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Biomechanics of the through-knee prosthesis

J. HUGHES

National Centre for Training and Education in Prosthetics and Orthotics, Glasgow

Abstract
The biomechanics of the through-knee prosthesis is considered in detail and comparisons made with the above-knee case. Socket shape and suspension are discussed and comment offered on knee function in both stance and swing phases.

Introduction

The through-knee (TK) is the most distal amputation in which normal knee function is completely lost by the patient on the amputated side. In biomechanical terms the problems are very similar to those of the above-knee (AK) level. There are, however, physical differences which may be of considerable importance in the prosthetic procedure. The end of the stump is composed of tissue normally adapted to weight-bearing in the kneeling position. A long lever arm is available for the exertion of control forces by the hip muscles and the muscles themselves are, for the most part, intact and operating in a physiological condition.

The biomechanics of the AK prosthesis is well documented and understood (Radcliffe, 1970 and 1977), while relatively little attention has been paid to the biomechanics of the TK. It is, therefore, useful to view the problems in terms of the differences which may exist between this level and the AK case.

Socket

The design of any prosthetic lower limb socket must take into account those forces which will be applied between prosthesis and stump during walking. It must provide for the application of these forces in such a way that load is transmitted to pressure tolerant areas while avoiding pressure sensitive. It must also be fashioned to apply these forces to tissues of varying stiffness, avoiding excessive pressure on the stiffer tissues and minimizing unwanted relative motion due to the excessive compression of soft tissue (Hughes 1983).

A good TK stump can take the largest part of the vertical support load on the end-bearing surface instead of, as in the case of the AK stump, requiring transmission of load to the pelvis by way of the ischium and gluteal muscles. In the medio-lateral plane, the body weight does, of course, exert a toppling effect about the point of support, during the stance phase (Fig. 1). This toppling effect is approximately equal to that which would be experienced were the

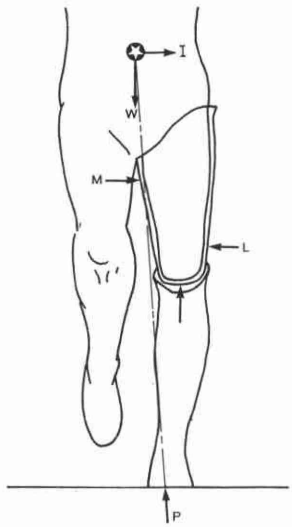

Fig. 1. TK socket—forces in the frontal plane during stance phase.

All correspondence to be addressed to Prof. J. Hughes, Director, National Centre for Training and Education in Prosthetics and Orthotics, University of Strathclyde, The Curran Building, 131 St. James' Road, Glasgow G4 0LS, Scotland.

support ischial and gluteal, as the support point is in almost the same vertical line. The summated effects of the lateral and medial stabilizing forces will, therefore, be required to act in the same general area, i.e. the lateral will be distal and the medial proximal. Because of the greater length of stump, compared to the AK, the area available to apply the forces is increased, implying lower contact pressures between socket and stump. Further, the greater stump length increases the couple arm attainable between the summated force effects therefore reducing their magnitude and further reducing the pressure.

The application of these forces is an important factor, in the design of the socket. If the lateral wall of the socket is not fitted properly to transmit force L (visualize the situation where there is a space between the stump and the socket wall) then the force will be transmitted as a shear force in the tissue of the end of the stump. This is because the end of the stump will still be forced into contact with the socket by the support force.

Figure 2 demonstrates the importance of designing the socket to transmit the medial proximal stabilizing force, M. Figure 2 (left) shows the situation in an AK socket where the support point is in the region of the ischial tuberosity. There will be a tendency for the socket to rotate about this support point. This in turn will tend to move the socket relative to the femur at its distal end. Therefore the femur must be properly stabilized in the socket. However, in the case of the knee-disarticulation prosthesis (Fig. 2 right) since there is usually no ischial support, the rotation will tend to take place about the support point at the end of the stump.

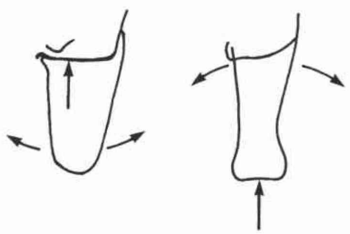

Fig. 2. Comparison of AK (left) and TK (right) sockets illustrating rotational effects about the point of support.

Ineffective stabilization will lead to relative movement between the socket and the top of the femur producing a fleshy "roll". Therefore, the force M must be applied in such a way as to avoid creating painful pressure and there should be a generous flare at the medial proximal brim to minimize the "roll".

Figure 3 shows the force pattern early in stance phase in the plane of progression. To maintain stability the resultant force, R, between ground and foot must pass in front of the knee joint otherwise the external force will tend to flex the knee. The amputee may influence the line of action of this force by extending his hip and "digging" his heel into the floor. This will have the result of reducing RH, the horizontal component of R and thus changing the slope of R so that its action line is more anterior. Consequently the socket must be designed to allow the transmission of this extension moment (that is to transmit force on the anterior proximal and posterior distal aspects).

In the swing phase, (Fig. 4), the forces tending to pull the socket off the stump—due to gravity and inertia—may be balanced by fitting the sockets over the flare of the condyles. In most sockets manufactured from a "window" is provided to allow the bulbous end to pass the narrow area above the condyles. This "window"

Fig. 3. TK socket—forces in the sagittal plane during stance phase.

