

The Biomechanics of the Syme Prosthesis¹

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The purpose of any limb prosthesis is to replace, to the reasonable satisfaction of the wearer, as much as possible of the normal form and function lost through amputation. To provide a suitable prosthesis in any particular case, therefore, the several cooperating professional persons—physicians, prosthetists, therapists, others as appropriate—must have an intimate knowledge of just what losses have been incurred and just what new circumstances, if any, have accrued as a result of the losses. Among these are the losses of structural elements, of joint motion, and of muscle function; the decrease in proprioceptive sense as well as in sensory perception; the development of persistent or recurrent pain in one form or another; the impairment of circulation; and the losses of what in the normal would be the weight-bearing areas; not to mention numerous other matters purely medical and not necessarily associated with the amputation. Any one of these factors, or any combination of them, may influence the way in which an amputee will use a given type of limb prosthesis—that is, a device intended as a limb substitute.

In the case of the Syme amputee, where the patient has suffered loss of the foot and ankle while retaining essentially the full length of the shank and more or less of the typical weight-bearing characteristics of the normal heel, the obvious problem is to restore foot and ankle function (or to supply the equivalent of foot-ankle function), to extend the stump so as to accommodate the loss of the tarsus and of the calcaneus, to furnish adequate support for the

body during standing and during the stance phase of walking, to provide suitable suspension for the prosthesis during the swing phase, and to do all these things in such a way that the final result is acceptable to the wearer under both static and dynamic conditions. As with prostheses for other levels of amputation in the lower extremity, determination of the requirements of the Syme prosthesis takes its departure from a review of the normal pattern of locomotion and proceeds toward assessment of the means through which such a pattern may best be reproduced by application of inanimate devices. Discussion is here limited to the pertinent features of straight and level walking in the normal person and to the corresponding circumstances in a Syme amputee enjoying good general health, using a prosthesis, and having a stump itself free from any inherent medical complications such as excessive scar tissue, or neuromas, or skin disorders, or sensitive joints, or other conditions ordinarily beyond control of the limb designer.

LOCOMOTION PATTERNS

In any analysis of bipedal locomotion such as that of man, it is common practice to divide the walking cycle into the two obvious phases through which the lower limbs pass alternately—the stance phase and the swing phase. Figures 1 and 2, based on averages from tests on four normal young males during straight and level walking (1,3), show five different kinds of data—angular motion at the knee and ankle joints, moments about the knee and ankle joints as a result of muscle activity, muscle activity as measured by electromyographic techniques, energy level at the knee and ankle joints at a given instant, and change in energy level. Correlation of the energy data (Fig. 2) with motions of the joints (Fig. 1) provides an insight into knee-ankle interaction

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in normal human locomotion and is useful in determining the compensation required to make up for the losses incurred by Syme's amputation.

The terms "work done on/" "work done by," "input," and "output" used in describing energy requirements can best be defined by citing examples. In the simplified sketch of musculoskeletal joint action (Fig. 3), the musculature exerts an internal moment M which resists the load W . If the load W is sufficient to overcome the moment M and thus to cause the joint to rotate in opposition to the muscle action, then work is done *on* the joint, *i.e.*, the joint absorbs energy. If the moment M is sufficient to cause the joint to rotate in the same direction as the muscle action and thus to move the load W in a direction opposite to its sense, then work is done *by* the joint, *i.e.*, the joint provides an energy output.

THE STANCE PHASE

Comparison of the stance phase of the normal with that of the Syme amputee wearing a prosthesis reveals an excellent example of compensation by one joint (the knee) for loss of a second joint (the ankle) in the same extremity.

Shock Absorption

During the subphase designated "shock absorption" (Figs. 1 and 2), the ankle in the normal subject undergoes plantar flexion while the knee flexes, both under load. Thus, an energy input results at both knee and ankle (work is done *on* both joints during the first part of the stance phase). As summarized in the bar graph of Figure 2, the work done on one joint is approximately equal to that done on the other. It could therefore be stated that in bipedal walking the knee and ankle contribute equally to the cushioning of the shock transmitted to the body at the beginning of

