THE UNIVERSITY OF CALIFORNIA FOUR-BAR LINKAGE KNEE FOR THE ABOVE-KNEE AMPUTEE

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Summary

The design of a prosthesis for the active above-knee amputee is described which incorporates systems for stance phase stability control, swing-phase motion control, and provision for relative pelvis/foot rotation. The prosthesis is the result of over 20 years of development at the Biomechanics Laboratory, University of California, Berkeley, with support from the Veterans Administration.

Introduction

In 1957 Radcliffe /1/ reported the development of an early model of a four-bar linkage knee for the above-knee amputee. The first unit incorporated a four-bar linkage similar in function to the present unit although larger in overall dimensions. The linkage has since been optimized by computer analysis resulting in improved functional characteristics as well as the present dimensions.

The swing-phase control in the earlier models was accomplished with use of a vane-type hydraulic damper acting about the uppermost pivot in the four-bar linkage. Difficulties with oil leakage over a period of several years of experimentation led to abandonment of the hydraulic vane-type damper and subsequent development of a piston/cylinder pneumatic damper unit as described below.

Knee Stability

In the 1957 report Radcliffe derived the knee stability equation:

$$M_H = \frac{L}{h} \quad (Pd - M_K) \qquad (1)$$

where, as shown in Figure 1,

M_H - the muscle moment about the hip joint on the amputated side required to maintain the weight-bearing knee in a stable flexed position;

M_K - the knee moment created by mechanical or hydraulic friction in a "brake" type design (M_K = 0 in the UC-BL four-bar knee);

L - the total length of the prosthesis from the hip to the bottom of the heel;

P - the load carried along the long axis of the leg;

h - the vertical height of the instantaneous center of knee rotation measured from the bottom of the heel;

d - the distance forward from the hip/heel line to the instantaneous center of knee rotation.

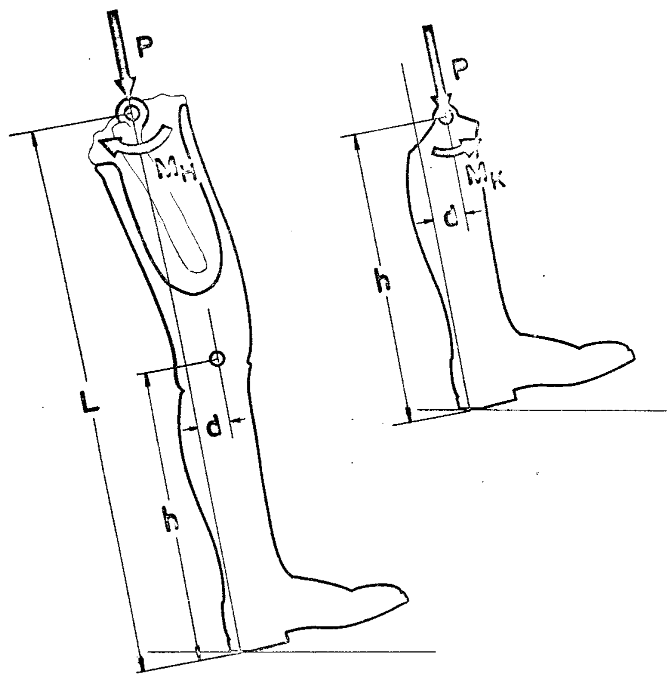

Fig. 1. Dimensions used in knee stability equation.

A complete discussion of the derivation of this equation has also been given more recently /2/.

Knee stability is influenced by three independently variable parameters: M_K, d, and h in Equation /1/. For a given kinematic arrangement it is obvious that the required M_H can be reduced to zero if the mechanism is capable of developing a brake moment.

$$M_K = Pd \tag{2}$$

This principle has been exploited successfully in a wide variety of friction-stabilized knee brake mechanisms. It should be recognized that the friction which resists flexion to provide knee stability at the beginning of stance phase may interfere with knee flexion near the end of the stance phase. As the amputee attempts to flex the knee, the negative sign in Equation 1 becomes positive because the friction moment is acting in the opposite direction, and the hip moment required to initiate knee flexion is increased as compared with a non-brake mechanism.

The parameter *d* in Equation 1 governs the so-called alignment stability.

It is possible to reduce the variable *d* to zero by aligning the prosthesis in such a manner that the knee center lies on the hip/heel line. With such an alignment, a muscle moment about the hip would not be required to maintain knee stability at heel contact. Unfortunately, however, such an alignment is not satisfactory because it provides too much stability later in the stance phase when weight is borne on the forefoot, thereby interfering with desirable flexing of the knee just prior to toe-off.

The third parameter *h* in Equation 1, in combination with dimension *d*, provides what may be described as kinematic stability, as discussed below.

The kinematic characteristics of a four-bar linkage are determined by the length and arrangement of its links. An infinite variety of arrangements is possible.

