The Biomechanics of Below-Knee Prostheses in Normal, Level, Bipedal Walking

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HUMAN locomotion involves the transformation of a series of controlled and coordinated angular motions occurring simultaneously at the various joints of the lower extremity into a smooth path of motion for the center of gravity of the body as a whole. Though largely taken for granted, it is an extremely complicated process, the complexity becoming evident when one considers that the path of motion is influenced by six major factors: knee-ankle interaction, knee flexion, hip flexion, pelvic rotation about a vertical axis, lateral tilting of the pelvis, and lateral displacement of the pelvis. A thorough study of walking in the orthograde attitude would therefore include not only the influence of each of these factors on the total displacement pattern but also a complete analysis of the action of major muscle groups of the lower extremity. The present discussion is limited to a consideration of the hip, knee, and ankle joints and of their interaction during level walking—first in the normal person and then in the case of the below-knee amputee wearing the patellar-tendon-bearing prosthesis with and without additional impedimenta in the form of thigh corset and sidebars.

PHASES OF THE WALKING CYCLE

The upright, bipedal walking cycle may be divided into two phases—the stance (or weight-bearing) phase and the swing phase. The stance phase of any given leg begins at the instant the heel contacts the ground, ends at toe-off when ground contact is lost by the foot of the same leg. The swing phase begins at toe-off and ends at heel contact. The two feet are in simultaneous contact with the walking surface for approximately 25 percent of a complete two-step cycle, this part of the cycle being designated as the "double-support" phase.

Figure 1 gives a graphic account of the interaction between the knee and ankle joints and of the phasic action of major muscle groups during a typical walking cycle. The particular curves shown represent the average of actual measurements recorded during studies (J) of four male college students considered to be representative of a larger population sample. The sequence of events is arbitrarily started at heel contact and followed until the next heel contact of the same foot. The term "knee moment" refers to the action of the muscle groups about the knee which tends to change the knee angle, either in flexion or extension. Similarly, "ankle moment" refers to the muscular action about the ankle joint which may cause either plantar flexion or dorsiflexion, The mechanics of major muscle groups of the lower extremity is indicated in Figure 2.

EVENTS JUST PRIOR TO HEEL CONTACT

In reference to Figure 1, and particularly to the curves in the region corresponding to the end of the swing phase (about 95 percent of a complete cycle), it may be noted that the knee joint reaches its maximum extension just prior to heel contact and that a period of knee flexion then initiated continues on into the stance phase. As seen in the curves of muscle activity, this decrease in the rate of knee extension at the end of the swing phase, in

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preparation for the contact of the foot with the floor, is due primarily to the action of the hamstring muscle group, which is attached to the pelvis behind the hip joint and to the tibia and fibula below the knee joint. Tension in the hamstring group may cause either hip extension or knee flexion or the two simultaneously.

HEEL-CONTACT PHASE

As the heel makes contact, the hamstring action tends to bring it forcibly backward into contact with the floor, while the knee continues to flex rapidly. The activity in the hamstring group continues, but with decreasing magnitude, while the quadriceps action begins to build up quickly. The quadriceps group, acting anteriorly about the knee joint, and the pretibial group, acting about the ankle joint, serve to control the knee-ankle interaction and thus to effect a smooth motion of the forepart of the foot toward the floor. The major function of both knee and ankle during this phase is smooth absorption of the shock of heel contact and maintenance of a smooth path of the center of gravity of the whole body. Although the function of the knee as a shock absorber is often overlooked, energy studies (1) have shown that the knee and ankle contribute equally to shock absorption.

