vs. knee angle during swing phase.

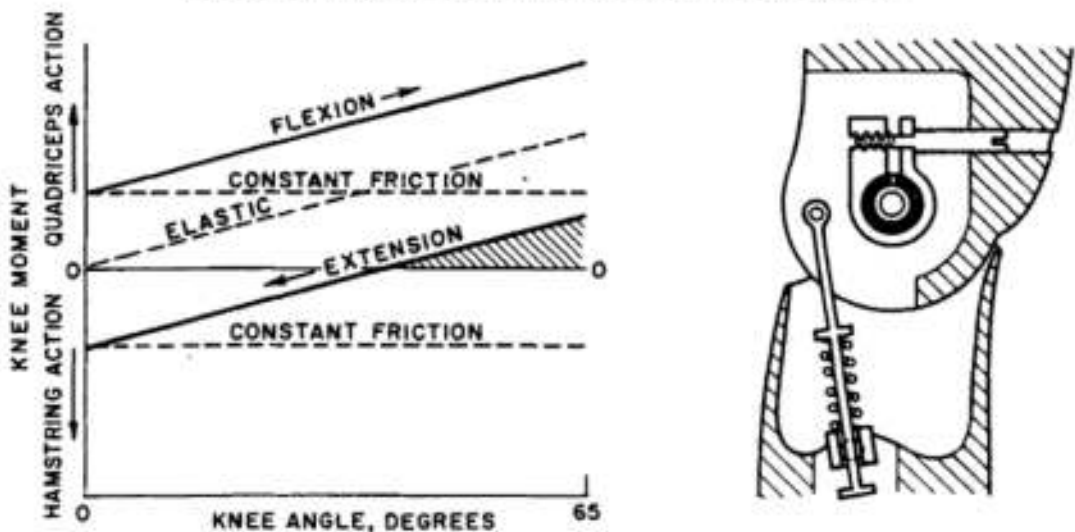


Fig. 13. Mechanical swing control with linear extension bias.

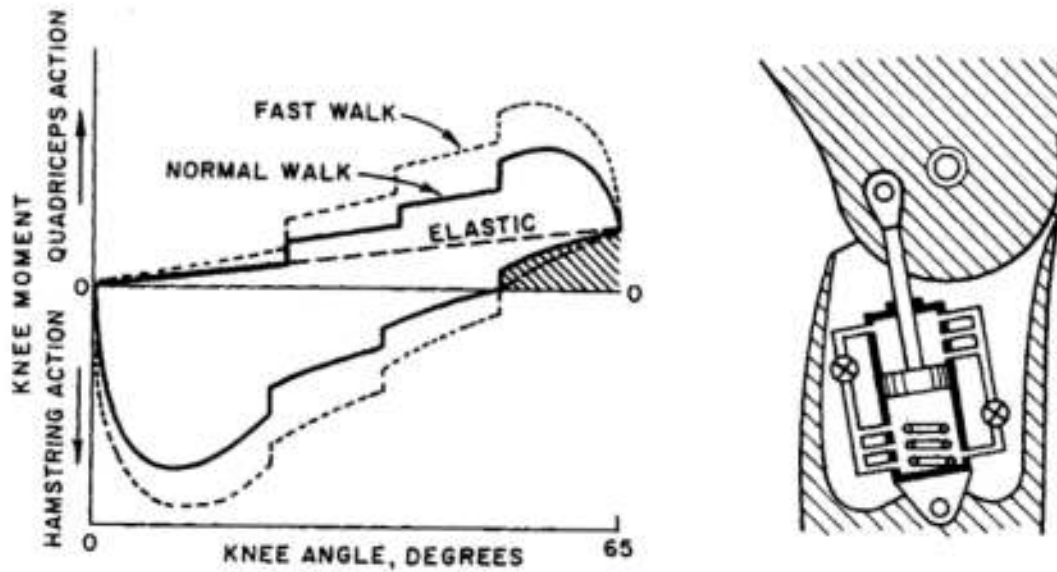


Fig. 14. Hydraulic swing control with linear extension bias.