One measure of the functional characteristics attained with a particular four-bar linkage is a plot of the successive locations of the relative center of rotation of the socket with respect to the shank as a function of the knee angle, as shown in Figure 2. A larger value for *h* and/or a smaller value for *d* at a particular angle of knee flexion will result in a smaller hip moment required to maintain knee security. However, during knee flexion a large *h* also results in a cosmetically undesirable forward translation of the knee block in the region of the anatomic knee center. The knee center must eventually move downward with knee flexion or the knee will not bend in the proper place during sitting. It is desirable to maintain a large value of *h* and thereby keep the knee center high over the first 10 degrees of knee flexion in order to help the amputee recover from a stumble or other situation when knee stability is accidentally disturbed. The linkage shown in Figure 2 has been selected as a compromise which tends to maximize the desired functional benefits of the linkage while minimizing the cosmetic problems due to translation of the knee block. The kinematics of the four-bar linkage has been found to be useful to a wide variety of young male amputees, including one bilateral above-knee amputee, over periods of use ranging up to 20 years.

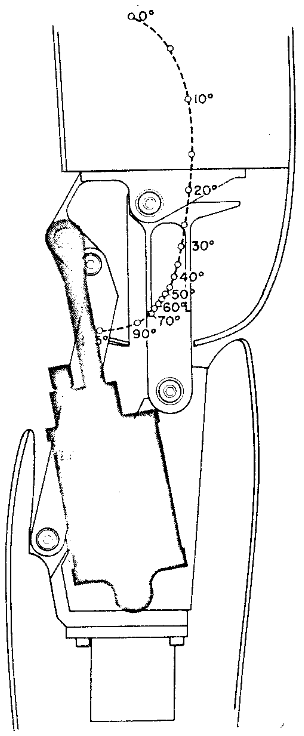

Fig. 2. Linkage arrangement and path of knee center

Figure 3 indicates the manner in which the rotation center (plotted on the moving socket) changes position during a full walking cycle. Note that it is possible to have a relatively large dimension d in full extension, which is beneficial to the amputee near the end of the stance phase. The elevated position of the center of knee rotation compensates for the large value of d by allowing the amputee to make better use of available muscle moments at the hip, and therefore provides improved voluntary control of knee stability. It is interesting to note that voluntary control is equally important in both flexion and extension, particularly to the younger, more active amputee.

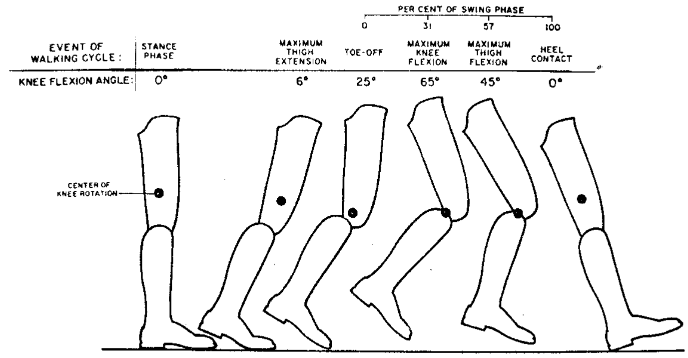

Fig. 3. Change in knee center position during walking

Swing Phase Control

The swing-phase control is provided by an improved version of the pneumatic swing-control unit described by Radcliffe and Lamoreux /3/ in 1968. The present unit has an improved valve design in which a single multifunction valve replaces the two needle valves and a check valve of the 1968 design.

The earliest pneumatic dampers used for swing-phase control in knee mechanisms were provided with separate flow paths for the two directions of motion of the knee joint. Flexion resistance and extension resistance were individually adjusted by two separate needle-valve/check-valve combinations. This arrangement was simple in concept, but, in practice, it proved to be difficult to adjust. It also had the disadvantage that it required an adjustment of both valves to accomplish a change in the overall level of swing-phase control.

Though it was not obvious from theoretical considerations, it soon became clear from experience that an optimal adjustment of the damper always required greater resistance to flexion than to extension. Consequently, the elimination of one of the check valves became feasible, with one needle valve functioning during both directions of knee motion and the other only during extension of the knee. This minor change resulted in a simplified adjustment procedure as well as a reduction in parts in the assembly. Once a well-balanced adjustment of both valves had been obtained, changes in the overall level of resistance could be accomplished by adjusting only the valve which functioned in both directions.

The new design is a logical extension of the previous system. Experience had indicated that only two different valve settings were required, and that the larger valve opening was consistently required during extension of the knee. Consequently, it was possible to eliminate the check valve entirely and have only a single airflow path and needle valve by providing a small amount of backlash, which allowed the needle to move slightly in and out of the valve seat when the direction of airflow changed. During flexion of the knee, the pressure drop across the valve forces the needle as far as it will go into the seat, providing maximum flow resistance. During extension, the process is reversed, with the needle being forced as far as it will go out of the valve seat. Simple screw adjustments of both stops are provided as shown in Figure 4. Note that the backlash adjustment is separate, but is contained within the main adjusting knob so that adjustment of the overall resistance level, achieved by turning the outer knob, does not affect the backlash setting.