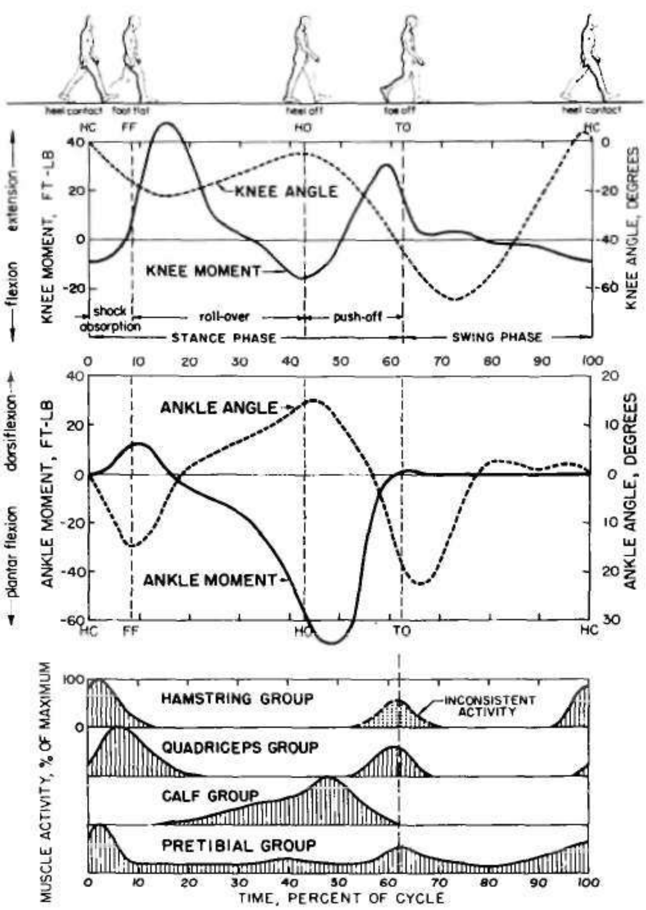

Fig. 1. Correlation between joint action and muscular activity in the lower extremity during normal, level walking.

MID-STANCE PHASE

The controlled knee flexion of the heel-contact phase continues into the mid-stance phase (between foot flat and heel-off), and the maximum angle of knee flexion, approximately 20 deg., occurs in the first part of the mid-stance phase. As the body rides over the stabilized knee, the upward thrust of the floor reaction moves forward on the sole of the foot, thus gradually increasing the dorsiflexion of the ankle and causing the knee to begin a period of extension. In this period, control of the leg is through ankle-knee interaction, there being only minimal muscular activity in the groups acting about the hip and knee. The knee reaches a position of maximum extension about the time the heel leaves the ground, the calf group providing the resistance to knee extension and ankle dorsiflexion. As the heel leaves the ground, the knee again begins a period of flexion, chiefly because of muscular action about the hip joint. This sequence of controlled flexion at heel contact, release to allow gradual extension in mid-stance, and controlled flexion preparatory to swing is important in accomplishing a smooth and energy-saving gait in normal persons.

