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A Prosthesis for Very Short Below-Knee Stumps

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Amputation surgery is improving constantly, and it is our impression, based on limited statistics, that the ratio of below-knee to above-knee amputees has increased significantly during the last few years.

The advantages of amputation below the knee versus amputation above the knee are self-evident when gait patterns of both types of patients are compared. From the physiological and mechanical aspects, the knee joint provides the activation of the prosthesis and a "feedback" signals source. In any level of amputation below the knee a certain amount of the musculature and most of the proprio receptors are preserved. However, the mechanical solution for the prosthetic replacement becomes more complicated with the reduction of stump length (1). The transfer of power via the stump-prosthesis interface is almost impossible with very short stumps when conventional fitting techniques are used. The proprioception however is almost fully retained, and this, in itself, is important enough to justify very short BK amputations.

In the present work, the research and development of a BK prosthesis for very short stumps was carried out with the aid of a bilateral amputee. The patient, a war veteran, was amputated bilaterally at equal level, resulting in two 5 cm-long stumps (Figs. 1 and 2). Assuming that for walking purposes a socket longer than the stump can be provided, three different solutions were explored. The major obstacle in the use of such a socket is painful compression of the posterior wall against the hamstrings during knee flexion. This was compensated for in the various solutions. The option of using a thigh corset with side bars was ruled out since the eccentricity of the axis of the side bars in relation to the natural knee centrode together with the geometry of the stump would result in a very limited knee function.

Fig. 1. Anterior view of stumps of the patient.

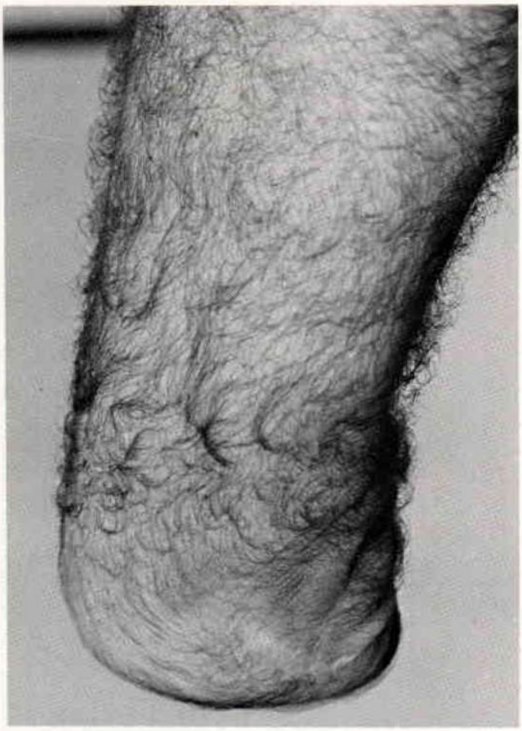

Fig. 2. Lateral view of right stump of the patient.

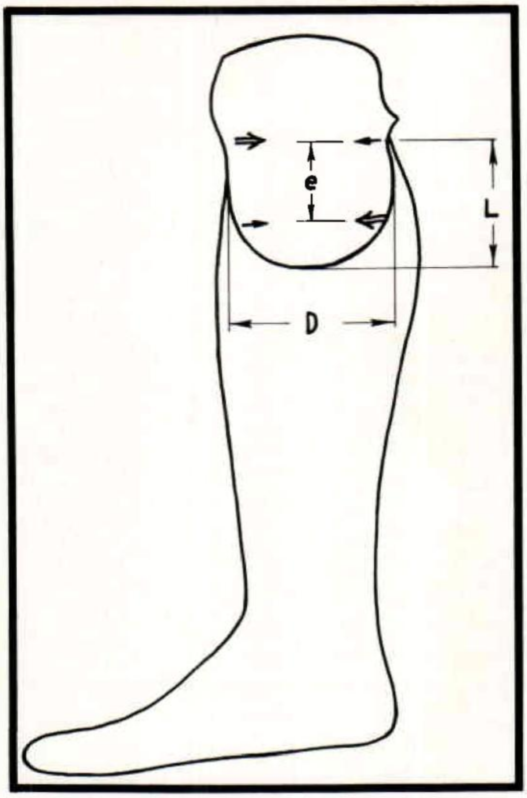

Fig. 3. Geometry of the socket for a short below-knee stump. L—stump length; D—stump width; e—eccentricity of the forces.