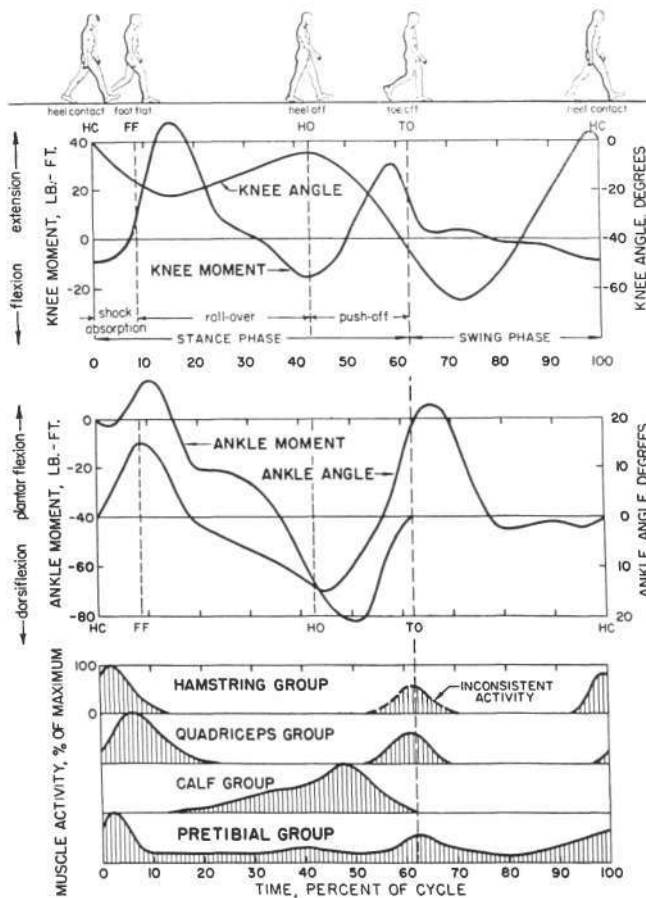


Fig. 1. Correlation between joint action and muscular activity in normal locomotion in man.

the stance phase when the leg first assumes its function of support.

In the Syme amputee, ankle function has been lost and some way of compensating for it must be found. Because of the inherent space limitations in conventional Syme prostheses, use of articulated ankle joints and elastic compression members has been for the most part unsuccessful. It is known that, in order to keep stresses in elastic bumpers within reasonable limits, the bumpers must contain a certain minimum volume of material. Otherwise the energy-absorption requirements per unit volume are excessive, and overheating and fatigue occur rapidly. The alternatives are to increase the volume of shock-absorbing material so as to reduce the unit stresses, or to

transfer shock absorption to some other area, or both.

The volume of shock-absorbing material can be increased by eliminating the articulated ankle joint and using in the heel the greatest possible volume of suitable sponge-rubber cushion—as in the SACH foot (2). In general, function may be improved over that supplied by an articulated joint, but owing to the space limitations the Syme amputee cannot be given the same degree of shock absorption as can be afforded the above-knee or below-knee amputee wearing a SACH foot.

To compensate for the lack of adequate function in the artificial foot, the knee joint on the side of the amputation must assume a greater proportion of shock absorption by increasing

the amount of knee flexion under load just after heel contact. If the knee does not assume this function, the amputee must tolerate a definite impact force from prosthesis to stump and must also accept the deviation from normal gait that might be expected to accompany such a circumstance.

Roll-Over

The roll-over portion of the stance phase in normals may in turn be subdivided into three parts corresponding to the direction of knee motion. During the first part, the knee continues to flex under load and thus prolongs the period of its function as a shock absorber for the initial support of the body weight. The ankle, acting as a controller, is required to supply energy during this time, as indicated by the rising curve of energy level and the positive bar for the ankle (Fig. 2). In the Syme amputee, the heel cushion of the modified SACH foot contributes some of its energy of compression and thereby simulates normal ankle action, but again the knee joint must compensate for the shortcomings of the prosthetic foot-ankle unit. Because of the lack of active plantar flexion in Syme amputees, maximum knee flexion during this subphase is in general less in persons wearing a Syme prosthesis than it is in normal persons.

While in normal locomotion the body continues to roll over the foot, which for the time being continues in full contact with the floor, the knee begins a second period of active extension, a circumstance that results in work being done on the body as a whole (*i.e.*, the knee exhibits energy output). Meanwhile, the ankle absorbs about half the energy output of the knee. In a typical Syme amputee wearing a prosthesis, the foot-ankle unit is neither absorbing nor supplying energy during this period, and the energy requirement of the knee during this interval is thus reduced as compared with that of the normal person.