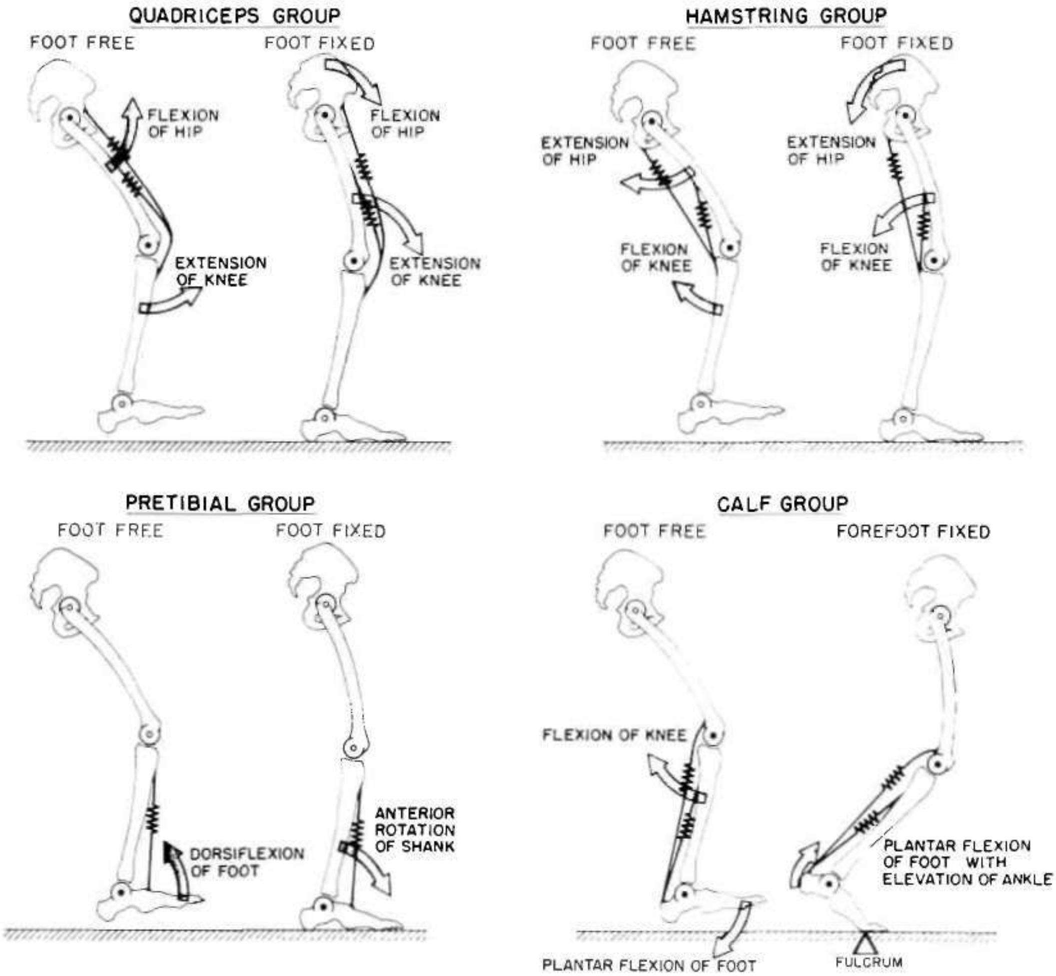

Fig. 2. Major muscle groups of the normal lower extremity (schematic), showing the major mechanics in the parasagittal plane.

PUSH-OFF PHASE

During the push-off phase, a phase complex and often misunderstood, the knee is brought forward by action of the hip joint, and a sensitive balance is maintained by interaction of hip, knee, and ankle joints. The combined action has two purposes—to maintain the smooth forward progression of the body as a whole and to initiate the angular movements in the swing phase that follows. As the knee begins to flex (about the time the heel leaves the ground), the knee musculature must first resist the external effect of the force on the ball of the foot which passes through space on a line ahead of the knee joint. Then, as the knee is brought forward by hip-joint action, so as to pass through and then anterior to the line of the force acting upward on the foot, the knee must reverse its action to provide controlled resistance to flexion by increasing quadriceps activity. Some inconsistent hamstring activity is noted as an antagonist. The calf group continues to provide active plantar flexion during the entire push-off phase. At the time the toe leaves the floor, the knee has flexed 40 to 45 deg. of the maximum of 65 deg. it reaches during the swing phase. In normal persons, knee flexion in the swing phase is not due primarily to hamstring action, as might be supposed. Complete prosthetic restoration of normal function in the push-off phase is difficult, if not impossible. A proprioceptive sense of knee position by the amputee is necessary, as well as an active source of energy in the ankle. Because of lack of an active source of ankle energy, initiation of knee flexion in amputees wearing a prosthesis must come from active hip flexion.

SWING PHASE (QUADRICEPS ACTION)

The over-all objective in the swing phase is to get the foot from one position to the next in a smooth manner while clearing the usual obstacles of terrain. At the start of the swing phase, the leg has just completed a period of rapid increase in kinetic energy caused by the active extension of the ankle and flexion of the hip during the push-off phase. The knee is flexing and continues to flex after toe-off. During rapid walking this would result in excessive knee flexion and heel rise were it not for the action of the quadriceps group in limiting the angle of knee flexion to approximately 65 deg. and then continuing to act to start knee extension. Knee extension continues as a result of a combination of pendulum effects owing both to muscle action and to the weight of the inclined shank and of the foot. Little quadriceps action is required, since other factors are of equal importance. For example, the iliopsoas muscle contributes by developing active hip flexion, which in turn accelerates the knee forward and upward.

MID-SWING

During mid-swing there is a period of minimal muscular activity, and the leg accelerates downward and forward like a pendulum with forced motion of its pivot point.