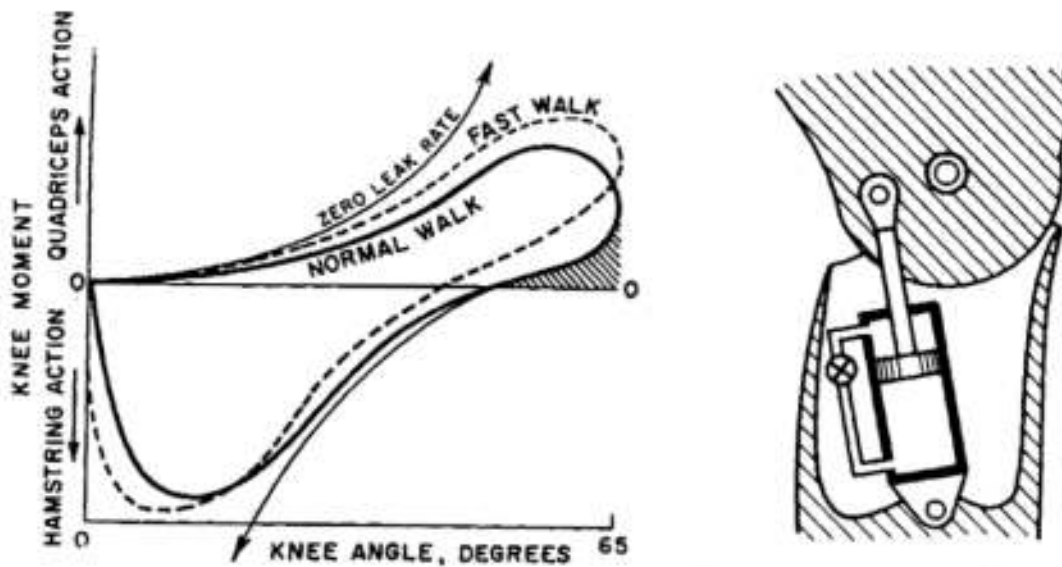


Fig. 15. Pneumatic swing control.

patterns shown are the result of the active use of the hip musculature to control stability at heel contact and initiate flexion at the end of the stance phase. Examination of such patterns reveals the need to pay particular attention to the proximal anterior brim area. The socket must provide sufficient anterior support to prevent the ischium from sliding forward and downward into the socket with inevitable painful contact of the pubic ramus on the medial

wall of the socket. At the same time, the fitting must anticipate the movement of the femur stump within the soft tissue as the femur first presses posteriorly to maintain knee stability then moves anteriorly to initiate knee flexion in the swing phase. The ischial seat should be sloped slightly inward and rounded to avoid skin irritation, but must provide a definite contribution to support of the body weight. The gluteal area of the posterior brim should fit snugly