Problem Definition

The design of the conventional patellar tendon bearing (PTB) prosthesis is based on specialized regions of load transfer. The vertical load is carried mainly on the impression of the socket in the region of the patellar tendon. The regions on both sides of the tibial crest, the distal posterior surface, and the poplitea provide additional support in standing and ambulation. When the stump is short it becomes spherical and the moments transfer mechanism becomes quite inefficient (Fig. 3). In order to reduce the sphericity of the socket a supracondylar suspension type of PTB socket can be used with an extra long posterior wall. The posterior wall will then limit the ability of the knee to flex due to its projection against the distal portion of the hamstrings. However, with an extra extension of the posterior wall of about 5 cm the knee is still capable of flexing to 30 or 40 degrees. This amount of flexion is nearly satisfactory for level walking but may cause severe inconvenience in walking up and down inclined surfaces. For steep inclines, stair climbing, and sitting this type of arrangement is unsatisfactory. It was assumed however that a dynamically variable socket wall could provide satisfactory compensation for excessive flexion.

Initially, the sitting position seemed to be the most troublesome but later this proved to be only part of the problem. It was assumed that the stretched hamstring

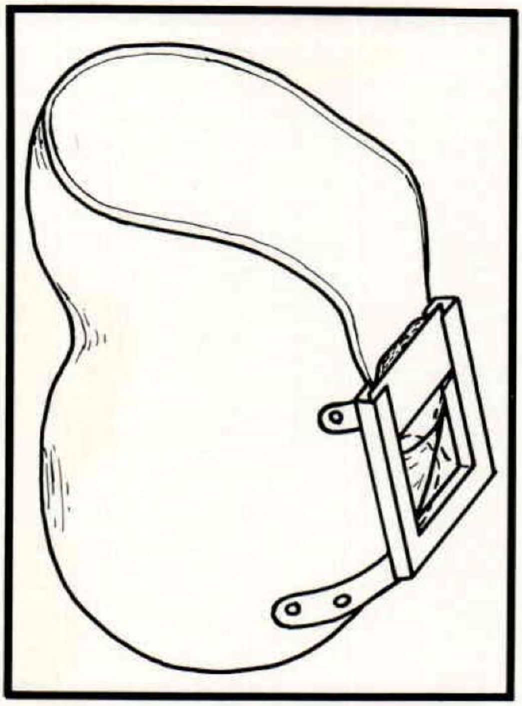

Fig. 4. Schematic diagram of a sliding posterior flap arrangement.

during excessive knee flexion forms a triangular wedge pressing obliquely against the edge of the prosthetic wall. This pressure could be relieved by providing a translatory movement of the upper section of the wall distally from the hamstrings. It was found later that the real source of the problem was a combined wedging and bulging of the soft tissue in the popliteal region during flexion.

From the point of view of efficiency of moments transfer to the prosthesis during ambulation, a long posterior wall is superior. However, when an adjustable wall is considered, the length is limited by the obliquity of the compression of the soft tissues. Three alternative solutions to the problem were therefore outlined:

1. A sliding posterior wall with limited translation of 25 mm (Fig. 4).

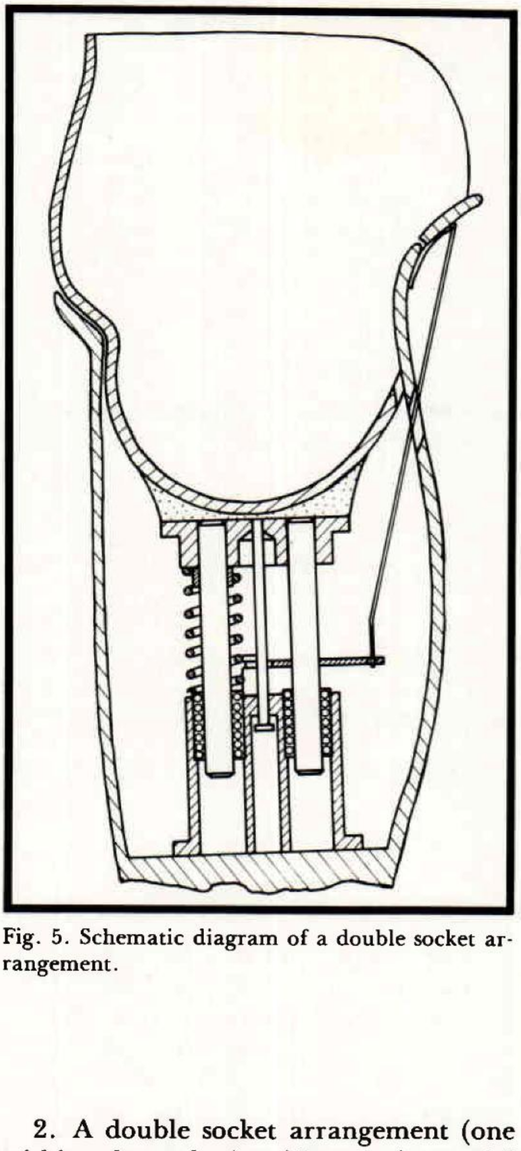

Fig. 5. Schematic diagram of a double socket arrangement.

2. A double socket arrangement (one within the other) with relative axial movement between them. The posterior support being provided by the outer socket while the rest of the support (anterior and mediolateral) is provided by the inner socket (Fig. 5). When excessive flexion occurs, the inner socket is unlocked automatically from the outer socket, thus relieving the pressure.

3. A stabilized socket attached above a specially designed four-bar mechanism (Fig. 6). At a predetermined degree of

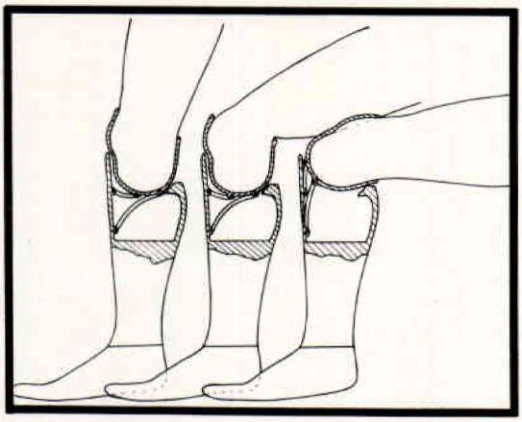

Fig. 6. Schematic diagram of a stabilized rotatory socket arrangement.

knee flexion the socket unlocks and rotates with respect to the shank guided by the four-bar linkage. In this way the amputee can sit with his anatomical knee extended while the shank is vertical. Because the stumps are very short the cosmetic appearance arising from this arrangement is not disturbing.

The advantage of the second and third solutions over the first is the ability to use an extra large extension of the socket wall; in the order of 50 mm and more. The disadvantages are in the technical complexity of these solutions and some functional restrictions.

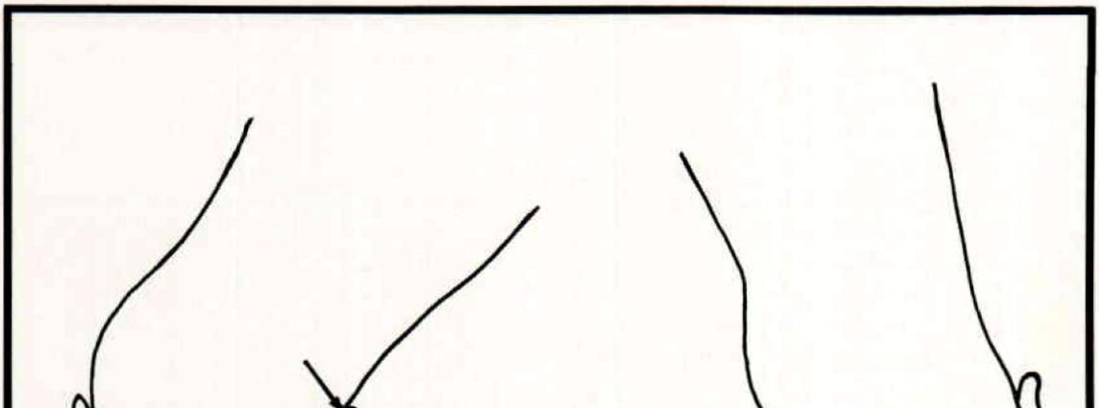

Fig. 7. Schematic Illustration of the concentrated force resultants applied on the socket wall by the stump; 7a relates to flexion under load; 7b relates to active pushing.

The Mechanical Solutions

Adjustable Posterior Wall

Figure 7 describes the forces acting on the proximal part of the posterior socket wall. Figure 7a relates to flexion of the knee when the major forces acting on this section are those applied by the hamstrings and the bulging of the soft tissue. Figure 7b relates to the push-off phase of ambulation.

