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Functional analysis of the UC-BL shank axial rotation device[1]

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Origins of concept

"Present indications are that transverse rotations of the various segments of the leg are an important factor in the ease and rhythm of walking of normal individuals. In order to improve function, reduce fatigue, and prevent more or less continual abrasion at critical points of the stump of the amputee, provision in the prosthesis for allowing transverse rotations of the same order of magnitude as present in normal legs has the possibility of being a major contribution to the improvement of artificial legs."

The above statement is a direct quote of the opening paragraph in the 1947 report on fundamental studies of human locomotion by the University of California Prosthetic Devices Research Project established at Berkeley in 1945. The goal of these studies was to provide a basis for improvements in prosthetics technology. From their results nothing stood out more clearly than the need for a device to allow passive axial rotation between socket and foot in lower-limb prostheses.

Design criteria

Design criteria for such a device were established on the basis of amputee tests of an experimental unit with adjustable stops and variable return spring characteristics. The final recommendation was that a lightweight unit be designed which would permit rotations of up to 20 degrees in either direction about the long axis of a prosthetic limb, with a centering spring torque of approximately 0.23 Nm (2 lb in) per degree of rotation. This range of motion was found to accommodate most locomotor activities without contact against the rigid rotation stops. The centering spring torque was a compromise: a stiffer spring would tend to negate the purpose of the device by allowing large torques to be applied to the residual limb; a softer spring would not return the foot consistently to centre during the swing phase.

In addition to providing these basic characteristics, any practical axial rotation unit must have bearings capable of supporting the amputee's weight and the severe bending moment that occurs during the latter part of stance phase when weight is supported on the forefoot. The friction torque in the bearings under load must be small compared to the centering spring torque if the unit is to perform its function of allowing axial rotation during stance phase. The return spring also should have adequate damping to prevent excessive vibration of the foot during swing phase. The axial rotation device should be compact to permit inclusion in the widest possible variety of prostheses, and it should be light in weight to minimize the increase in inertia of the distal part of the prosthesis. Finally, a truly practical device must be simple, reliable, and low in cost.

Previous designs

Since the need for an axial rotation device has been clearly recognized for 27 years, it is quite appropriate to ask why satisfactory devices have not been developed and made widely available to amputees. The reason seems to be that the requirements for high strength, low friction, soft but well-damped return spring, and light weight are difficult to achieve simultaneously. A number of designs have appeared, but all have lacked at least one of these essential features. The original University of California design was too large and heavy, and lacked adequate damping of the return spring (University of California, 1947). A subsequent design was compact but had excessive bearing friction which prevented rotation at the very time it was needed (Mullby and Radcliffe, 1960).

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Other designs have been unable to withstand the severe bending moment which occurs when weight is borne on the forefoot, or they have had excessively stiff return springs, or they have been excessively large and heavy (Staros and Peizer, 1972, 1973).

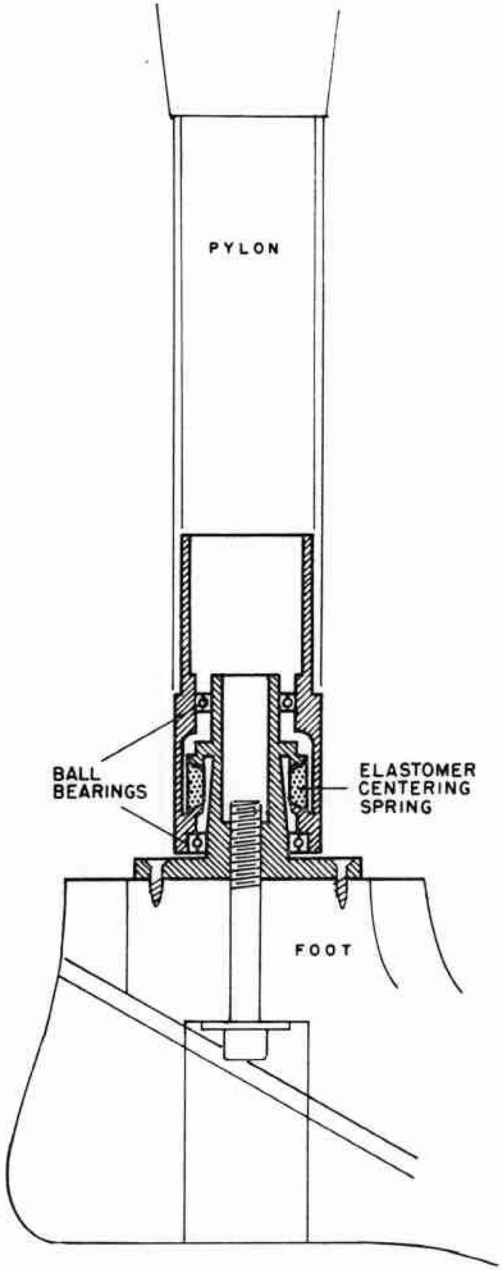

Fig. 1. The UC-BL shank axial rotation device in cross section. Although the unit is shown mounted on a foot, an alternative and preferred installation would invert the device and mount it as high as possible in the prosthetic shank. This preferred installation moves the mass of the unit proximally and minimizes bulk at the ankle.