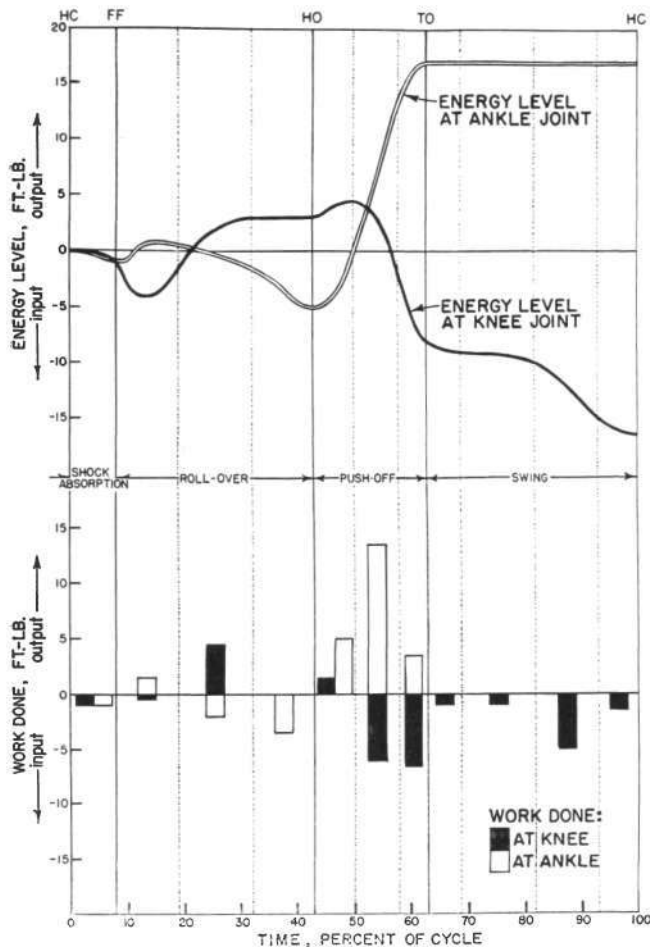


Fig. 2. Energy levels and work done at knee joint and ankle joint during normal, level walking.

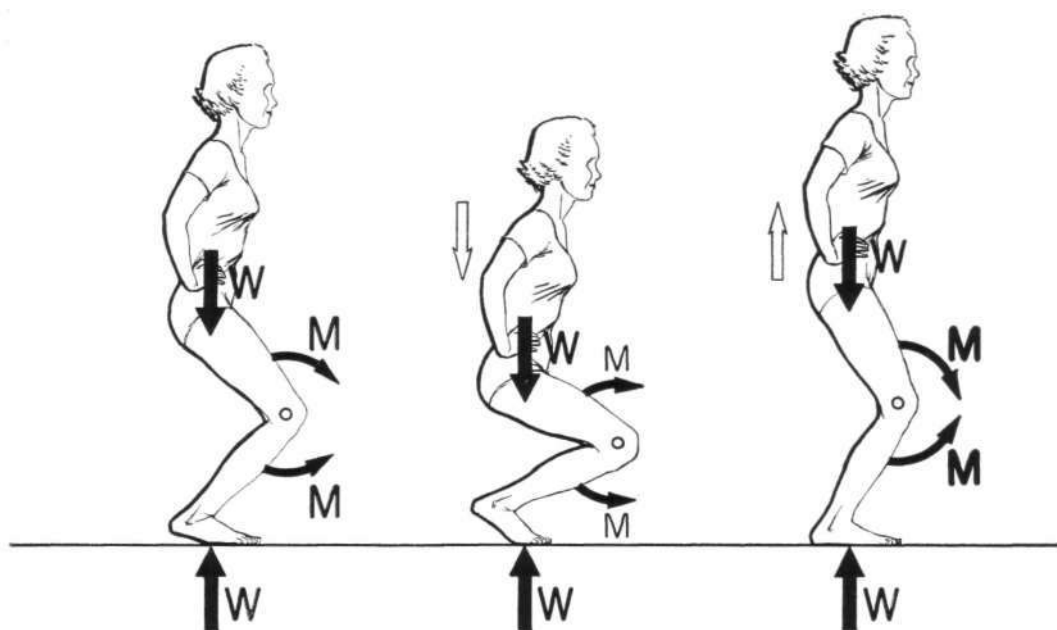


Fig. 3. Energy input and output at a typical joint. Left, equilibrium; center, energy in at knee joint, *i.e.*, work done *on* the joint; right, energy out at knee joint, *i.e.*, work done *by* the joint.

