

Four-bar linkage prosthetic knee mechanisms: kinematics, alignment and prescription criteria

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Introduction

Four-bar linkage knee mechanisms for the trans-femoral amputee are widely available, but although they may offer functional advantages to certain amputees, they are fitted in a limited number of cases. It may be assumed that one reason for this is that persons responsible for prescription and fitting may not be familiar with the kinematic characteristics and possible advantages of such mechanisms and are reluctant to use a device which they do not understand completely.

This paper will describe the kinematics of several types of four-bar mechanisms, and discuss the differences and prescription criteria for three different classes of four-bar linkage mechanisms currently available for fitting to amputees. Before beginning the discussion of four-bar prosthetic knees, it will be helpful to review some fundamental concepts.

The load line

The line along which the equivalent single load force acts on a weight-bearing prosthesis seldom, if ever, acts along a line directed from the hip joint to the ankle. Neither does it, in general, act from a single point at the level of the socket brim to the centre of pressure on the sole of the foot. The location and direction of the load line can be measured by a force plate during walking and it is constantly changing its location and direction with respect to the geometric long axis of the prosthesis (or anatomical long axis for the non-amputated side).

The direction of the load line as seen from the medial or lateral side for a trans-femoral amputee is directly related to the stability of the prosthetic knee. When the load line passes anterior to the knee joint axis the prosthetic knee is forced into full extension against the extension stop. In order for the knee to flex while bearing weight at push-off the load line must shift to a position where it passes posterior to the knee centre. The amputee can actually control the direction of the load line as seen in the mediolateral view by active use of the flexion-extension musculature about the hip joint of the stump. This leads to the concept of "voluntary control of knee stability" which is of particular interest in the design and use of certain four-bar or other polycentric knee mechanisms. The same concepts are also important to any trans-femoral amputee using a non-locked single axis knee. A trans-femoral amputee with a weak hip who is unable to exert the necessary muscle effort would obviously have greater difficulty in maintaining knee stability without dramatic changes in alignment stability or installing a brake type mechanism.

A lesson in basic engineering mechanics may help to explain how the muscle moments exerted by the amputee influence the direction of the load line. Consider the schematic diagrams of Figure 1. An outline of a trans-femoral prosthesis is shown in Figure 1(a) at heel contact, the most critical period of the stance phase for knee security. The upper reference point is arbitrarily selected at the hip joint which allows the analysis to proceed without considering the manner of socket fitting. In diagram (a) a prosthesis is shown at the instant that weight bearing begins. In this case the amputee is not exerting an extension

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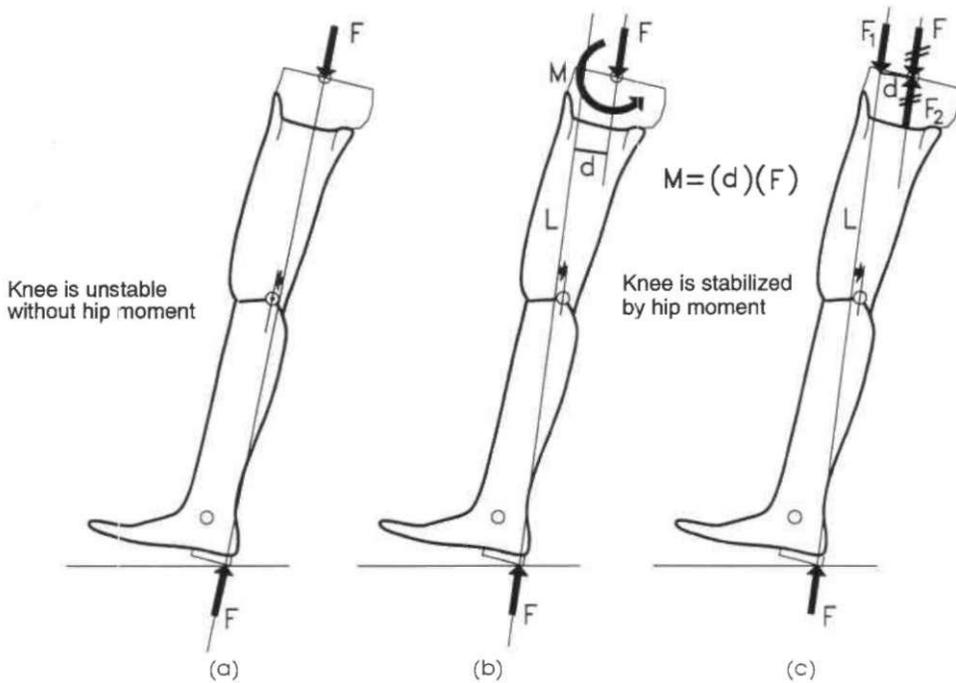


Fig. 1. A single force F_1 acting along the load line L is equivalent to force F and moment M acting about the hip of trans-femoral amputee.

moment about the hip and the load on the prosthesis would be a direct thrust from hip joint to heel contact point. In diagram (a) the "load line" passes behind the knee centre and the knee would buckle under load. What is the mechanism by which the knee is made secure so that it will not buckle?

Consider Figure 1(b). In this case the amputee is exerting an extension moment about the hip. This tends to drive the heel into the ground and the ground pushes back on the heel. Since Newton's law states "action = reaction", the result is a second component of force acting forward on the heel and the load line inclines in front of the knee centre giving a stable knee.

This phenomenon can also be explained by considering the combination of joint force and extension moment acting at the hip joint. Consider Figures 1(b) and (c). One may replace the hip extension moment M in (b) by a pair of equal and opposite forces of magnitude F separated a distance D which has the same extension moment as M , i.e., $M = (D) \times (F)$. This pair of forces F_1 and F_2 are now placed on diagram (c) in a position such that F_2 is in-line with the actual force F . The moment M has

been replaced hypothetically by the forces F_1 and F_2 offset by the distance D and the forces F and F_2 cancel each other. The equivalent inclined load line from the heel would pass ahead of the hip joint by the distance D . F_1 and the heel force F act along the same load line.

This variable location of the load line during the dynamic events of the stance phase of a walking cycle make the definition of a "load line", which can be drawn or visualized on a lower limb prosthesis, dependent on knowing the complete set of forces and moments acting on the prosthesis at each phase of the walking cycle. In the standing at rest position, it also requires more data than is available to the prosthetist and many assumptions would have to be made. It is therefore best to leave the concept of an exact load line to laboratories with the instruments to measure it and use a "vertical reference line" to describe the geometry of a lower limb prosthesis relative to this line.

Bench alignment vertical reference lines

A vertical line, or plumb line, is the reference line used by prosthetists for the "bench

alignment" of the prosthesis. This line is used to assemble the components of the prosthesis such that the prosthesis will provide stability in weight bearing during the first walking trials when fitted to the amputee. It is anticipated that small changes in the bench alignment settings will be made during the first walking trials, a procedure known as "dynamic alignment". With careful bench alignment, the changes during dynamic alignment should be small, and necessary only to fine tune the prosthesis for the needs of an individual amputee.