TERMINAL DECELERATION (HAMSTRING ACTION)

Near the end of the swing phase, the rate of knee extension must be reduced in order to decelerate the foot prior to heel contact. This "terminal deceleration" of the normal leg is due primarily to the extension resistance of the hamstring group.

KNEE ACTION IN AMPUTEE GAIT

In the past a common cause of difficulty in the use of the so-called "muley" below-knee prostheses (2) has been the "breakdown" of the stump, in particular of the knee joint on the amputated side. It has been due in part to overstraining of the ligamentous structures of the knee by excessive hyperextension under load. In order to protect these ligamentous structures on the amputated side, it is necessary to maintain within safe limits the forces and moments about the knee which tend to force it into hyperextension. In normal individuals a precise sense of knee position limits the hyperextension moment by maintaining the knee center close to the line of the force transmitted through the lower extremity. Since in many below-knee amputees the knee action is unaffected by amputation, it is reasonable to expect such an amputee to walk with a normal knee action. When this potential is anticipated and accounted for in the fitting and alignment procedure, a below-knee amputee of average-to-long slump length can make use of the controlled flexion-extension-flexion sequence of knee action required in absorbing shock and smoothing the path of motion of the center of gravity (Fig. 1). The socket must be fitted to accommodate the dynamic forces, and the amputee must contribute voluntary control of the knee by action of the musculature.

ANALYSIS OF STUMP-SOCKET FORCES

The contact pressures between the stump and socket of a below-knee amputee are influenced by a combination of factors. In the case of the patellar-tendon-bearing prosthesis (or of any other below-knee prosthesis without thigh corset and sidebars), the two major factors are the fit of the socket and the alignment of the prosthesis, *i.e.*, the location of the foot with respect to the socket. When the thigh corset is used, there are certain modifying effects even when optimum alignment of sidebars and corset with respect to the socket is obtained. In discussing the relationship between fit and alignment, it is often helpful to discuss alignment factors first, since the method of fitting a socket to an amputee's stump is dictated largely by the manner in which he

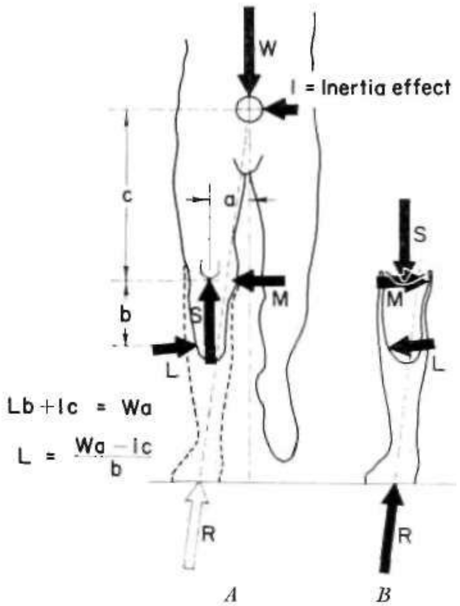

Fig. 3. Mediolateral force diagram for a below-knee amputee wearing the patellar-tendon-bearing prosthesis with supracondylar cuff only. *A*, Forces on the amputee; *B*, forces on the prosthesis.

can be expected to perform while wearing his prosthesis. His performance, in turn, is influenced considerably by the structural relationship between the elements of his prosthesis, *i.e.*, the alignment. The patellar-tendon-bearing cuff-suspension below-knee prosthesis, without side joints or corset, is here discussed first. Thereafter the modifying influences resulting from the addition of the side joints and corset are considered.

The following analysis is based on the assumption that a below-knee amputee with a stump of at least average length can be expected to walk in a manner similar to that of a normal person. That is, if the prosthetic foot is properly designed to minimize the effects of the loss of normal ankle function, the amputee can compensate by hip and knee action so as to achieve a gait which closely approximates the normal. Accordingly, he should be expected to go through the following sequence of knee motions:

1. Control of knee flexion from the time of heel contact until the foot reaches a stable position flat on the floor.

2. Control of knee flexion-extension during roll-over. The foot-shank serves as a firm base during this portion of the stance phase. The position of the knee relative to the force acting on the foot can be gauged accurately by properly trained amputees. The muscular moment about the knee required to maintain a particular knee position serves as an excellent source of proprioceptive sensation if the socket fit is intimate enough to reduce lost motion to a minimum.