It is evident from the orientation of the resultant force vector that a linearly sliding posterior flap such as the one described in Figure 4 will be subjected to various locking effects. Such an arrangement was tested and failed to produce the desired results. It was therefore decided to produce semi-linear displacement by rotating the flap about hinges which were positioned in the civinity of the center line of the socket. The socket posterior flap was cut from the socket itself and was inclined slightly obliquely to the fixed wall. The hinges and the guiding bars were positioned on the line normal to the flap, in order that the forces and the displacement be compatible. Pushing action of the stump ensured full recovery of the socket length, and knee flexion pushed the socket wall distally and posteriorly. Location of the hinge and geometry of the wall is illustrated in Figure 8. The final socket and prosthesis are illustrated in Figure 9.

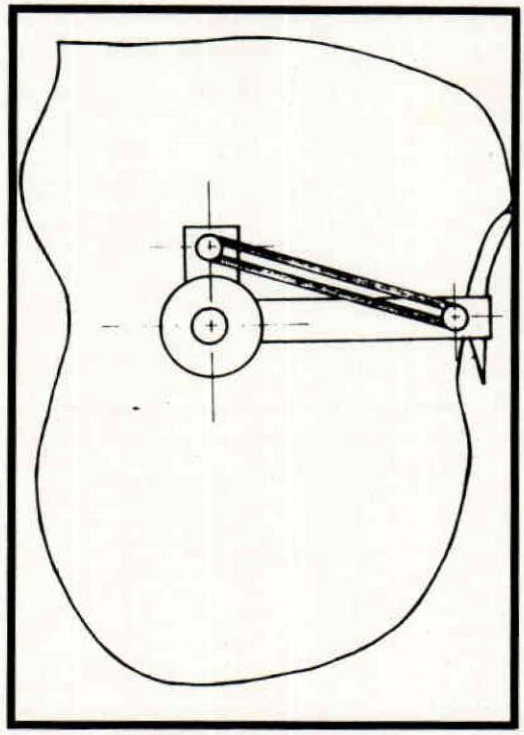

Fig. 8. Geometry of the rotatory flap arrangement.

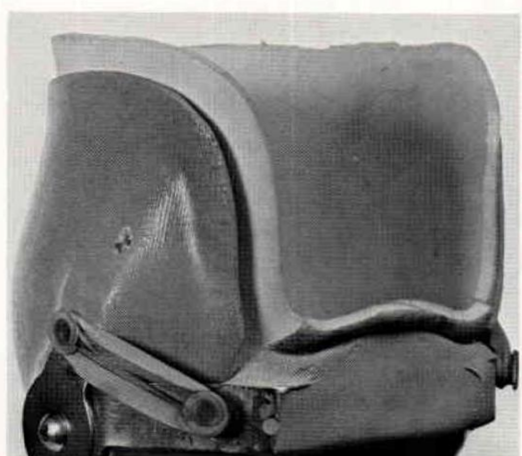

Fig. 9. The rotatory flap socket as built for trials.

Double Socket Arrangement

A PTB socket, conventional in every respect except for an extended posterior wall and supracondylar suspension, was made. An extension to the socket imitating the layout of a section of the leg was made of foamed polyester. A second extended socket was then produced by employing the ordinary casting procedure and using the first socket as a casting form. When the polyester was removed