The German bench alignment system

In Germany, and some other European countries, a bench alignment system is often used which shows the prosthesis in a reference position corresponding to the highest position of the hip joint as the amputee rolls over the ball of the foot in the stance phase. This results in a vertical reference line which extends downward from the hip joint (sometimes assumed to pass through the centre of stump cross-sectional area at the brim level) to approximately the mid-length point of the foot. The bench alignment procedure involves specifying offsets to locate the knee joint centre and ankle joint centre posterior to this vertical reference line. This system was developed over 50 years ago and was considered necessary to account for the large number of knee and foot mechanisms used in Germany. Each knee-foot combination has a specific pair of offsets.

The reference position, with the foot assumed to be bearing weight on the ball of the foot, also requires the incorporation of a specific air space under the heel (the "safety factor") for each style of foot. The illustrations used show an exaggerated heel height similar to a cowboy boot. This is not the heel height for the shoe but the sum of the heel height plus the safety factor. The safety factor improves the stability of the knee at heel contact during walking since it is equivalent to plantar flexion of the foot during bench alignment.

The UC Berkeley bench alignment system

At the Biomechanics Laboratory, University of California at Berkeley, comparative studies in 1964 of the European bench alignment system with the then commonly used TKA line (Trochanter-Knee-Ankle vertical reference line as seen from the lateral side) concluded that the

actual optimal alignment obtained was very similar, particularly for single-axis knee mechanisms. The major source of difference was the difficulty in locating the trochanter or upper reference point relative to the upper socket brim. Note that to locate the trochanter relative to the socket brim the amputee must be present and fitted into the socket. Rotation contracture of the femur relative to the pelvis or the presence of excessive soft tissues can cause significant error in location of the trochanter point on the socket.

A new system was adopted which did not depend upon the location of the trochanter but arbitrarily used the Bisector of the Medial Brim (BMB point) as the upper reference point on the medial socket brim. This point can be located accurately and easily on a typical quadrilateral shaped trans-femoral socket and does not require that the amputee be present. If the socket deviates from a quadrilateral shape, the centre-of-area at the brim level can be projected to the medial wall of the socket as a reasonable approximation.

As viewed posteriorly the upper reference point on the socket is the point of contact of the tuberosity of the ischium. This point can be estimated by palpation of the point of contact of the ischium relative to the socket brim. Fortunately this point need not be located as accurately as the BMB point and a good approximation is usually adequate for bench alignment. This point is used to estimate the outset of the foot relative to the socket. The outset of the foot can vary from a negative value for very active walkers with a long stump to a positive 5 cm (2 in) or more for very short stumps. A positive outset is required whenever the amputee is not capable of using the stump for lateral stability and must bring the load line as seen posteriorly to a position which passes close to the hip joint.

In the Berkeley system the alignment of the prosthesis is referred to lines drawn on the medial and posterior sides of the prosthesis. The vertical reference line is the centre line of the shank pylon tube and the medial side of the prosthesis is much more convenient for viewing an extension of the shank tube centre line from above or checking the alignment by use of a plumb bob in a vertical fabrication jig. The vertical reference line, in both posterior and medial views of the prosthesis, becomes the

reference for offsets or angles for alignment of foot, knee, and socket during bench alignment. At Berkeley the use of endo-skeletal systems with aluminium tubing as the shank structure and the SACH (Solid Ankle Cushion Heel) foot without ankle joint was the preferred system dating back to the 1950s. It was convenient to use the centre of the shank pylon as the vertical reference line and refer the heel lever arm and forefoot lever arm for the foot, knee centre anterior-posterior location, the socket flexion and adduction, and the anterior-posterior location of the socket at the brim level to this line. The recent popularity of a variety of endo-skeletal systems and energy return feet without ankle joints clearly makes the use of the shank pylon centre line the system of choice for the vertical reference line.

Comparison of the German and Berkeley reference positions

As a demonstration that good bench alignment can be achieved using either the UC Berkeley system or the German system, the two systems are compared in Figure 2. This illustration is similar to one developed during the 1964 studies by the Danish prosthetist, Erik

Lyquist, while he was a visiting research prosthetist at the UC Biomechanics Laboratory. Note that the same basic diagram is used in both cases. The only difference is the reference position. The Berkeley system is much easier to check in the vertical alignment jig or by use of a plumb bob with the completed prosthesis.

Bench alignment as viewed in the anterior-posterior direction

Many modern lower limb prosthesis systems incorporate built-in alignment devices which allow changes in the angle of the foot relative to the shank tube, angular changes between knee mechanism and shank tube at the proximal end of the tube and angular (and sometimes translational) adjustments between knee mechanism and the distal end of the socket.

Given such a plethora of adjustments, the prosthetist is often tempted to use them all. All can be useful in certain circumstances, but it must be recognised that in some cases too many adjustments can cause conflicting results and often do more harm than good. Figure 3 shows medial and posterior views of typical bench alignment settings using the UC Berkeley system for a single axis knee mechanism (without a

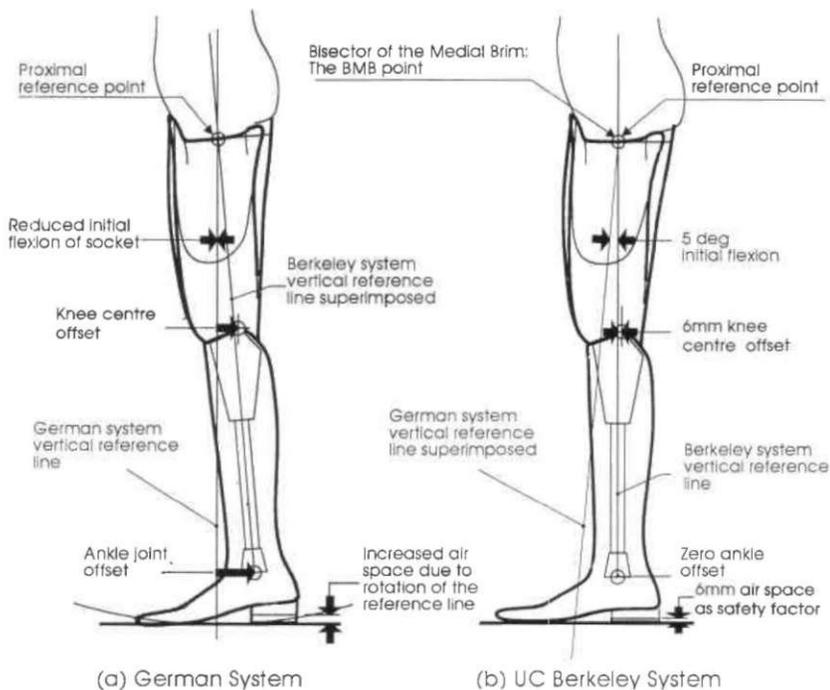


Fig. 2. Bench alignment - medial view comparison of German and UC Berkeley systems.