During the third part of normal roll-over, the knee is forced into full extension and maintained there by the external forces acting upward on the ball of the foot. The ankle continues to absorb energy as the tibia rotates forward over the stationary foot. To compensate for the inability of the prosthetic ankle to absorb energy during the last part of roll-over, the prosthetic foot must be designed so that the forward point of support corresponds to the ball of the foot, an arrangement which maintains the knee along a path corresponding to that of the normal. In other words, the knee should move forward smoothly, and no sensation of vaulting over the fore part of the foot should be experienced. In the amputee wearing a Syme prosthesis with a properly aligned SACH foot, knee action at the end of roll-over should be almost the same as it is in a normal person.

Push-Off

The push-off portion of the stance phase begins when the heel is lifted from the floor. During the first part of this subphase in normal persons, both knee and ankle contribute energy—the knee by virtue of energy that has been stored by passive stretching of the ham-

string ligaments and the ankle by virtue of active plantar flexion which continues throughout the push-off phase. In the Syme amputee, the ankle substitute cannot contribute energy by active plantar flexion, and accordingly other means must be found to maintain a smooth path of the center of gravity of the body. In the SACH foot, a comparatively simple keel contour, with a cylindrical or spherical surface on a 2-in. radius at the end of the keel, has been found practical for most adults. Under these circumstances, the hip and knee joints serve as the active elements in the kinematic chain which controls the pathway of the center of gravity.

In the second part of push-off, the normal knee absorbs about half as much energy as is supplied by the normal ankle joint, energy absorption by the knee being associated with the maintenance of a smooth path for the center of gravity of the body as a whole. At toe-off, for example, the knee in normal persons has flexed 40 deg. of the total of 65 deg. achieved at the point of maximum knee flexion. Energy absorption by the normal knee continues at about the same rate after active plantar flexion of the ankle has started to slow down. Since the foot-ankle unit in the Syme prosthesis must

maintain the pathway of the knee by proper keel contour rather than by active plantar flexion of the ankle, the amount of energy absorption required of the knee is less in the Syme than it is in the normal. The need to initiate knee flexion before the end of the stance phase remains, however, and the socket must therefore be designed to permit maximum control of knee motion by the stump in preparation for the swing phase.

THE SWING PHASE

Since in the patient with Syme's amputation the knee and hip joints are usually undisturbed, it might be assumed that the swing phase of the Syme amputee would always appear relatively normal. But the role of the ankle joint at the end of the stance phase must be considered. In normal locomotion, the knee starts to flex before the foot leaves the ground, and the controlled knee-ankle interaction provides a major source of energy for the forward propulsion of the knee. If this motion is smooth and precisely controlled, the thigh-shank-foot combination enters the swing phase normally. Anything that tends to disturb this smooth transition from stance to swing has a noticeable effect throughout the swing phase.

For the patient who has undergone Syme's amputation, poor function in the prosthetic foot and pain in the weight-bearing areas of the stump are the two most common sources of unstable or erratic action during transition from stance to swing phase. When, however, the prosthetic foot has been properly designed, aligned, and adjusted to allow the knee and hip to provide normal-appearing control of knee motion at the end of the stance phase, the amputee should, in general, have the ability to exercise complete control of his prosthesis during swing phase.

SOCKET DESIGN

ANALYSIS OF STUMP-SOCKET FORCES DURING THE STANCE PHASE

Analysis of the distribution of contact pressures between stump and socket at various times during the stance phase is useful in the design of a socket that will be comfortable for the amputee. Since pressure distribution varies

during each of the three subphases—shock absorption, roll-over, and push-off—each must be analyzed separately.