3. Control of knee flexion during the push-off phase as an aid in accelerating the prosthesis forward in the swing phase.

MEDIOLATERAL FORCES, CUFF-SUSPENSION BELOW-KNEE PROSTHESIS

Figure 3 is a front view of a below-knee amputee in a position corresponding to the mid-stance phase. Two force systems are shown. Figure *3A* shows the forces exerted on the amputee. These forces are of two types—the body weight due to the effect of the earth's gravitational pull and the forces applied through contact with the socket. Figure 3B shows the forces acting on the prosthesis.

If, as seen from the front, the prosthesis is considered as a means of supporting the body, it must be capable of providing both vertical support and mediolateral balance. It is apparent that vertical components of pressure are applied against the surfaces of many areas of the stump, but for purposes of simplified analysis the combined effect of all these forces is shown as the single support force *S*.

Considering the point of application of the support force 5 as a balance point, the lateral force *L* times the distance *b* equals the body weight *W* times the distance *a*, or, in equation form:

$$Lb = Wa \quad \text{and} \quad L = \frac{Wa}{b}$$

Unfortunately, the effect of the horizontal acceleration of the center of gravity cannot be ignored in this case, and hence in neglecting the horizontal acceleration equation 1 is incorrect.

As indicated in Figure 3, the horizontal acceleration of the body in a medial direction, due to the medial inclination of the total floor reaction *R*, results in a lateral inertia force which tends to oppose the acceleration. This inertia force must be included when consideration is given to balancing moments about the