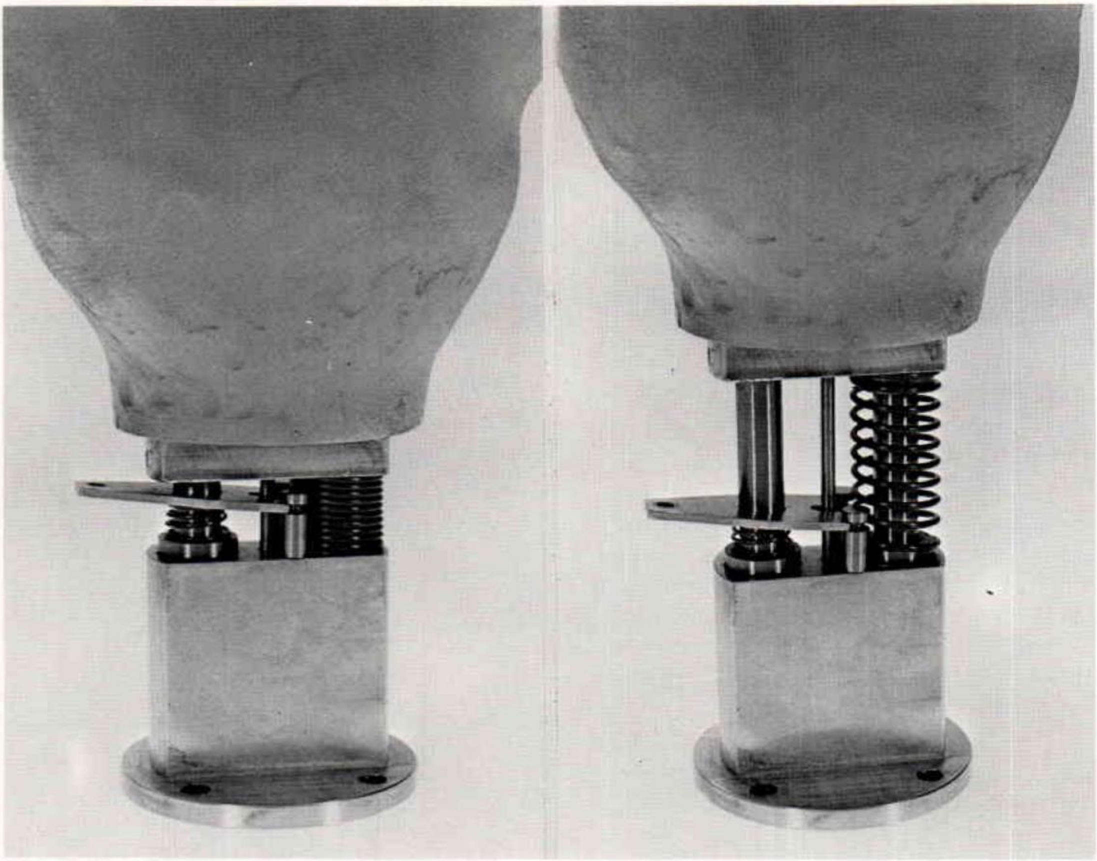

Fig. 10. A linear guiding mechanism with a self-locking arrangement.

the two sockets fitted exactly, one inside the other with a space between the distal ends. A section of the posterior wall was cut from the inner socket and glued to the corresponding part of the outer socket. A linear guiding mechanism with a self-locking arrangement was fitted in the space between the two sockets (Fig. 10). The mechanism is normally locked and the two sockets are held together tightly. When the unlocking lever is compressed, the inner socket is ejected with the aid of a compression spring up to a stroke of 40 mm. The lever is compressed directly by the action of the hamstrings via a connecting rod. The ejection of the inner socket exposes the recess in the posterior wall and the pressure on the hamstrings is relieved. This arrangement has the characteristics of a simple position servo, and the inner socket will continue to come out only as long as pressure is applied on the hamstring. It will therefore be normally exposed only to a fraction of its stroke which allows bending of the knee.

This mechanism allowed, therefore, adequate knee flexion during gait and full flexion during sitting. Additionally, it supported the stump sufficiently to provide adequate stability during ambulation.

Rotating Socket Assembly

The rotating socket design was meant primarily for sitting. A four-bar mechanism was synthesized to produce the