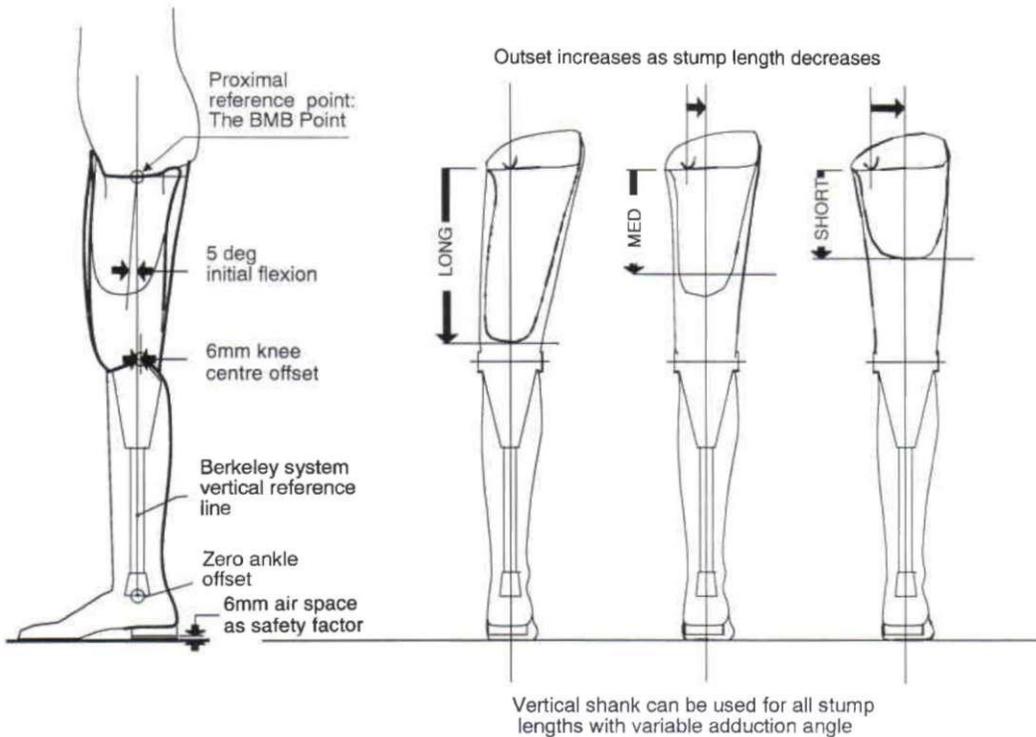


Fig. 3. Typical bench alignment for varying stump length UC Berkeley system.

brake or hydraulic stance control) and a SACH foot (with *soft* heel cushion) for amputees with short, medium, and long stumps. Only minor changes from these guidelines should be required if the amputee is fitted comfortably and has good control of the prosthesis through his or her stump-socket fitting.

Increasing knee-stability during dynamic alignment

The proper way to increase the "alignment stability" of the knee is to increase the posterior offset of the knee centre relative to the vertical reference line. In modular endo-skeletal systems this dimension is set by the manufacturer and built into the hardware. This makes it necessary to check the posterior offset for the knee mechanism being fitted. The vertical reference line dropped from the bisector of the medial brim (the BMB point) should pass a *minimum* of 6 mm ($\frac{1}{4}$ in) ahead of a single-axis knee centre as viewed from the medial side. Remembering that the knee bolt is typically rotated externally about a vertical axis by approximately 5 degrees, the knee centre as

viewed from the lateral side would be more than 6 mm ($\frac{1}{4}$ in) posterior, perhaps as much as 12 mm ($\frac{1}{2}$ in).

During dynamic alignment the prosthetist can increase knee stability by increasing the posterior offset of the knee axis using the following two-step procedure:

1. Extend the socket a small amount in approximately one degree increments using the adjustable coupling between socket and knee.

This will shift the upper reference point at the brim of the socket forward approximately 6 mm ($\frac{1}{4}$ in) per degree of socket extension. After the adjustment the amputee must bring the upper reference point back to the standing position by a posterior rotation about the point of support on the foot which will result in a decrease in the air space (safety factor) previously set during bench alignment.

2. Plantar flex the foot a small amount using the adjustable coupling at the ankle.

The plantar flexion adjustment will restore the air space to the desired amount. The final

result will be a posterior shift of the knee axis with minimal disturbance of the desired socket-foot relationship.

The amount of forward shift (S) of the BMB point per degree of socket extension (E) for a given BMB-knee dimension (D) can be computed easily using the following formula:

$$S = (D)(E)/57.3$$

For example to find S, given D = 380 mm (15 in), E = 1 degree

$$S = (380)(1)/57.3 = 6.6 \text{ mm (0.26 in) (per degree of extension)}$$

where 57.3 is the factor for conversion of degrees to radians

Therefore, it is apparent that small changes in the socket angle can have a major effect on the posterior offset. The compensating angular change at the ankle would be somewhat smaller. Assuming a length (L = 760 mm (30 in)) from BMB to ankle, the angular change (A) at the ankle would be computed from the formula:

$$A = 57.3(S)/(L)$$

$$A = 57.3(6.6)/(760) = 0.50 \text{ degrees plantar flexion}$$

To decrease the built-in offset, the socket is flexed a small amount and the ankle is dorsiflexed.

Effect of foot adjustments on knee stability

It should be emphasized that a bit more plantar flexion will probably be welcomed by the amputee since it will give more stability at heel contact. To achieve the optimal settings there are several foot function adjustments that will have to be checked with the amputee:

1. If the amputee complains that heel pressure is not felt while standing, have the amputee move the foot rearward and stand on the ball of the foot. Explain to the amputee that this is necessary if dynamic knee stability at heel contact while walking is to be achieved efficiently. This may require patience, practice and retraining, but the result will be worth the time and effort. When an air space is incorporated into the bench alignment, the amputee should not contact the heel on the floor when standing at rest. Clearly this system is not recommended for bilateral

trans-femoral amputees.

2. If the amputee complains of the leg being too long, first check the leg length by examining the height of the iliac crests and spinal curvature with equal weight distribution on both feet. If the prosthesis is too long, the shank tube may need shortening, the foot may have too much air space under the heel, or the ankle has been plantar flexed more than required for stability at heel contact.
3. If the amputee complains of lack of support on the ball of the foot during roll-over try increasing the amount of plantar flexion. This is particularly important with energy return feet which must store energy by deflection of the keel (forefoot) under load.

In walking, the foot should approach the floor in a slightly plantar flexed attitude, the centre of pressure should move quickly and smoothly forward from the heel to the ball of the foot, and the amputee should not have the sensation of either "walking over a hill" or "lack of support during roll-over".

The knee will be secure as soon as pressure is established on the ball of the foot. Most knee instability problems are due to 1) a dorsiflexed foot, 2) a stiff plantar flexion bumper, or 3) a firm heel cushion action. Any of these factors will prolong the time of weight bearing on the heel.