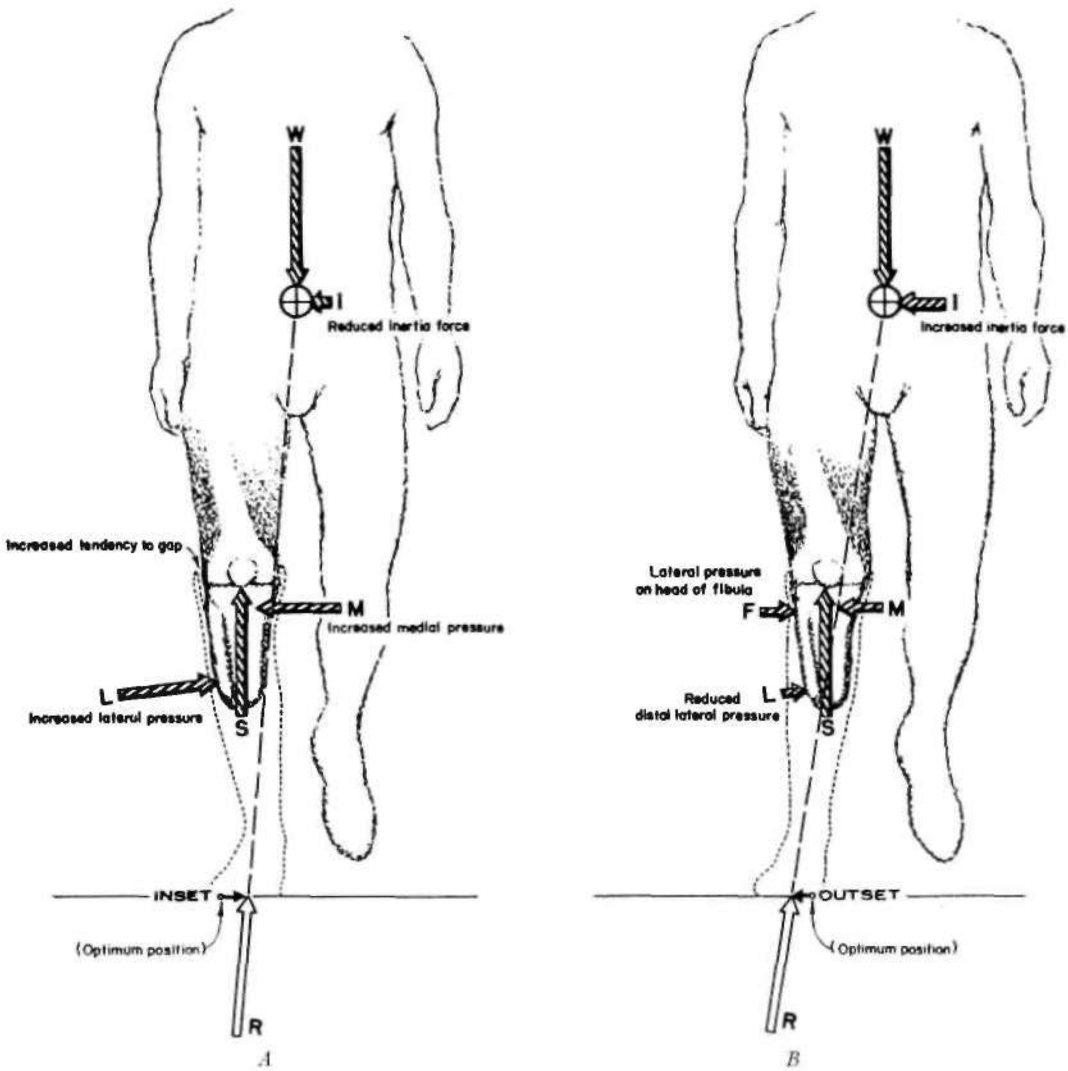

Fig. 4. Change in mediolateral force diagram owing to inset or outset of foot from optimum position, PTB prosthesis with cuff only, as in Figure 3. *A*, Inset; *B*, outset.

point of support. The correct relationship is therefore $Lb + Ic = Wa$:

$$L = \frac{Wa - Ic}{b}$$

Equation 2 shows that the magnitude of the required lateral stabilizing (balancing) force L can be reduced in one of two ways—by increasing the horizontal inertia force or by increasing the effective lever arm b. Increasing the horizontal inertia force requires that the horizontal acceleration be increased or, in other words, that the foot should be moved laterally so as to increase the medial inclination of the total floor reaction.

EFFECT OF FOOT INSET-OUTSET ON MEDIOLATERAL FORCES

The effect of changing the inset or outset of the foot is shown in Figure 4, where it is possible under special conditions, as shown in Figure 4B, to eliminate the need for the lateral

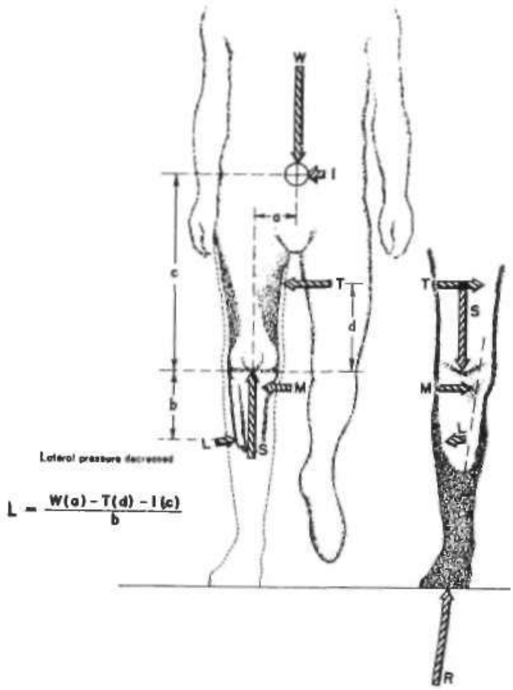

Fig. 5. Effect of thigh corset and sidebars on mediolateral stump-socket forces, PTB prosthesis. When the thigh corset applies a force against the medial side of the upper part of the thigh, the effect is similar to a force on the laterodistal side of the stump. Corset adjustment constitutes a possible means of modifying the magnitude and distribution of forces against the lateral side of the stump. This circumstance suggests that if the lateral sidebar is constructed with sufficient stiffness it may be of assistance in relieving excessive pressure on the laterodistal end of the stump.

stabilization force *L*, since in this case the weight and inertia force are seen to be in balance:

$$Wa = Ic$$

The force on the lateral aspect of the stump has shifted to the region of the head of the fibula.