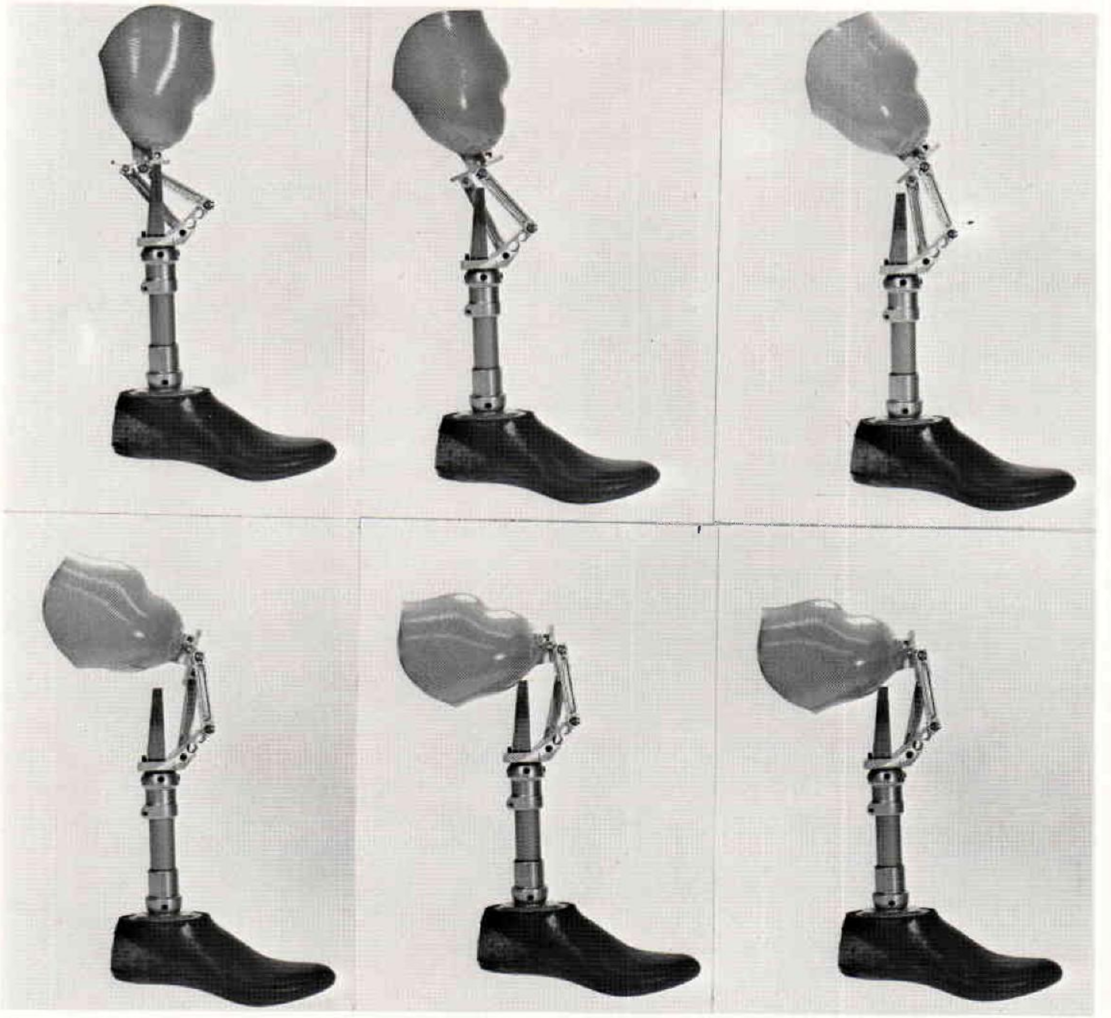

Fig. 11. A prosthesis incorporating the rotatory socket and a four-bar mechanism.

movement as specified (Fig. 6). The linkage was determined by performing a computer analysis of the path of the socket. The characteristics of a whole range of mechanisms were investigated and the optimal solution was selected. The linkage was assembled on a standard Otto Bock modular shank as illustrated in Figure 11. It can be seen from the figure that in the normal standing position the instantaneous center of rotation of the socket (the point of intersection of the two moving bars) is posterior to the center line of the prosthesis.

The initiation of flexion in the mechanism is accompanied by elevation of the socket, an energy consuming action that stabilizes the knee under load. In other words, when the mechanism is flexed slightly and a load is applied to the socket the mechanism will rotate toward full extension.

The rotating socket mechanism does not require an extension lock, and a simple elastic strap can be used to keep it in the extended position.

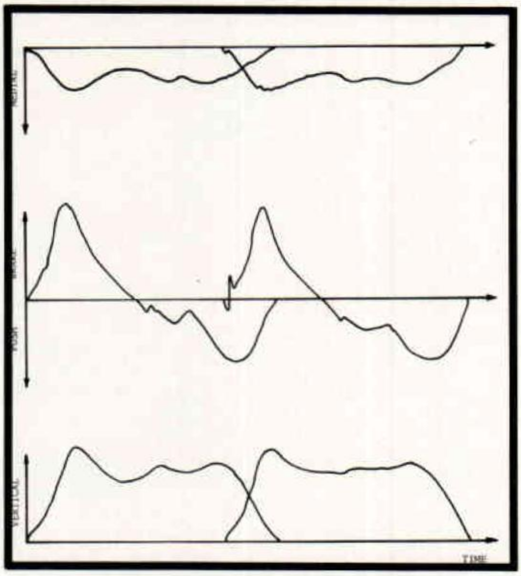

Fig. 12. Representative results of ground reaction forces during walking by a bilateral below-knee amputee using prostheses with the rotatory mechanism.