Shock Absorption

If it be assumed that body weight is supported at the distal end of the stump, it can be seen clearly from Figure 4A that during the shock-absorption subphase the major functional forces between stump and socket occur in the anterodistal and posteroproximal areas. During roll-over, the need for posteroproximal pressure decreases, and the contact pressure at the end of the stump shifts toward the center of that area. If the force system is to be in equilibrium, the paths of the forces P , D , and F must intersect at M and their vectors must form a closed polygon. Use of this principle makes it possible to estimate the relative magnitudes of the three forces.

Push-Off

Figure 4B shows the force system that develops as the Syme amputee rolls over the ball of the foot in the push-off subphase. At the instant shown, the hip joint is being used to help flex the knee against the force acting upward on the ball of the foot. Again, the principle of force equilibrium can be applied to estimate the magnitude of the forces. A posterodistal and an anteroproximal contact force between stump and socket are seen to be necessary to resist the floor reaction against the ball of the foot. It is essential that the anteroproximal force against the tibia be kept at as high a level as possible. Shortening of the distance a results in increased inclination of the line of the posterodistal contact force and in a transfer of the force away from areas surgically prepared for end-bearing.

Since some change in the inclination of the distal stump-socket force is unavoidable, it must be anticipated during the fitting procedure. If the line of the floor reaction is kept in a particular position relative to the knee, the amputee can use some voluntary control in shifting the distal contact point. Moreover, the anteroproximal force at push-off will be several times the posteroproximal force at heel contact. For this reason, the prosthesis must be