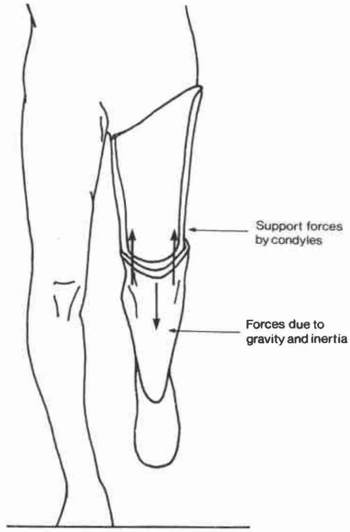

Fig. 4. TK socket—forces in the frontal plane during swing phase.

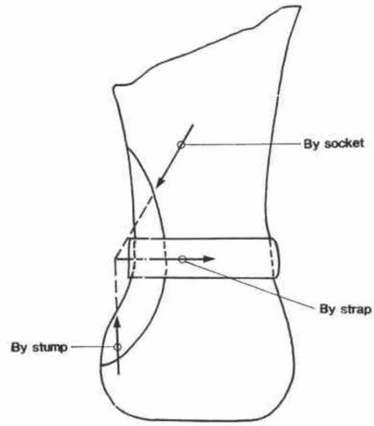

Fig. 5. TK socket—forces on the medial plate during swing phase.

will be placed on the medial side to avoid the area which must transmit the lateral stabilizing force, L. It will be seen from Figure 5 that the support force by the condyles during the swing phase will tend to displace the plate used to close the "window". A strap must be fitted to provide a horizontal force pulling the plate into the window and allowing the solid part of the socket to exert a balancing downward force on the plate.

Knee function

The functional loss suffered due to the absence of the normal knee joint is considerable for the TK amputee as it is for the AK. The knee joint is capable of a large range of motion, actuated by powerful muscle groups, and is of great functional importance in normal locomotion (Hughes, 1970). Restoration of locomotion may only be achieved by an adaptation in gait, optimum utilization of remaining musculature and the provision of passive mechanisms. The biomechanical problem may be considered in terms of (a), the stance phase and (b), the swing phase. Although the problems are the same as for the AK amputee (Radcliffe, 1977), it is useful to summarize them briefly to set the scene for the prosthetic procedures at this level.

Stance phase

The requirement is for stability at heel strike and as the body weight progresses over the support point, coupled with an ability to initiate flexion, preparatory to swing through. It is possible to provide these actions in some measure, and under voluntary control, by hip action—extensor muscle action at the hip can maintain knee extension and knee flexion may be initiated by hip flexor muscle action. There are basically three ways in which stability may be provided mechanically for the amputee who has a prosthesis with a "free" (articulated) knee. Firstly, the knee may be aligned in such a position that it lies behind the load line from hip to foot—causing the resultant force to pass in front of the knee joint and therefore render it inherently stable. Secondly, a so-called stabilizing device may be provided which develops, as a result of axial load applied to the prosthesis, a moment at the knee opposing motion. Thirdly, a polycentric knee device may be supplied. This raises the instantaneous centre of rotation of the thigh and shank in the extended position. The effect is to reduce the moment required of the hip to maintain stability in a potentially unstable situation. Of these three, only the last does not, theoretically, make it more difficult to initiate flexion in the later part of the stance phase.

Swing phase

There are two problems facing the TK amputee in the swing phase. These are replacement of quadriceps function in decelerating the upward movement of the shank after toe-off (avoiding excessive heel rise) and replacement of hamstring function in decelerating the shank in the later part of the swing phase (avoiding "slamming" into full extension).

These functions cannot be effectively provided under the voluntary control of the hip muscles but can be passively reproduced by mechanisms which develop a moment at the knee, resisting motion. Such a mechanism may operate by the development of mechanical friction in which case it will be suitable for only one rate of cadence since the friction effect will correspond to only one inertia condition. Alternatively the resisting moment may be produced pneumatically or hydraulically in which case an automatic adjustment to different rates of operation is inherent, and of a particularly suitable characteristic in the latter. Additionally there may be an elastic member biased to maintain knee extension. This aids in resisting excessive heel rise and initiating swing through but increases the rate of extension and so aggravates the problem at the end of the swing phase.

Examination of the functional requirements of the stance and swing phases shows that certain functions may be controlled or provided by the hip musculature, whereas others require the provision of mechanical devices—for the meantime available designs are all passive in nature. Furthermore, the problems are exactly the same as in the case of the AK amputee. In terms of those functions requiring the use of the hip muscle groups, the TK amputee is presumably at an advantage over the AK because of the physiological integrity of the thigh muscles and the length of lever arm available. On the other hand, because of the physical restrictions imposed on the provision of mechanical devices, he is in a less advantageous position. The femoral condyles are intact and may have the patella sutured distally. Thus the anatomical knee "centre" is considerably above the end of the stump. The medio-lateral width over the femoral condyles is anatomically normal. Thus the spatial distribution available for the provision of devices is, at best, restricted.

Summary

In conclusion the TK amputee is very similar biomechanically to the AK. The socket problems are apparently somewhat reduced with good end-bearing and large stabilizing areas. The functional replacement, required due to the loss of the knee joint, presents the same problem as for the AK, with enhanced hip control potential, but limitations in the provision of mechanical devices in terms of the relative physical relationship of stump and prosthesis.

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