Evaluation Procedure and Results

Altogether five sockets were produced. One socket was fitted with a sliding posterior wall. This solution was found inadequate because of its failure to produce smooth sliding under load. The mechanism was also rather bulky. Two other sockets were fitted with a hinged posterior flap. The recovering moment was provided by elastic bands as described in Figure 8. The sockets were assembled on Otto Bock modular shank units and the prostheses were supplied to the bilateral amputee for use outside the laboratory. During a routine checkup the patient was examined by the medical team and his gait was evaluated with the aid of a television system and force plate dynamometers.

The amputee expressed full satisfaction with the performance of the prostheses. This was also evident from his gait. The adjustable wall of the sockets could be seen moving during walking as well as during stair climbing and sitting.

The ground force characteristics as obtained from the "Kistler" force plate dynamometers during one routine check are illustrated in Figure 12.

The degree of symmetry of the forces of both legs is within the range of normal walking. This was expected since the patient is a symmetrical bilateral amputee. From the A-P force characteristics it can be seen that the left leg is more active in braking and less active in pushing than the right leg. The total impulse is balanced and the activity of both legs as assessed by impulse measurements is equal. There is a certain degree of A-P instability at mid-stance of both legs which is expressed by the "slowing down" of the force development. This is a result of the degree of "slackness" when the supporting surfaces of the socket change roles between "braking" and "pushing".

The mediolateral force characteristics are balanced with respect to impulses. The medial force is rather high, a characteristic typical of wide-base gaits. There is no component of lateral force, and during the whole cycle the forces act only medially. This is also a feature resulting from a broad-base gait and indicates a tendency of the amputee not to cause lateral instability by alternating moments on the stump.

The features however are very good compared to normal gait and indicate excellent control during most phases of the walk cycle.

Two double socket mechanisms were produced and assembled on Otto Bock shank units. The prostheses were tried on the patient in the laboratory only. As fair as gait stability and control were considered this arrangement was very good. However this solution which is practically an inversion of the previous one failed to produce the desired results when knee flexion under load took place. This was especially severe in stair climbing. The mechanism itself however functioned sat-

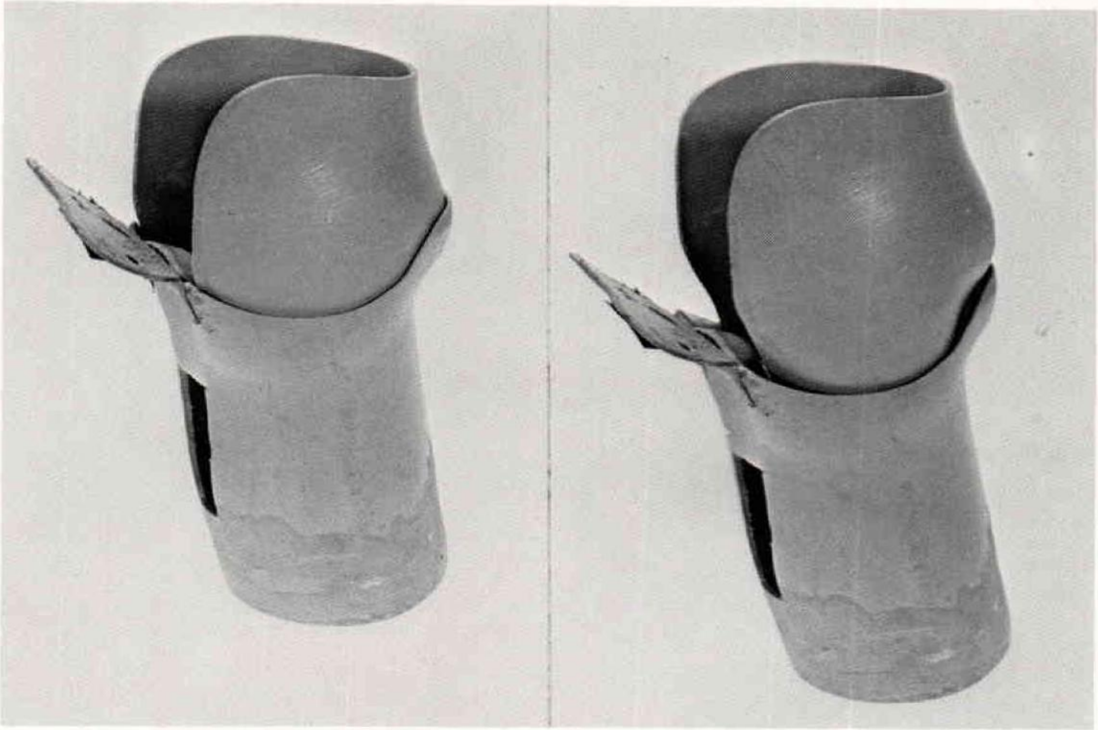

Fig. 13. The double socket arrangement in the compressed and ejected positions.

isfactorily and conformed with the design criteria. Figure 13 shows the socket in compressed and ejected positions.