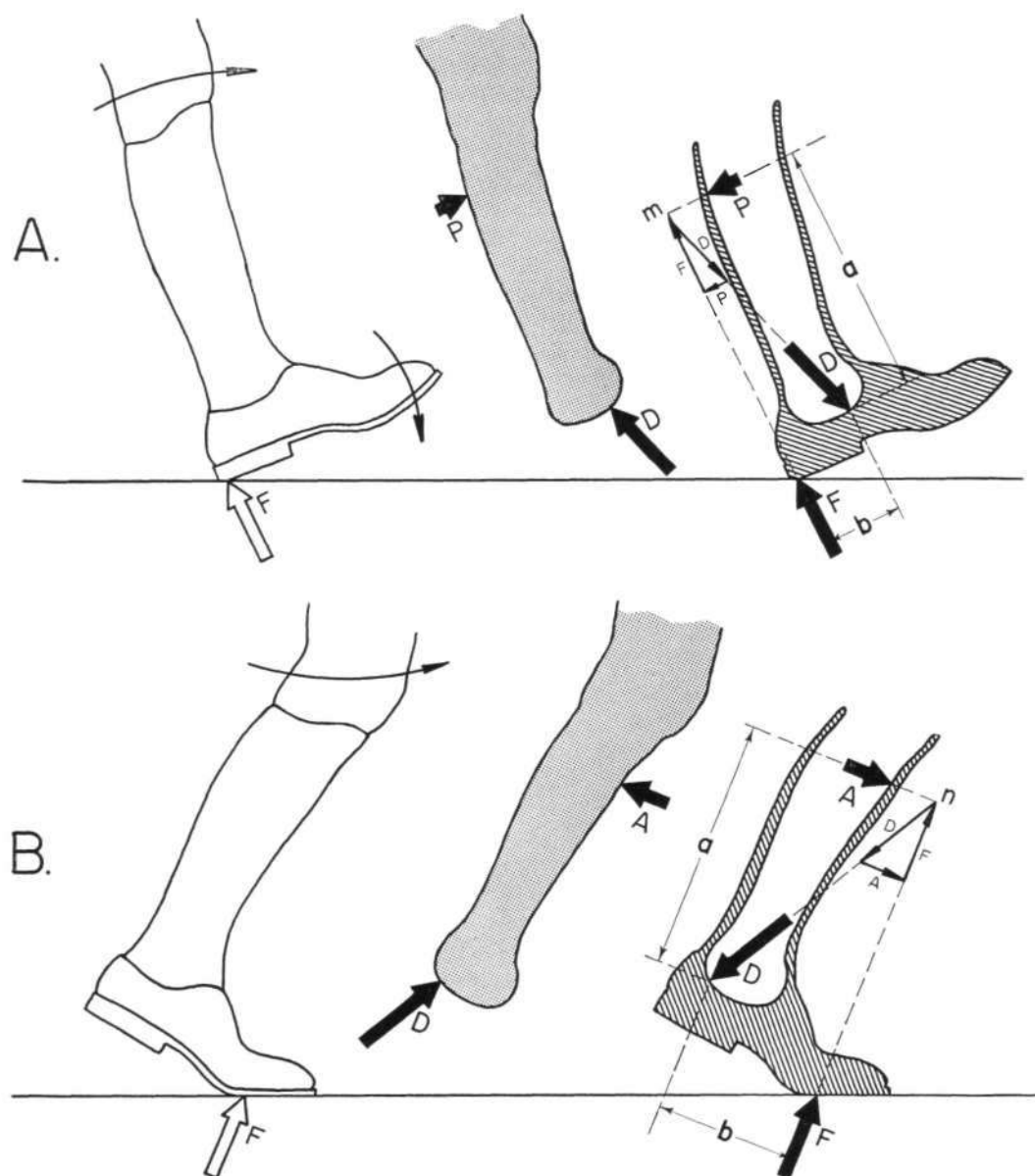


Fig. 4. Stump-socket forces during the stance phase. A, Shock absorption; B, push-off.

strong enough to resist the large bending moment in the ankle region during push-off. Suppose that in a 180-lb. man there is an increase of 30 percent (as compared with body weight) in the dynamic force against the ball of the foot during push-off and that dimension b is 4 in. Then the structure must resist a bending moment of $1.30 \times 180 \times 4 = 936$ lb.-in.

SOCKET MATERIALS

Because of the bulbous form of the typical Syme stump, any prosthesis devised for it will be bulky in appearance. To provide the least bulky socket requires that the thickness of the wall be kept to a minimum commensurate with structural demands. Plastic laminates

with high strength-weight ratios that can be molded easily over a plaster model seem ideally suited for construction of sockets for the Syme prosthesis.

Since a snug fit throughout the length of the stump is necessary if proper function is to be expected, a cutout must be provided in the narrow section of the socket to permit entry of the bulbous end of the stump. The question arises as to where to locate a cutout, which in any case obviously should not interfere with the functional characteristics of the prosthesis nor affect its structural properties unduly. Several possibilities have been suggested. Among others are the posterior cutout used at Sunnybrook Hospital in Toronto and the medial cutout proposed at the Veterans Administration Prosthetics Center (page 57). Some predictions as to the relative structural strengths to be had from the several approaches may be arrived at through the techniques of engineering stress analysis.

From a review of data on normal human locomotion it has been determined that in level walking maximum forces are brought to bear on the shank at the time of push-off. At this point in the walking cycle the center of pressure is eccentric with respect to the shank.

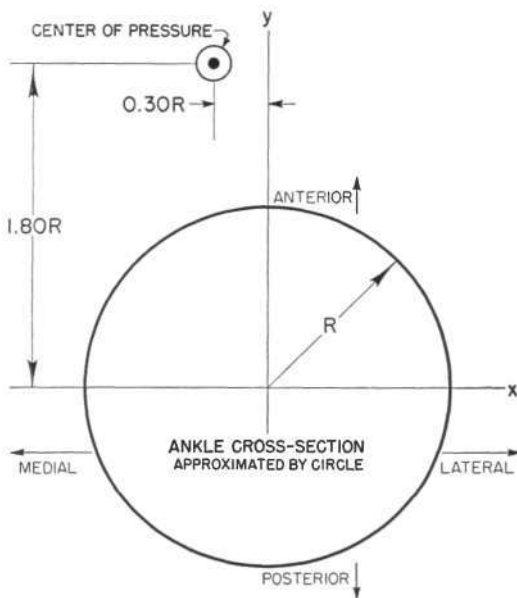
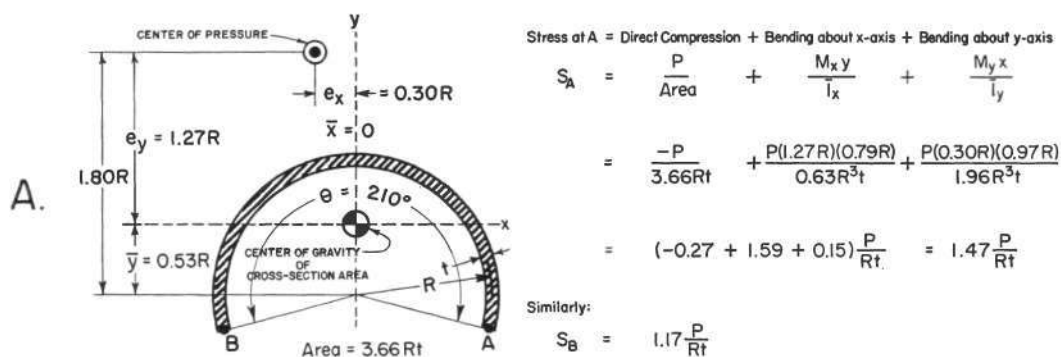


Fig. 5. Center of pressure as related to minimum cross-section of the ankle.