If a SACH foot is used the heel cushion must be softer than for a trans-tibial amputee. Use a soft grade and ensure the foot is fitted in the shoe to allow at least 9 mm ($\frac{3}{8}$ in) of heel compression. An energy return foot should be plantar flexed to account for the forefoot deflection at push-off. The anterior-posterior location of the foot relative to the shank centre line would also be important as related to the knee centre offset. But again, the foot position is usually determined by the mechanical design of the foot and the ankle connector.

In bench alignment the proper location of the upper reference point on the socket, along with the knee offset, should provide the necessary knee stability. Adjustments to knee stability should be done first at the foot and ankle.

A prosthetic foot of any type must provide three functions:

1. Shock absorption at heel contact
2. Smooth transition to a stable weight bearing mode on the ball of the foot

3. Smooth transition from stance to swing phase

An energy return foot may be important to athletic individuals but is not a necessary feature for many amputees.

The knee-stability diagram

One way to compare stability characteristics of either single-axis or four-bar linkage knee mechanisms is to visualize the contribution of the residual hip musculature on the amputated side to the stability of the knee during the stance phase of walking.

Figure 4 shows schematic diagrams of the equivalent forces and moments acting on the foot and about the hip joint of a typical transfemoral amputee fitted with a single-axis knee mechanism. Two diagrams are shown corresponding to two phases of the walking cycle: (a) heel contact and (c) push off. These two diagrams are then superimposed as shown in (b). Note that these diagrams would apply to any method of socket fitting, assuming that the socket is comfortable and allows the amputee to exert muscle moments about the hip joint as

seen from the side.

Note that the line of the floor reaction force does not pass through the centre of the hip joint at either heel contact or push off. At heel contact, the line of action of the force on the heel must pass ahead of the prosthetic knee centre in order for the knee to be stable during the heel contact – shock absorption phase. The amputee actually controls the orientation of the force line by actively exerting a small extension moment about the hip.

The same principles apply to the force system at push off. At push off the amputee should be able to initiate knee flexion for the transition into the swing phase without lifting the prosthesis from contact with the floor. This is accomplished by a flexion moment from the hip musculature which has the effect of shifting the force line originating at the ball of the foot to an orientation which passes behind the knee joint centre and would cause the knee to flex.

If the diagrams of Figures 4(a) and 4(c) are superimposed and it is assumed that the amputee is capable of exerting the required muscular moments about the hip joint, the stability diagram shown in Figure 4(b) is

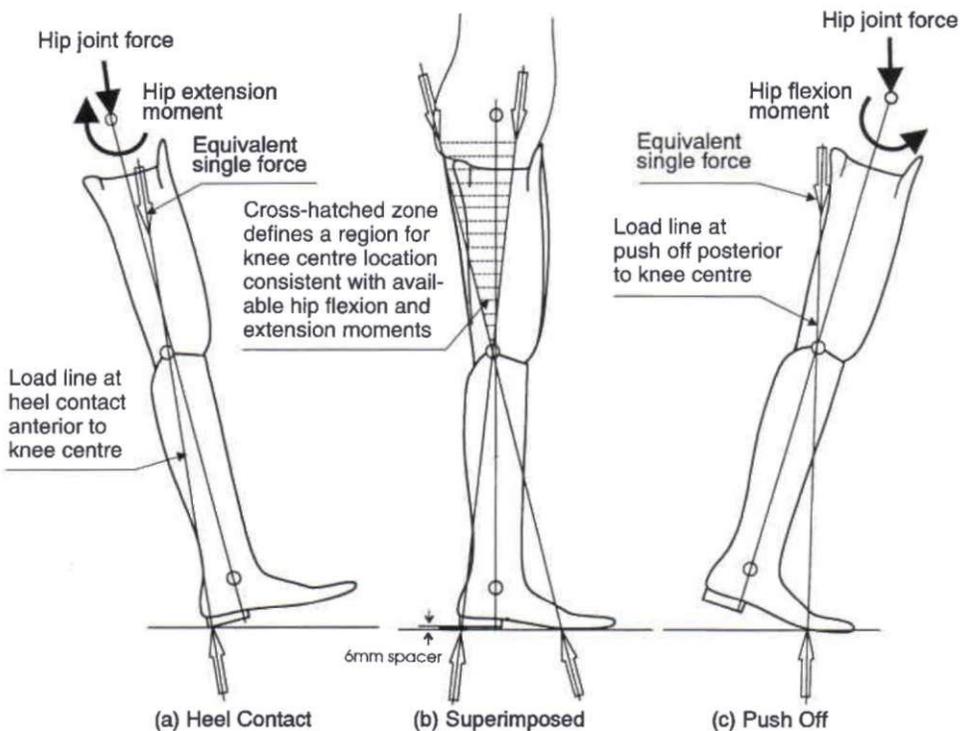


Fig. 4. Stability diagram – single axis prosthetic knee mechanism.

obtained. The cross-hatched area represents an area where the knee centre in full extension can be located and still maintain the two desired characteristics: 1) stability at heel contact and 2) ability to initiate knee flexion voluntarily just prior to push off. The actual required muscular effort on the part of the amputee will vary depending on the alignment of the knee joint within this V-shaped region.

The diagrams of Figure 4 have been drawn for a typical trans-femoral prosthesis with knee stability determined solely by alignment of the knee centre. No friction brake mechanism is assumed to be incorporated into the knee mechanism. Many active trans-femoral amputees have the ability to exert muscle moments about the hip joint much larger than the moments required in Figure 4. It should also be obvious that flexion and extension moments about the hip are absolutely essential. An amputee with a weak hip could not control a single-axis knee without a knee brake or lock.

The instant centre

The "instant centre", or more properly the "instantaneous centre of zero relative velocity", is a point where, for a very small change in the angle of knee flexion, the thigh section rotates about a point on an extension of the shank which appears to be temporarily fixed. For small angles of relative rotation one could imagine a temporary hinge connecting the shank and thigh sections at the instant centre. For larger angles of rotation the instant centre will change its location and a new temporary hinge must be imagined.

For a four-bar linkage knee, the instant centre (in any position of knee flexion) can always be located at the intersection of the centre lines of the anterior and posterior links which connect the socket section to the shank section of the prosthesis. As the knee flexion angle is increased the instant centre takes a series of positions which typically trace a path on an extension of the shank which progresses forward and downward toward the cosmetic or anatomical knee centre.

An elevated and posterior location of the instant centre will increase knee stability (see the Appendix). With a single axis knee, the location of the knee joint is also dictated by placing it at an approximate anatomical location with good cosmetic appearance while seated

with the knee at 90 degrees of flexion. A properly designed four-bar linkage knee also allows the possibility of locating the instant centre in full extension in a position within the desired stable region of the stability diagram, yet which maintains acceptable cosmetic appearance at 90 degrees of flexion. Small differences in link lengths and pivot locations can result in major changes in the kinematic behaviour of four-bar linkages, as will be discussed in the following paragraphs.