One socket was fitted with a four-bar mechanism and assembled on the Otto Bock shank unit (Figure 11). This arrangement was in doubt from the early stages of the evaluation of the other prostheses. The doubt arose from the change of criteria during the examinations. The mechanism however was developed in order to investigate its kinetic properties and its potential use in more severe cases of unilateral BK amputations. This mechanism which differed in its concept from the four-bar knee mechanism is worth exploring further. Its potential to serve as a polycentric stabilized knee is of great interest. The two major difficulties were associated with the fact that the four-bar unit acted as a polycentric toggle mechanism. The increased stabilizing moments of the mechanism in the early stages of flexion were good for level walking but inhibited the ability to initiate flexion for other purposes. The second inhibiting factor was the fact that the prosthesis was used for a bilateral amputee. Initiation of uncontrollable flexion in both knees could be hazardous. This solution however could be good for unilateral ultra short stump amputation when the good leg is capable of controlling the loadbearing required of the prosthesis.

Clinical Follow-up and the Patient's Subjective View

During a period of nine months the patient was invited to the clinic periodically for examination by the medical staff. In each meeting objective gait examinations

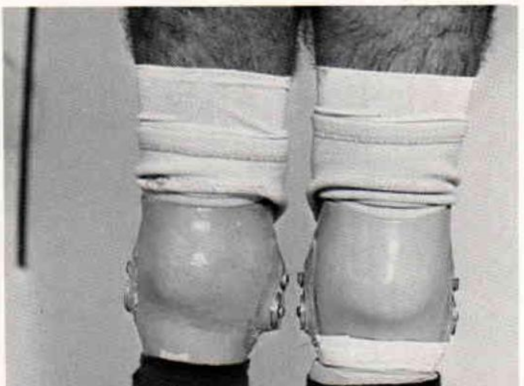

Fig. 14. Anterior view of the prostheses with the rotatory flap.

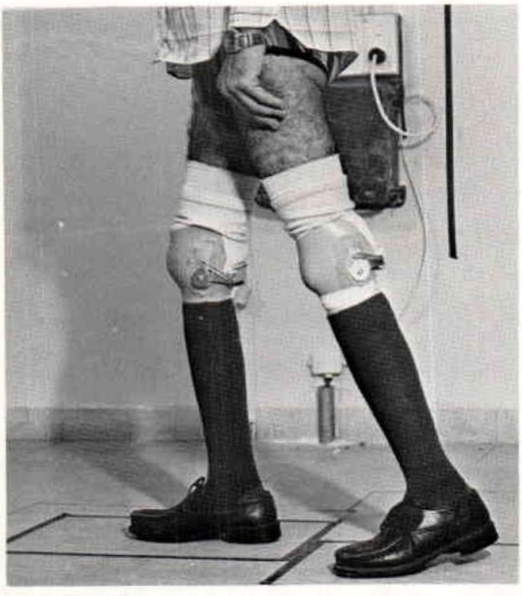

Fig. 15. Collection of data concerning the use of the prostheses with the rotatory flap.

were also carried out. The gait features were almost identical in all the examinations as described in Figure 12. Occasionally realignment of the prostheses caused slight alterations to the force records but this is outside the scope of the present work.

Medically the examinations gave excellent results. All the pressure sores and erosion of soft tissue in the stump, especially in the hamstrings and the popliteal region disappeared. A general weakness of which the patient complained while using his previous prostheses also disappeared presumably due to reduced energy requirements and reduced mental strain. A certain degree of excessive valgus was detected in one of the knee joints of the patient which seems to have been caused by the inadequate, unstable prosthesis worn previously.

The patient was satisfied fully and insisted on demonstrating to the staff his ability to control the limbs in accelerated gait, stair climbing and ladder climbing. So far, after a period of a year, the patient has not found it necessary to visit the clinic at his own initiative or to complain about faulty action or mechanical breakdowns.

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Footnotes

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