The present design

The device described here has been conservatively designed to satisfy the original 1947 design criteria in the hope of providing increased opportunity for clinical evaluation of the concept of passive axial rotation in lower-limb prostheses. It consists of a pair of thin-section, full-ball-complement ball bearings which support axial loads and bending moments, and an elastomer torsion spring in a ring configuration as shown in Figure 1. The spring is shaped with conical end plates to allow uniform shear strains throughout the volume of the elastomer.

Amputee trials have been conducted primarily with above-knee (AK) amputees. Their response has been uniformly favourable with dramatic relief of skin abrasions and epidermoid cysts in some cases. The design criteria appear to be well suited to the needs of the AK amputee. Some questions remain whether the design is optimum for below-knee (BK) amputees, who, unlike the AK, can safely flex the knee during stance phase. Further clinical trials are needed to evaluate the specific requirements of the BK amputee. In any event, the need is less critical for the BK amputee because he generally has a normal hip joint which allows him much greater freedom of axial rotation.

Measurement procedure

The axial rotation device was installed in an above-knee prosthesis with a UC-BL polycentric knee and SACH foot. This prosthesis was then instrumented to allow measurement of axial rotation occurring within the device, axial torque in the shank, pelvis rotation about a vertical axis, and relative internal-external rotation between the socket and the pelvis. Measurements were obtained as the subject walked at a comfortable speed (100 steps/min) on a power-driven treadmill, both with the device operating and with its motion locked out.

Measurement results

With one exception, the measurements were much as expected. When the axial rotation device was operating, pelvic rotation about a vertical axis increased slightly to a total of 6 degrees, the shank rotated externally 8 degrees over the foot during stance phase, and the step length increased slightly. What was not expected was that the relative internal-external rotation between pelvis and socket during stance phase *increased* from 3 to 8.5 degrees when the device was operating.

Because of the unexpected nature of this measurement, the measurements were repeated with another AK case wearing a single-axis hydraulic knee. The results of this second set of measurements were essentially the same as the first. It is the measurements of this second amputee which are shown in Figures 2 and 3.

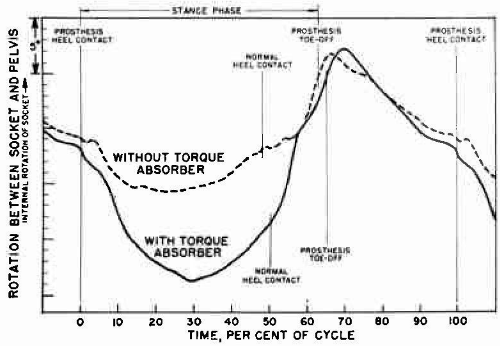

Fig. 2. A comparison of relative axial rotation between the pelvis and the socket of an above-knee prosthesis, with and without the axial rotation unit (torque absorber) operating. Note the unexpected increase in relative rotation which accompanies use of the device.

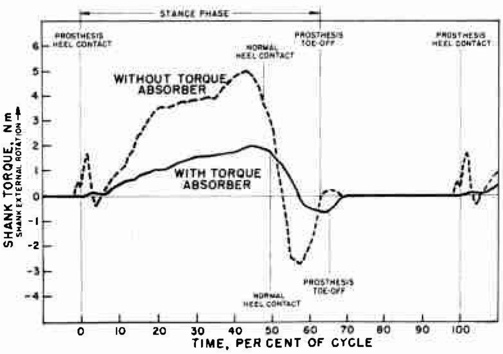

Fig. 3. A comparison of axial torques occurring in an above-knee prosthesis with and without the axial rotation unit operating. Note the decrease in torque which results from use of the device.

Discussion

Previously it had been tacitly assumed that the condition of minimum strain to the tissues around the brim of the socket would correspond to minimum relative rotation between the pelvis and the socket brim. The measurements do not support such an assumption.

When the axial rotation device is operating, axial torques are reduced, as shown in Figure 3. This reduction in torque, however, is accompanied by an increase in relative axial rotation between the pelvis and the AK socket during stance phase. This increased range of internal and external rotation can be attributed to the effect of muscle action within the confines of the quadrilateral AK socket.

At the instant of heel contact on his prosthesis, the amputee is actively extending the hip on the amputated side to assure stability of the prosthetic knee. This hip extension moment must be transmitted to the socket by means of increased contact pressures over the proximal-anterior and distal-posterior regions of the socket and stump. Since the gluteal muscles are largely responsible for the hip-extension moment there is a simultaneous generation of contact force in the gluteal region due to bulging of the contracting gluteal musculature.