The knee stability diagram is useful in comparing the characteristics of three different classes of four-bar linkage prosthetic knee mechanisms: 1) the four-bar mechanism with elevated instant centre, 2) the hyper-stabilized four-bar, and 3) the voluntary control four-bar. Each of these three classes of four-bar knee mechanisms has a place in the fitting of different groups of trans-femoral amputees.

The four-bar linkage with elevated instant centre

The four-bar prosthetic knee with elevated instant centre has been available for many years and has the general appearance shown in Figure 5. It typically has a long anterior link and a short posterior link. A four-bar linkage knee of this class offers considerable stability at heel contact and is of primary benefit to geriatric amputees or other amputees with limited ability to control stability through active and voluntary control using residual hip function on the amputated side. Devices of this type can be designed to give a variety of functional characteristics depending upon the arrangement and length of the link lengths, pivot locations, and extension stop adjustment.

The links of the hypothetical device shown in Figure 5 have been designed to give extreme stability at heel contact by having the instant centre for full extension of the knee located considerably posterior to the load line at heel contact as shown in Figure 2(a). The knee is forced into extension and is essentially kinematically locked in extension. No hip extension moment exerted by the amputee is required.

At push off the hip flexion moment exerted by the amputee, with help from the offset load on the ischial seat, is easily capable of shifting the load line behind the instant centre as required to initiate knee flexion. The elevated

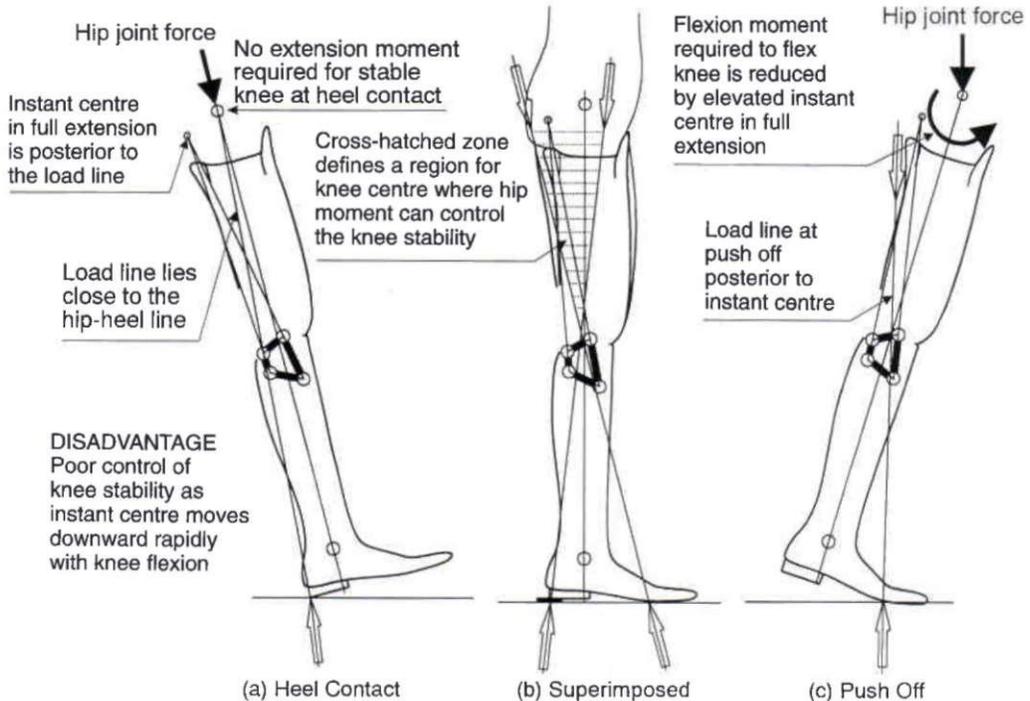


Fig. 5. Stability diagram – four-bar knee with elevated instant centre.

position of the instant centre has been shown to contribute to the ease by which the hip moment can maintain a stable weight bearing knee. Thus the elevated posterior location of the instant centre allows the possibility of initiating knee flexion with minimal effort by the amputee. At first glance this class of mechanism appears to have several advantages for the trans-femoral amputee.

However there are some limitations. All classes of four-bar linkage knees can be designed to provide a reasonable cosmetic appearance at 90 degrees of flexion. Note however, that to allow good cosmesis in sitting, an elevated instant centre must move downward very rapidly with knee flexion. This sudden shift of the instant centre does not allow the amputee to maintain control of the weight bearing knee if the knee flexes a few degrees as a result of some unforeseen event. However, a knee mechanism with an elevated and posterior instant centre in full extension will essentially be locked in extension at heel contact. At the end of the stance phase, the amputee typically initiates knee flexion by initiating swing-through from the hip after lifting the prosthesis from contact with the walking surface. This

class of knee mechanism has a definite place in providing knee stability similar to a locked knee for geriatric amputees or other amputees with limited physical capability.

It should be noted that the position of the instant centre in full extension is very sensitive to small changes in the full extension angle. This feature can be used as a means of adjustment of the knee stability in full extension by small changes in the position of the knee extension stop. With proper adjustment the initial instant centre location can be moved to the desired offset from the vertical reference line and serve as simple means of adjusting knee stability at both heel contact and push off.

The hyper-stabilized four-bar knee mechanism

The physical arrangement of the hyper-stabilized four-bar knee mechanism will often be similar to the four-bar with elevated instant centre. In this case the kinematic behaviour can be dramatically different with only small changes in dimensions. Figure 6 shows the kinematics of a typical device of this class. Note again that the initial location of the instant centre, hence the stability at heel contact, can be