Complete elimination of the lateral stabilizing force *L* by outset of the foot is generally undesirable, for the resulting wide-based gait is abnormal and unnecessary. Actually, a narrow-based gait with a definite need for the lateral force *L* (and corresponding lack of pressure on the head of the fibula) is definitely indicated for stumps 4 in. or more in length, the wide-based alignment being then reserved for very short below-knee stumps. It must be remembered, however, that planning the fit and alignment of a below-knee prosthesis to accommodate a narrow-based gait requires that the need for a definite lateral stabilizing force be recognized and accounted for in the fitting of the socket.

EFFECT OF THIGH CORSET AND SIDEBARS ON MEDIOLATERAL FORCES

Figure 5 shows the modifying effect of the thigh corset and sidebars on the pressures between stump and socket. If the sidebars are stiff enough it is possible to develop against the medial thigh a force *T* which acts in cooperation with the lateral-distal socket contact force *L* in providing mediolateral stabilization. In fact, with judicious use of bending irons the lateral pressure can be greatly reduced. In the past, this has been done to compensate for uncomfortable lateral-distal stump pressure. With a good socket fit against the lateral aspect of average-length stumps, however, the need for lateral stabilization by the thigh corset is minimized. Use of a thigh corset is indicated only for amputees with very short stumps or those in whom other medical factors require reduction in stump-socket contact forces.

ANTEROPOSTERIOR FORCES, CUFF-SUSPENSION BELOW-KNEE PROSTHESIS

Figure 6 shows a side view of a below-knee amputee and the cuff-suspension prosthesis under three conditions—at heel contact, during the shock-absorption portion of the mid-stance phase, and during push-off. At the instant of heel contact, and for a short time corresponding to about 5 percent of the walking cycle, knee stability is maintained primarily by active extension of the hip joint. The tendency of the external load on the prosthesis to extend the knee is resisted by hamstring action. During this phase, forces are acting as shown in Figure *6A*.

Analysis of the forces acting during the shock-absorption portion of the mid-stance phase shows that it is typical for the floor-reaction force *R* to be acting along a line which passes posterior to the knee center. Under such circumstances, a completely relaxed knee would buckle, but the amputee is able to resist this