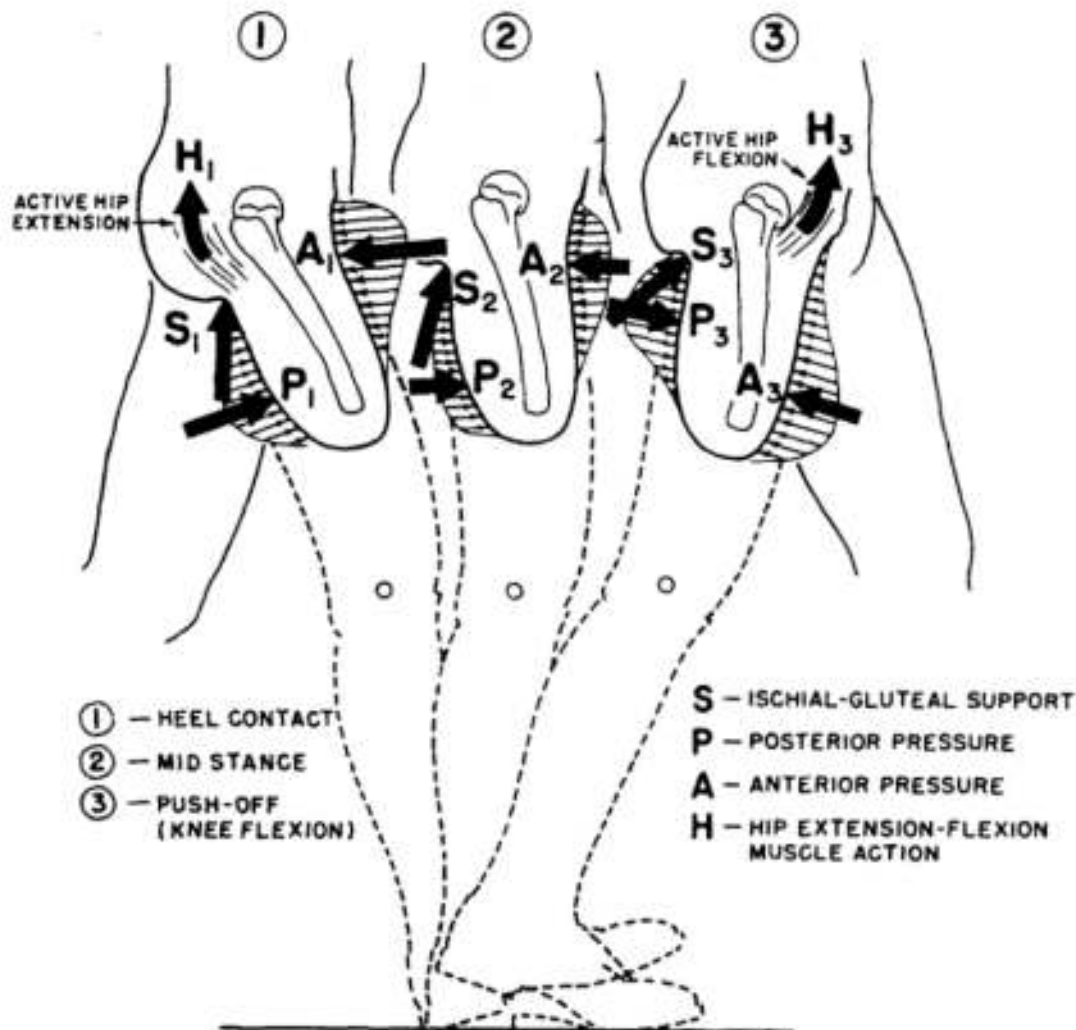


Fig. 16. Anterior-posterior stump-socket pressures during the stance phase of walking.

against the musculature to distribute the pressure on the socket brim and relieve the ischial pressure particularly in those cases where the ischium may be sensitive to excessive pressure.

Figure 17 illustrates the pressure patterns on the stump as viewed from the rear. In this case the need for lateral contact between socket and stump is clearly indicated. The lateral pressure pattern is associated with the balancing of the body weight above the level of the socket brim. The support point in the region of the ischium serves as a fulcrum for the body weight acting downward on the pelvis. The gluteus medius muscle acts to stabilize the hip joint which results in abduction of the stump against the lateral socket wall. This type of pressure pattern

always accompanies a narrow base gait with erect torso. The need for lateral pressure on the stump is diminished if the amputee walks with a wide walking base or leans the torso over the support point during the stance phase.

The quadrilateral socket shape

The socket shapes shown in Figure 18 are based upon the requirements as discussed in the previous section. The shape is not a duplicate of the free stump since soft tissue must unavoidably be compressed in areas where higher pressures must be tolerated on soft tissue, particularly in the Scarpa's triangle area along the anterior aspect of the stump and in the

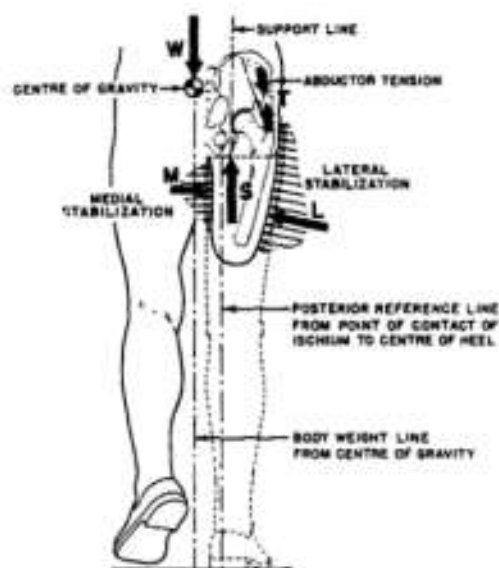


Fig. 17. Use of hip abductor musculature for lateral stabilization of the pelvis.

gluteal region of the posterior brim. Regions of firm musculature such as along the rectus femoris muscle are channelled to avoid excessive pressure as required. Tendon insertions on the pelvis such as the hamstring tendons or adductor longus must be accommodated in appropriate channels in the posterior-medial and anterior-medial corners of the socket. The width of the medial wall of the socket is defined by the anatomical dimension measured between ischium and adductor longus. If this dimension is incorporated into the socket accurately, the pubic ramus will always be supported well above the medial brim of the socket. If the medial dimension is too wide, the ischium will slide forward during weight bearing with inevitable painful pressure against the pubis.

The contours of the socket must always be considered in three dimensions. One of the most difficult problems in the fitting process is to achieve the proper degree of tightness of fit along the length of the stump. With open-end sockets carved from wood the socket circumference dimensions were often less than the free stump circumference with the result that the soft

tissues of the stump were pulled deeply into the socket as the socket was donned. With the modern total-contact suction socket, this reduction in socket circumference is often less and the socket dimensions more nearly approximate the free stump measurements.

The pressure pattern created between the socket and the tissues of the stump and pelvis is a function of the forces and moments being transmitted at any instant during the walking cycle. The pressure pattern varies dramatically over the walking cycle and it is remarkable that a socket of fixed shape can transmit these pressures comfortably for long periods of use.

Conclusion

Much progress has been made over the past 30 years in the refinement of principles, devices, fitting and alignment procedures and materials used for above-knee prostheses. Prosthetists all over the world have been trained in the application of modern methods and components and in many cases have successfully adapted local materials in the construction of prostheses.