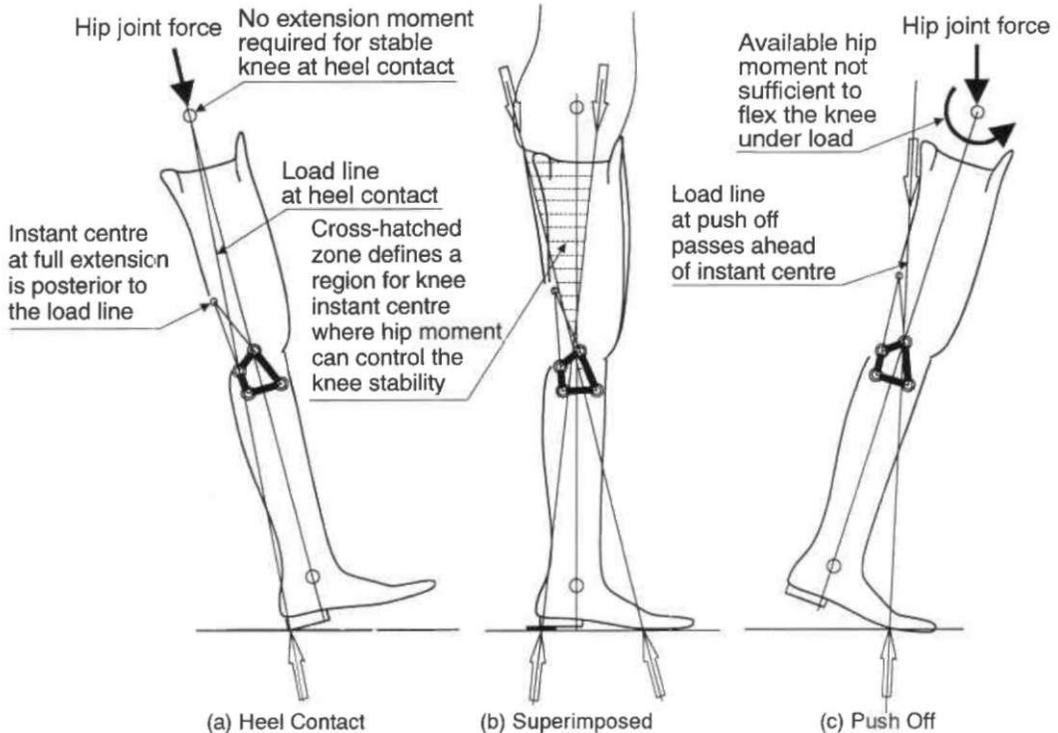


Fig. 6. Stability diagram – hyper-stabilized four-bar knee mechanism.

fixed easily by the designer at the intersection of the centre lines of the anterior and posterior links with the knee in full extension. The determination of actual link lengths and initial pivot locations which will allow for good cosmesis during knee flexion and at 90 degrees of flexion is a greater challenge.

The term “hyper-stabilized” refers to the very positive alignment stability built into devices of this type. As shown in Figure 6(a), the instant centre in full extension is located well behind the load line, which, with no hip extension moment required, lies close to the hip-heel line at heel contact. At push off, even with the maximum possible hip flexion moment exerted by the amputee, the instant centre is still behind the load line and it is not possible for the amputee to initiate knee flexion while the prosthesis is weight bearing. Again, this class of four-bar knee mechanism is of primary interest for geriatric or otherwise less active amputees who require the equivalent of a knee lock in the stance phase of walking.

It should be noted that an over-stabilized prosthetic knee with excessive alignment stability can lead to problems for the active

amputee. There would be many situations where activities of daily living will be difficult if not impossible due to the inability to flex the knee in a controlled manner while weight bearing. These activities would include stair and ramp descent foot over foot, sitting from a standing position in cramped quarters such as a theatre seat, entering and exiting automobiles, etc.

The voluntary control four-bar mechanism

Figure 7 shows a four-bar linkage for which the instant centre lies within the stability zone at both heel contact and push off. In this case the initial elevation of the instant centre is not as high as for the linkage of Figure 5 and the downward path of the instant centre stays somewhat elevated and within the zone for the first few degrees of knee flexion. Figures 8 and 9 show the path of the instant centre for devices of this class. Each point on the path is labelled with the corresponding angle of flexion of the socket relative to the shank of the prosthesis.

The voluntary control four-bar knee is designed to give the amputee the ability not only to control knee stability at both heel

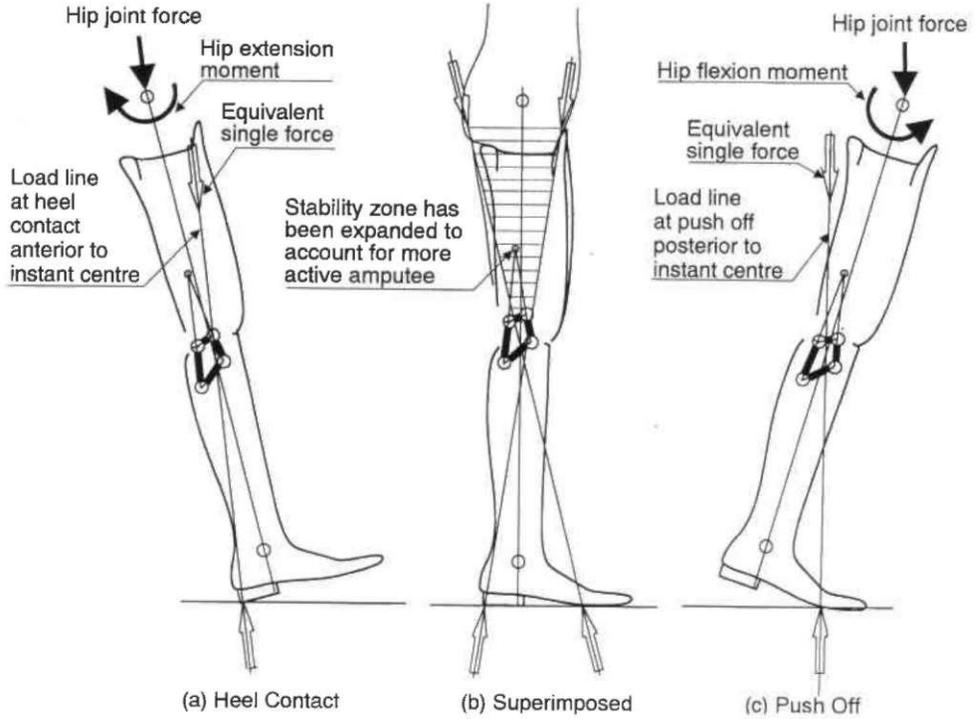


Fig. 7. Stability diagram – Hosmer voluntary control four-bar knee mechanism.

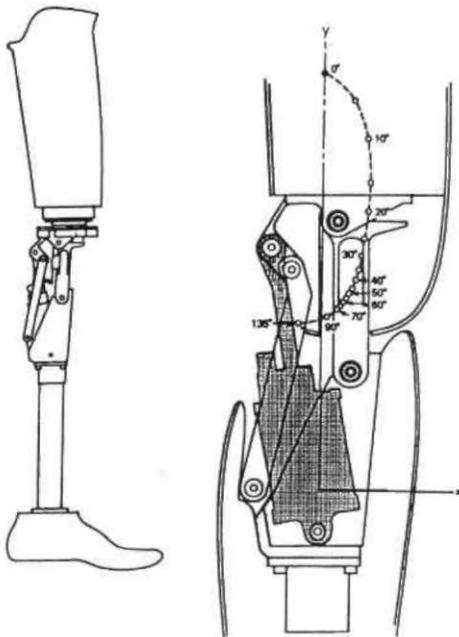


Fig. 8. Path of instant centre UCBL four-bar polycentric knee.