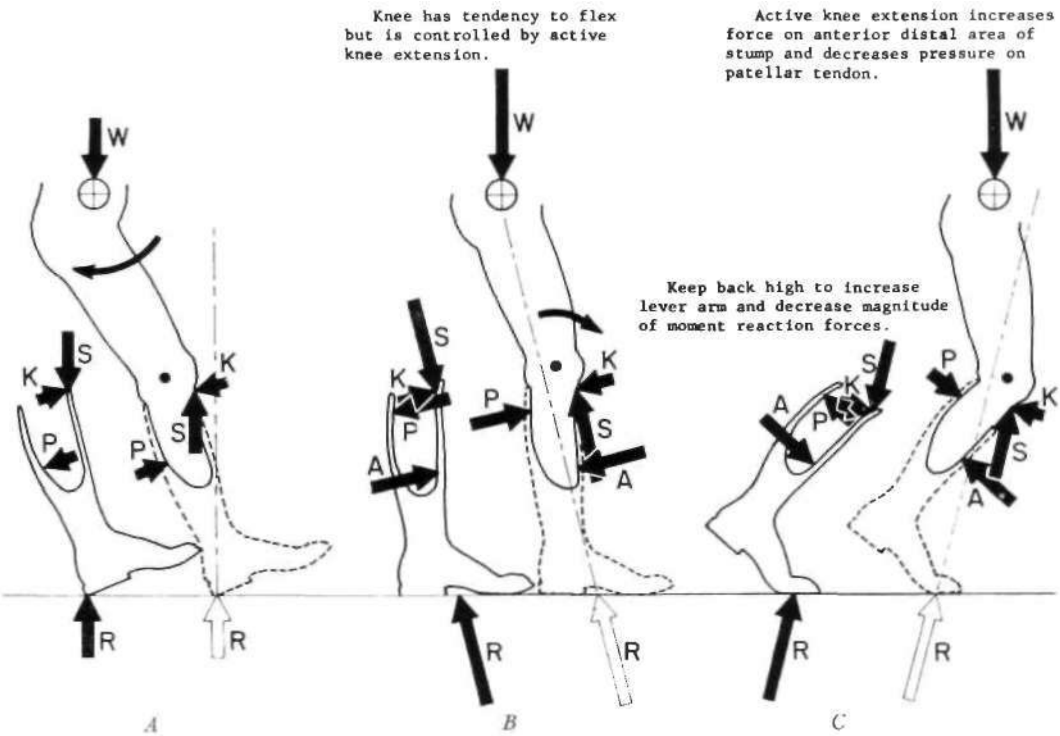

Fig. 6. Anteroposterior force diagrams for a below-knee amputee wearing the patellar-tendon-bearing prothesis -with supracondylar cuff only. *A,* At heel contact; *B,* during shock absorption (foot flat in midstance); *C,* during push-off.

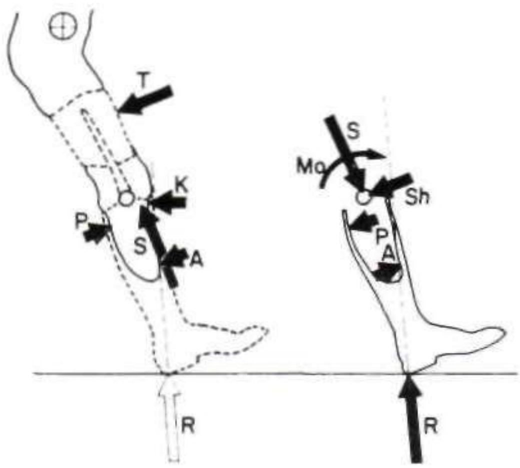

Fig. 7. Effect of thigh corset, sidebars, and back-check on anteroposterior stump-socket forces, PTB prosthesis. Shear force, *Sh,* is absorbed by mechanical side joint. Moment reaction forces on the stump are reduced through absorption of moment by knee stop. Without a knee stop, the stump would have to resist moment due to floor reaction passing ahead of knee joint. The resulting high pressure on the patellar tendon can be eliminated if the knee is allowed to flex (Fig. 6) instead of being forced into full extension.

tendency by active knee extension. The resulting force pattern on the stump (disregarding end-bearing) is as shown in Figure *6B,* where the forces are concentrated in three areas—around the patellar tendon, on the anterodistal portion of the tibia, and in the popliteal area. The socket fit must be designed to accommodate the resulting functional pressures.

During the push-off phase, the floor reaction continues to pass behind the knee, and the anteroposterior forces are concentrated in the same three areas, as shown in Figure *6C.*

EFFECT OF THIGH CORSET AND SIDEBARS ON ANTEROPOSTERIOR FORCES

If a below-knee amputee is fitted with a thigh corset and back-check so that he relies on the mechanical action of the back-check to resist knee extension, the force pattern is altered considerably. Figure 7 shows the effect. The floor reaction *R* must now be assumed to pass anterior to the knee, since otherwise the knee would not be extended against the back-check. If the knee joint is considered as a moment center, the effect of the force *R* is resisted by the back-check moment *Mo* and the two forces *A* and *P* exerted by the stump within the socket. Under the proper conditions, it is possible for the mechanical back-check to provide the total resistance to the floor reaction, the stump being suspended freely in the socket. This would indicate that, by proper adjustment of thigh corset, sidebars, and back-check, it is possible to modify the pattern of anteroposterior stump-socket contact pressures.

SUMMARY

Thus it may be seen that, while normal skeletal and neuromuscular structure of the lower extremity is so organized as to accommodate the complex and precisely phased performance needed for erect, bipedal locomotion, the below-knee amputee, even though provided with a well-fitting prosthesis of the patellar-tendon-bearing cuff-suspension type, is unavoidably destined to experience in walking a continually changing set of stump-socket forces in both the anteroposterior and the mediolateral directions. Successful fitting of the below-knee amputee means, therefore, the resolution of stump-socket forces in such a way as to provide both comfortable support and adequate stabilization throughout the walking cycle. Whenever addition of thigh corset and sidebars is required, there occurs a change in the pattern of motion, and hence a change in stump-socket forces to be anticipated, and accordingly suitable modifications are required. Allowance for such factors calls in every case for the sound judgment of the prosthetist if fully satisfactory results are to be obtained.

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