Mechanical Design

Figure 5 shows the major design features of the present unit. The anterior link is constructed in the form of a U-shaped yoke to provide excellent strength and torsional rigidity. The pre-production model shown incorporates machined links and a welded sheet metal upper shank structure. It is anticipated that the production model will substitute aluminum castings or extrusions where appropriate. All beatings are of the molded nylon insert type. This type of bearing has proved very satisfactory in service and is easily replaced when required. Bearing shafts are aluminum with a hard anodized wear surface.

The swing-control unit is of all-aluminum construction except for the brass needle valve housing and Delrin valve. The unit requires minimal lubrication. Once adjusted to the needs of a particular amputee, the pneumatic unit will maintain a particular resistance characteristic for long periods.

Shank Torque Absorber

An optional functional element is the shank torque absorber installed between the upper shank welded structure and the lower pylon tubing. The torque absorber is designed with a torsional spring constant of 2.0 in-lbs. per degree of relative socket-foot rotation. It allows ± 20 degrees of rotation before reaching a mechanical stop. The shank torque absorber allows the amputee to rotate his pelvis over the fixed foot in standing or walking activity. In walking this increased freedom of pelvis motion results in improved gait and reduced chafing of the skin at the brim of the socket.

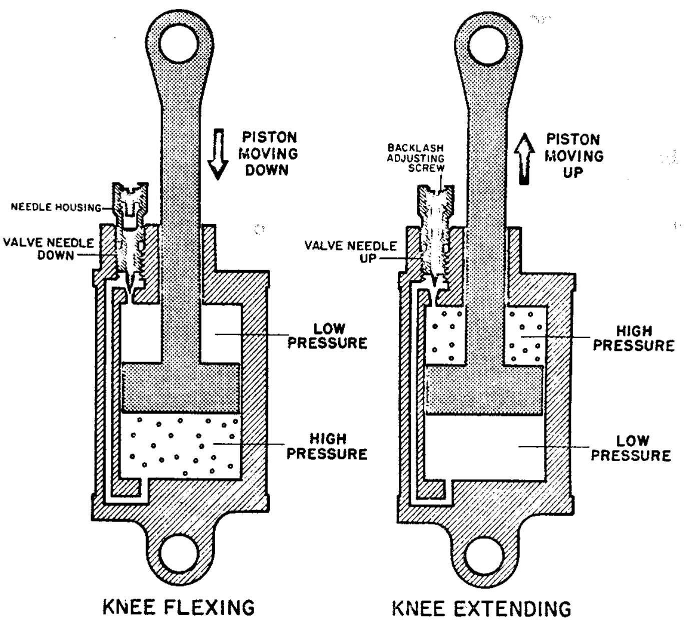

Fig. 4. Dual-action valve for pneumatic swing control

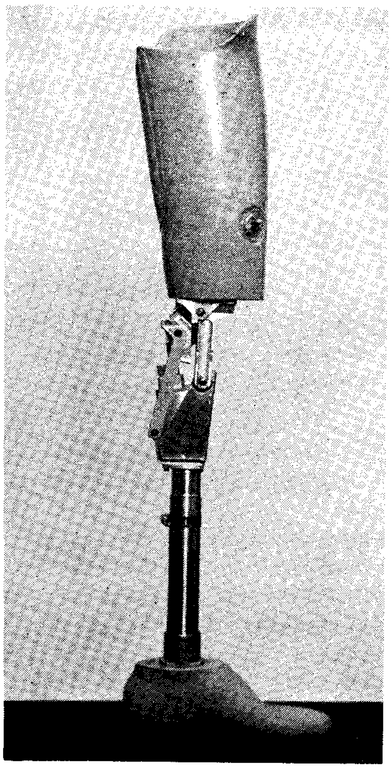

Fig. 5. The UC-BL four-bar linkage knee with pneumatic swing control and shank torque absorber

References

/1/ Radcliffe, C.W.; "Biomechanical Design of an Improved Leg Prosthesis," Oct. 1957, 72 pp. (Issued by Biomechanics Laboratory as Series 11, Issue 33.)

/2/ Radcliffe, C.W.; "Prosthetic Knee Mechanisms for Above-Knee Amputees," *Proc. Conf. on Priorities in Prosthetic and Orthotic Practise*, University of Dundee, Scotland (June 16-20, 1969) Edward Arnold (Publishers) Ltd., London, 1970.

/3/ Radcliffe, C.W. and Lamoreux, L.; "UC-BL Pneumatic Swing-Control Unit for Above-Knee Prostheses: Design, Adjustment, and Installation," *Bull. Prosthetics Res.*, Veterans Administration, BPR 10-10: 73-89, Fall 1968.