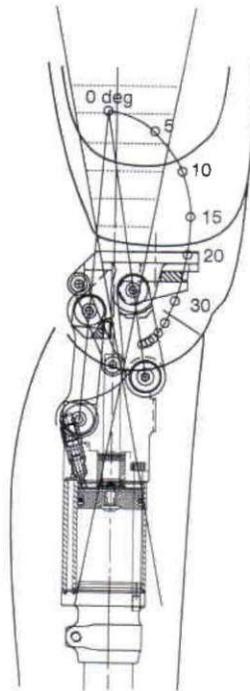


Fig. 9. Path of instant centre – VC Hosmer voluntary control four-bar knee.

contact and push off, but to have complete control of knee stability over a limited range of knee flexion. The actual ability to control the motion and stability of a flexed knee depends upon the physical capabilities of the amputee. It is desirable for the amputee to be able to react to an event which might disturb the stability of the weight bearing knee, particularly at heel contact, to arrest the tendency toward uncontrolled flexion and voluntarily move the knee to a stable position in full extension. As the amputee gains experience in the use of a voluntary control four-bar prosthetic knee, these reactions become almost involuntary since this is the way a non-amputee senses and reacts to control knee stability.

Voluntary control of the stance phase stability is most important over the first 10 degrees of flexion from a position of full extension. Figure 8 illustrates the path of the instant centre for the University of California four-bar polycentric knee. This design has been extensively tested and the kinematics have been shown to offer many advantages to active patients with a desire to be aggressive walkers. The voluntary control is perceived by the patient as a "gliding motion" of the knee block forward and backward over the shank with controlled flexion and extension of the knee. This has been shown to offer benefits in controlling stability while the patient walks on rough ground, sloping surfaces, or stair descent. Other advantages include the ability to bear weight on a slightly flexed knee while taking short steps around a counter, dancing, playing golf, etc. Voluntary control may not be optimal for geriatric patients, although devices of this type have been fitted successfully to both active older patients and active patients with very short stumps.

The path of the instant centre for a voluntary-control four-bar knee mechanism does not begin at the extremely elevated and/or posterior position in full extension, as has been shown for the other two classes of four-bar mechanisms. The full extension location of the instant centre is approximately 100 mm (4 in) above and 6 mm ($\frac{1}{4}$ in) posterior to the vertical reference line. This location allows the path of the instant centre to move smoothly forward and downward with increasing angles of knee flexion yet stay in an elevated position within the stability zone for as much as 10 degrees of

flexion. The angles noted at points circled along the path of the instant centre indicate the corresponding angle of knee flexion.

The arrangement shown in Figure 9 is a new design of a four-bar prosthetic knee, with voluntary control characteristics very similar to the original University of California model, which has been developed by Hosmer-Dorrance in cooperation with the author. The new model is designed to provide equal or better kinematics in a package which is well suited to installation in current modular prostheses. The original pneumatic swing phase control cylinder and two-way valve have been retained with minor modifications to allow installation as part of the shank structure.

There are several functional advantages for the mechanisms of Figures 8 and 9 which may not be immediately apparent:

1. Ease of slope and stair descent. The experienced and active user can descend stairs using the "jack-knife" method without having to place the heel on the outer edge of the stair tread. The technique involves placing the foot in the normal position then flexing the knee under load and allowing the opposite foot to drop down to the next step. This can actually be done by locating the heel of the prosthetic foot rearward against the riser of the step, even with poor visibility.
2. Approximately 130 degrees of knee flexion. This feature becomes very important in entering and exiting automobiles. The amputee can sit on the seat then reach down and lift the flexed prosthesis into place rather than place the extended prosthesis into place before sliding into the seat. The extra degrees of knee flexion can also be helpful in many kneeling activities.
3. Increased toe clearance during swing phase. Rotation about an elevated instant centre creates a posterior translation component for the motion of the shank-foot which as the knee flexes effectively raises the toe of the foot more than 25 mm (1 in) higher than would be seen with a single axis knee. It is virtually impossible to stub the toe of the prosthetic foot in level walking.

Alignment of the voluntary control four-bar knee

The voluntary control four-bar should be

aligned in full extension with the soft tabs on the extension stop compressed to give contact on the firm portion.

Using the methods of the following paragraph will automatically locate the instant centre properly. This will give the average active patient excellent control of knee stability at heel contact. Figure 9 indicates that such a location will allow the instant centre to remain within the stable region over more than 5 degrees of flexion. The 90 degree rotation centre corresponding to the single-axis or anatomical centre is located at the same level and slightly behind the upper pivot on the anterior link.

Figure 10 shows the preferred method of bench alignment of the socket relative to the knee unit. A vertical reference line is shown drawn upward through the centre line of the shank pylon tubing ending at a point at the bisector of the medial brim (the BMB point) of a quadrilateral socket. The socket is shown flexed approximately 5 degrees. The universal socket adapter has been located on the four-bar mechanism to anticipate the necessary forward

location of the distal portion of the socket. Regardless of the amount of flexion built into the socket, the BMB point must lie on or close to the vertical reference line extending upward through the shank centre line. A forward shift of the socket without a secondary plantar flexion adjustment at the ankle will not give a more stable knee. Some improvement in standing stability may be noted but at the expense of a decrease in stability at heel contact while walking. If the knee is unstable at heel contact the most likely cause is unsatisfactory foot function! The prosthetist must always remember that a forward shift of the socket is perceived by the amputee as a rearward shift of the foot!

The adduction of a socket for a long stump will automatically result in a narrow walking base as shown in Figure 3. The shorter stump will require a wider walking base. In general the axes of the mechanism should be parallel to the floor and perpendicular to the shank tube. The shank tube should be perpendicular to the floor in midstance as viewed anteroposteriorly. The heel stiffness and lever arm have the most

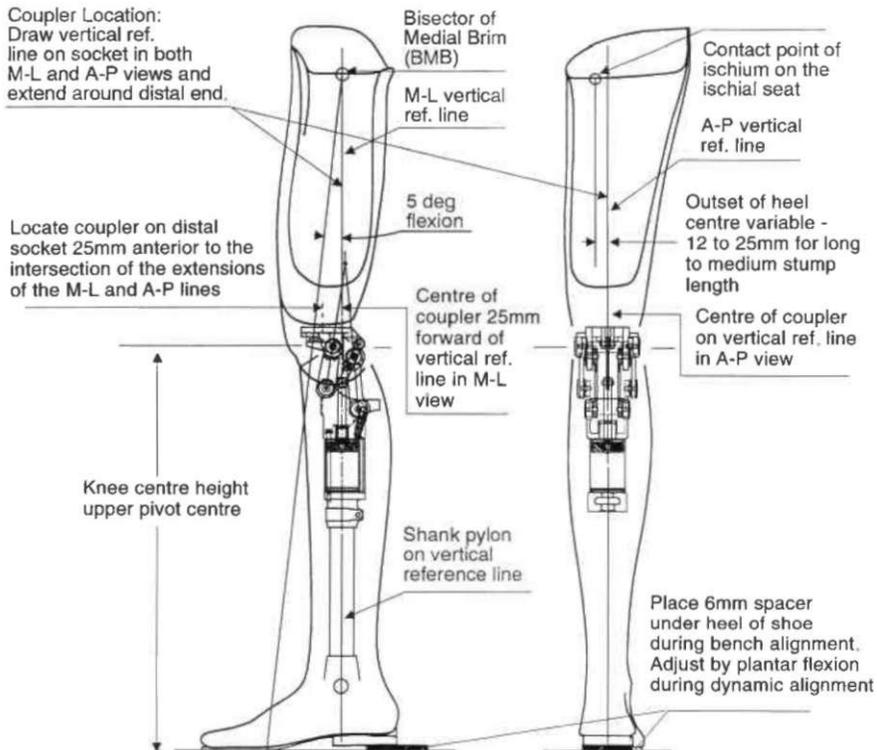


Fig. 10. Bench alignment of the Hosmer voluntary control four-bar knee.

obvious influence on knee stability at heel contact. The knee may feel unstable to the amputee during the period when all weight is supported on the point of the heel. The function of the foot should be to minimize this period and move the support point to the ball of the foot smoothly and rapidly. The heel cushion or plantar flexion stiffness should be as soft as possible without creating foot slap and/or difficulty in rolling over the ball of the foot.