There is considerable current interest in the design of modular components with international standards for dimensions which will allow all nations to benefit from advances in component design and manufacture.

The major problems to be solved are generally associated with the increasing age of the amputee population. Lighter weight prostheses which can be produced quickly and at minimum cost are required. Problems associated with cosmesis remain to be solved. Improved patient management procedures to reduce the time between amputation and fitting with a permanent prosthesis must be developed if the elderly amputee is to be rehabilitated. Amputee training procedures which have proved successful with young vigorous amputees must be re-examined to account for lower levels of physical capability. Finally, we must encourage the dissemination of knowledge on an international basis so that all persons concerned with the welfare of above-knee amputees can benefit from the success of our more progressive institutions.

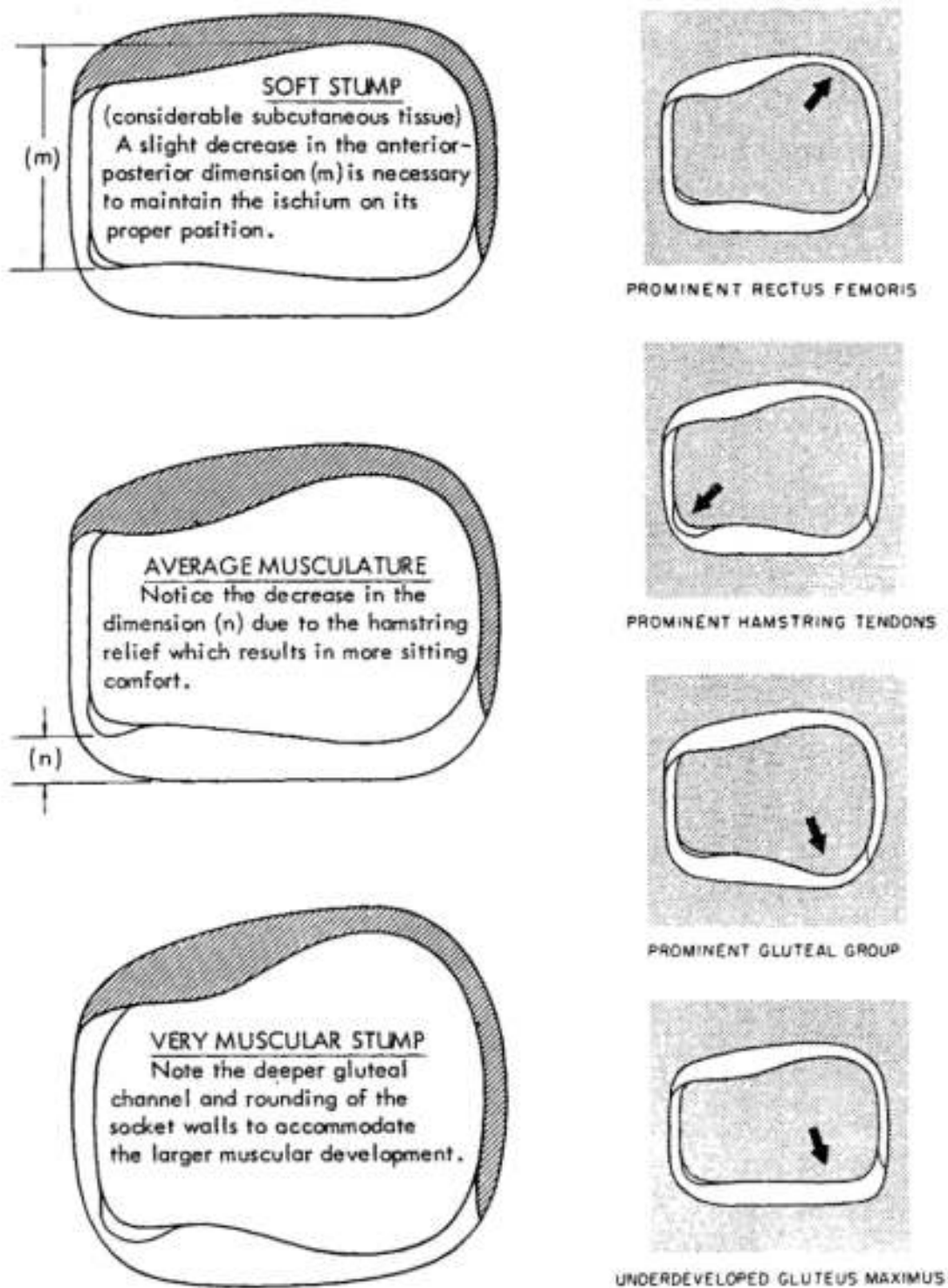


Fig. 18. The quadrilateral suction socket shape.