The combination of these two effects gives rise to a contact pressure distribution similar to that shown in Figure 4. As illustrated, this pressure distribution generates an external rotation torque about the long axis of the socket.

When the prosthesis incorporates an axial rotation device, this net torque acting about the long axis of the socket is able to rotate the socket externally over the fixed foot, the only resistance to such rotation being the relatively weak return spring in the axial rotation device. This axial rotation tends to relieve the contact pressures which caused the torque and thereby reduces pressures in the critical antero-medial region of the brim.

When the prosthesis does not contain an axial rotation device, the hip extension moment still gives rise to the pressure distribution shown in Figure 4. Now, however, the prosthesis is very stiff in torsion and pressures cannot be relieved by external rotation of the socket. Consequently, the pressure pattern tends to rotate the stump internally within the fixed socket with little or no relief of pressure. For many amputees this combination of slight internal rotation and high contact pressure will give rise to skin trauma and the development of sebaceous cysts in the antero-medial region of the socket brim or chafing in the region of ischial contact.

As the stance phase continues and the centre of pressure moves forward to the ball of the foot, the hip action changes from active extension to active flexion as required to initiate knee flexion near the end of stance phase. This flexion moment tends to reduce the magnitude of the anterior force and increase the magnitude of the posterior force producing the socket-brim pressure pattern. At the same time the gluteal muscles relax their tension in the posterior region and the rectus femoris actively bulges against the anterior lateral brim area, allowing a posterior shift of the stump within the socket. The medially located hamstring tendons tend to resist this displacement and a pressure distribution similar to that shown in Figure 5 results. As illustrated, this pressure distribution which results from active hip flexion generates an internal rotation torque about the long axis of the socket.

Again, without an axial rotation device in the prosthesis, the socket cannot move to relieve these changing forces and the torques must be passed through the skin at the stump-socket interface, giving rise to the aforementioned skin problems which are familiar to limb fitters. With an axial rotation device the socket is free to respond to the demands of the stump and relieve the pressures and torques caused by cyclic action of the musculature.

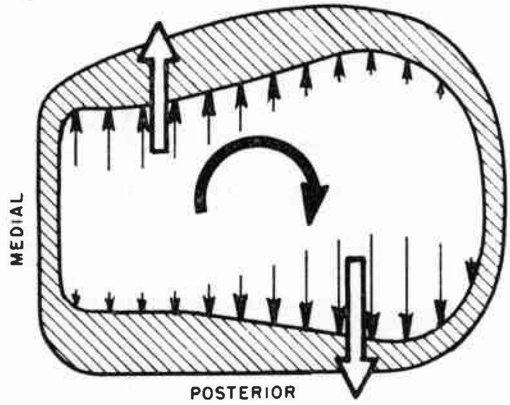

Fig. 4. Estimated pressure distribution around the brim of an AK socket at heel contact. Note the resultant external rotation moment.

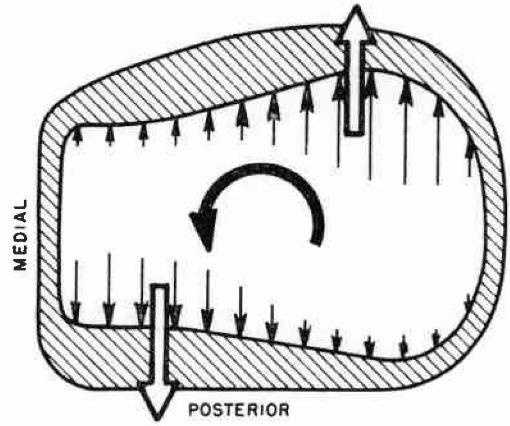

Fig. 5. Estimated pressure distribution around the brim of an AK socket just prior to toe-off. Note the resultant internal rotation moment.

Conclusions

The installation in an above-knee prosthesis of a device which allows rotation of the socket over the fixed foot has been shown to offer the amputee the following advantages:

1. Improved gait symmetry.
2. Reduction of axial torques between the stump and the socket.
3. Reduction of the frequency of occurrence, or elimination of, sebaceous cysts due to skin trauma at the socket brim.
4. Improved freedom of movement when changing direction of motion, working at a bench or counter, and in sports activities.

The usefulness of an axial rotation device for above-knee amputees has been demonstrated clinically with tests on research patients over a period of 30 years. The present design has been used successfully by several amputees for periods of up to 18 months without malfunction and it appears to have overcome the shortcomings of previous units. More extensive amputee trials will begin in the near future under the auspices of the Veterans Administration Prosthetics Center in New York City.

It is possible that experience may demonstrate that assymetrical spring rates, non-linear elasticity, or a change in rotation axis may improve future devices, particularly for the below-knee amputee.

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