Use of the SACH foot with a voluntary control knee

The SACH foot has been described by some as not being optimal for trans-femoral amputees. This would apply if the foot was fitted with too hard a heel cushion or without air space under the heel. Either error will extend the period of heel contact. It has been the author's experience that a properly selected and fitted SACH foot will provide excellent function when used with a voluntary control four-bar knee.

The heel cushion stiffness must be carefully selected. In general, a trans-femoral amputee must be fitted with a softer grade of heel cushion than a trans-tibial amputee of similar weight and activity level. Many SACH feet, or other similar solid-ankle feet, have been used successfully with trans-femoral amputees.

Some comments on walking training

The walking training of the amputee in the use of a voluntary control four-bar knee must begin with a careful explanation to the amputee of the concept of the instant centre and the principles of voluntary control. The stability of a voluntary control knee is not automatic and the amputee must be instructed carefully in how to participate in the control of knee stability by use of hip musculature. An active amputee with a properly aligned single-axis knee will, most probably, already be using the hip musculature and the initial sensation of the amputee will be that the voluntary control simply makes the control of knee stability a lot easier.

For those amputees who have relied on excessive alignment stability with a single axis or over-stabilized four-bar knee there will be a period of readjustment. The amputee cannot relax the hip at heel contact and rely on the alignment stability to provide knee security. The amputee must be instructed to extend the

hip and press the stump backward in the socket just enough to maintain the knee in full extension against the stop at heel contact. If the stump is not pressed backward the knee will be unstable. The amount of hip moment required is small and the amputee will quickly learn to adopt this very natural method of knee stability control.

It is assumed that the initial training will be done during the dynamic alignment of the prosthesis. If the amputee experiences a feeling of lack of knee stability the foot function and alignment should be checked, then, if necessary, the socket should be extended in about one degree increments and the foot plantar flexed until the knee feels stable. Once the amputee feels secure the training in the use of hip musculature can continue.

Every effort should be made to move the amputee outside the parallel bars as soon as possible. One effective technique for walking training is as follows:

1. Have the trainer providing instruction walk alongside the amputee with the amputee's arm over the shoulder of the trainer to provide support in case the knee buckles.
2. The trainer should use both hands to control the flexion and extension of the socket as the amputee walks. Place one hand on the anterior surface of the socket and the second hand posteriorly. The trainer provides the stability control for the initial walking trials by gently extending and flexing the socket at the appropriate time.
3. As the amputee walks the trainer continues to demonstrate and describe the control action to the amputee. As the amputee becomes more familiar with the necessary hip moments the trainer can gradually reduce moments provided manually and the amputee will continue to walk unaided.

Appendix: The knee stability equation

Figure 11 shows the derivation of an equation, (Eq. 4), which gives the magnitude of the hip extension moment M_h , which would be required to provide knee stability as a function of the axial load P , the magnitude of an existing brake moment M_b , and the x -coordinate (forward offset) and y -coordinate (elevation) of the instant centre at heel contact. Note that a friction brake will typically provide a moment M_b which exceeds the value of P times x hence

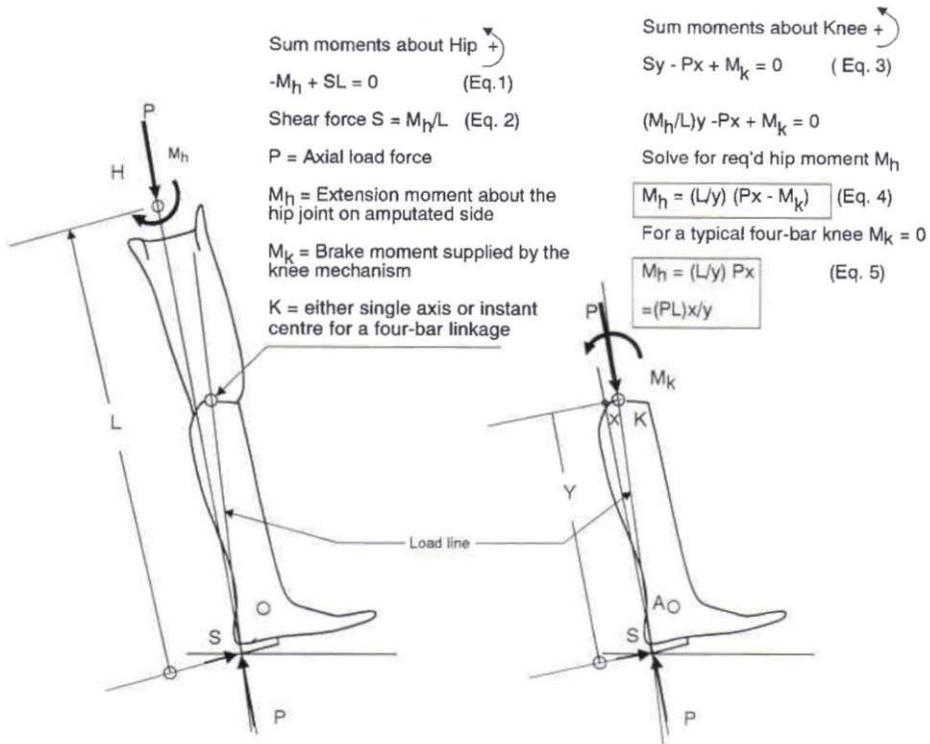


Fig. 11. Derivation of the knee stability equation $M_h = (L/y)(Px - M_k)$

the required hip moment M_h becomes zero.

A four-bar linkage typically does not incorporate a brake mechanism hence the simplified Equation 5 can be used to estimate the hip extension moment required as the position of the instant centre changes.

Both the x and y coordinates of the instant centre are important in the design of a voluntary control four-bar knee linkage. The required hip moment for voluntary control can be reduced in two ways:

1. Reduce the x -coordinate, i.e. locate the initial position of the instant centre closer to the heel-hip line. The x -coordinate of the instant centre should not increase rapidly as a function of knee flexion.
2. Increase the y -coordinate. To gain optimal benefit from the elevation of the instant centre the y -coordinate must not decrease too rapidly as a function of the knee flexion angle.

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