Obviously the highest unit stress will occur at the level of the shank where the cross-sectional area is smallest. The relationship at push-off between the center of pressure acting upward on the ball of the foot and the minimum cross-section at the ankle is indicated in Figure 5, where the ankle is approximated by a circle of radius R and where all dimensions are expressed in terms of R . If the same loading conditions be assumed to be present when a Syme prosthesis is worn, the result is a combination of three different types of stresses in the structure of the prosthesis: compression stresses resulting from the direct thrust load carried by the structure, bending stresses resulting from a tendency for the structure to bow laterally, and bending stresses resulting from a tendency for the structure to bow posteriorly. If the loading conditions and the dimensions of the cross-section are known, the magnitudes of the stresses can be calculated, as indicated in Figure 6A. In such calculations, a plus sign indicates that a fiber of the material would be in tension at the point being investigated. A minus sign shows that the fiber would be compressed.

Summarized in Figure 6B are the results of a number of calculations based on stresses in a hypothetical Syme prosthesis with a circular cross-section of radius R , with a material thickness t , carrying a load P , and with a constant eccentricity. An interesting feature is that, even when the values for direct compression as a result of proximal weight-bearing are included, in general the posterior cutout results in tensile stresses at critical points whereas the medial cutout results in compressive stresses at critical points. The posterior cutout with $\theta = 210$ deg. and the medial cutout with $\theta = 270$ deg. are perhaps most nearly representative of actual conditions.

These results would indicate that, when Syme prostheses are constructed with a posterior opening in the socket (tensile stresses at critical points), a material with the highest possible tensile strength should be used. A laminate of Fiberglas cloth with epoxy resin, such as is used by Canadian makers of Syme prostheses, would be an efficient material, particularly when reinforced with roving along the edge of the cutout. A laminate of Fiberglas



B.

POSTERIOR CUTOUT	MEDIAL CUTOUT	LATERAL CUTOUT
$\theta = 180^\circ$ $S_A = 2.34 P/Rt$ (TENSION) $S_B = 1.96 P/Rt$ (TENSION)	$\theta = 180^\circ$ $S_A = -3.47 P/Rt$ (COMPRESSION) $S_B = -1.17 P/Rt$ (COMPRESSION)	$\theta = 180^\circ$ $S_A = -2.19 P/Rt$ (COMPRESSION) $S_B = -0.23 P/Rt$ (COMPRESSION)
$\theta = 210^\circ$ $S_A = 1.47 P/Rt$ (TENSION) $S_B = 1.17 P/Rt$ (TENSION)	$\theta = 210^\circ$ $S_A = -2.20 P/Rt$ (COMPRESSION) $S_B = -0.42 P/Rt$ (COMPRESSION)	$\theta = 210^\circ$ $S_A = -1.45 P/Rt$ (COMPRESSION) $S_B = 0.33 P/Rt$ (TENSION)
$\theta = 270^\circ$ $S_A = 0.92 P/Rt$ (TENSION) $S_B = 0.78 P/Rt$ (TENSION)	$\theta = 270^\circ$ $S_A = -1.08 P/Rt$ (COMPRESSION) $S_B = -0.19 P/Rt$ (COMPRESSION)	$\theta = 270^\circ$ $S_A = -0.66 P/Rt$ (COMPRESSION) $S_B = 0.24 P/Rt$ (TENSION)

Fig. 6. Summary of stress calculations for various socket cutouts. A, Sample stress analysis for Canadian-type posterior cutout, $\theta = 210$ deg. B, Comparison of stresses at edge of cutout for varying degrees of cutout at three locations about the circumference; P , R , and t constant.

cloth and polyester resin would also be satisfactory if fabricated carefully. Either material would provide great strength and minimum thickness with more than sufficient tensile strength. Nylon stockinet with polyester-resin laminates has lower tensile strength, and the lamination would have to be thicker.

When the stresses at critical points are compressive, such as in the case of medial opening,

a material with the greatest compressive strength should be used. In situations involving compressive loading of thin-walled columns (as in a proximally loaded Syme prosthesis), failure may be due either to failure of the laminate at the area of direct compression or to buckling of the material in a localized area, such as near a free edge carrying a compression stress. The sides of the cutout in the Syme

socket with medial opening would constitute free edges of this type. To increase resistance to local buckling, the wall thickness of the laminate should be increased. Doing so will also increase resistance to direct compression because the area of the cross-section will be increased proportionally.

Since in practice it is more convenient to use nylon stockinet as a laminating material, and since the thickness must be increased to overcome the effects of buckling, nylon stockinet is probably the material of choice for the medial opening. Although theoretically Fiberglass laminates would have sufficient direct compressive strength even with thin walls, resistance to local buckling would be lower than in the case of a thicker nylon laminate. Moreover the compressive strength of a structure made of thin-walled Fiberglass laminate depends mainly on the quality of the laminating technique.

It should be pointed out that in Syme prostheses direct end-bearing has been used more often in Canada than in the United States. Since end-bearing tends to increase the critical tensile stress in the posterior-opening socket by eliminating the direct compressive stresses due to proximal loading, the need for an extremely strong laminate such as one of Fiberglass cloth, Fiberglass roving, and epoxy resin is obvious. When direct end-bearing is used with the medial opening, the critical compression stress is reduced, sometimes to the extent that it is converted into tension of some low value. Nylon stockinet and polyester resin should be an adequate material for the medial-opening socket, although such a socket is more bulky in appearance.

CONCLUSIONS

To ensure a satisfactory period of use, the ankle of any prosthesis must be so designed that the elastic members resisting dorsi- and plantar flexion have adequate volume to provide sufficient fatigue strength. Furthermore, the foot must be designed to permit the knee and hip joints to move smoothly through space during the roll-over and push-off phases. The SACH-type foot, with its sponge-rubber heel wedge and a keel of proper proportions, has

proved useful in meeting most of the requirements for use in a Syme prosthesis, but, like all other known foot-ankle units, its inability to provide energy at push-off requires that the remaining musculoskeletal system compensate for functions lost in amputation.

To satisfy the requirements of a comfortable transmission of functional stump-socket contact forces, the socket must provide the following features:

1. Comfortable support of the body weight on the distal end of the stump or on the proximal part of the socket brim or both.

2. Firm support against the anteroproximal surface of the leg at the time of push-off. Careful fitting against the wedgelike medial and lateral surfaces of the tibia can satisfy this requirement.

3. Similar support against the posterior surface of the leg at the time of heel contact. This requirement can be satisfied by pressure in the region of the gastrocnemius. Here the main interest is to prevent lost motion between socket and stump as the reaction point shifts from the posterior to the anterior surface of the leg.

4. Provision for shifting of the center of pressure against the distal end of the stump, as indicated by the force analysis. If a cuplike receptacle is provided for the stump end, it must extend around and up the sides of the bulbous stump far enough to prevent relative motion between stump and socket in the anteroposterior direction. It is particularly important to provide for the horizontal component of the force against the posterodistal region of the stump during push-off.

5. Adequate stabilization against the torques about the long axis of the leg. A three-point stabilization against the medial and lateral flares at the anteroproximal margin of the tibia and a flattening of the posteroproximal contour can be highly effective in providing the necessary torque resistance. If the needed stabilization is not provided, torques acting on the distal end of the stump will result in skin abrasion and other associated difficulties in more proximal areas.

Either the posterior cutout of the socket favored by the Canadian workers or the medial cutout proposed by the VA Prosthetics Center will result in a socket of adequate strength if a laminate of the correct type is used. When a posterior cutout is incorporated, the laminate must be capable of resisting high tension stresses. Fiberglass-epoxy laminates are therefore indicated. When a medial cutout is used, particularly in those cases where a large proportion of proximal weight-bearing is provided, the critical stresses are compressive. When compression stresses are involved, the thicker

nylon-polyester laminate may have advantages.

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