FLUID-CONTROLLED KNEE MECHANISMS
CLINICAL CONSIDERATIONS

Earl A. Lewis, M.A., R.P.T.
Prosthetic Research and Education Officer,
Research and Development Division,
Prosthetic and Sensory Aids Service,
Veterans Administration, New York, N. Y.

INTRODUCTION

The relatively recent introduction of fluid-controlled knee mechanisms for use in above-knee and hip-disarticulation prostheses has raised a number of important considerations for clinicians involved in prescribing and fabricating an artificial limb and training an amputee in its use. Until recently, it was a relatively uncomplicated process to prescribe a prosthesis for a patient, since the available mechanical-resistance knee mechanisms were of quite similar function. The considerations involved in prescribing a fluid-controlled device, however, are more complex. Their design and function differ radically from mechanical knees, many of which have provided excellent function through the years in well-fitted and aligned prostheses. Moreover, the currently available fluid-controlled mechanisms differ from each other in design and function. Conventional knee mechanisms for above-knee prostheses present certain finite limitations in the level of performance capable of being attained by an amputee. The swing characteristics provided by fluid-controlled mechanisms offer higher capacity for achieving higher levels of performance. The prescription process may have to consider these two levels of performance, one for the conventional limb and a second but higher level for users of fluid-controlled systems.

In addition, the clinician has the responsibility of selecting the particular knee mechanism that will be most suitable for the patient. This requires a good understanding of the patient's physical as well as psychological problems. If the prescription of a fluid-controlled device is based on a good understanding of the patient, if the prosthetist who fabricates the prosthesis is qualified to install this mechanism, and if the therapist knows how the device will influence training procedures, then the performance of the amputee wearing the device should evidence an improvement over his performance on a non-hydraulic system. These considerations plus the significant factors of cost, maintenance, ease and types of adjustment of
fluid-controlled mechanisms as compared to mechanical knees would be the determining factors in the selection of a prosthesis.

**AMPUTEE PERFORMANCE WITH MECHANICAL-FRICTION AND FLUID-CONTROLLED DEVICES**

**Mechanical-Friction Devices**

The knee mechanism of a typical above-knee or hip-disarticulation prosthesis which is controlled by so-called conventional means, is basically a single-axis free-swing pendulum. The swing characteristics are controlled or modified by any one of a number of techniques. Of course, we could do nothing and just allow the leg to swing at its normal pendulum action. This is called the “free knee,” where the only resistance to swing is the inherent mechanical resistance between moving parts. While a large number of amputees do walk with a free knee, an even greater number have some type of attachment at the knee mechanism which either speeds up or limits portions of the swing. These devices and attachments have been discussed by Murphy in BPR 10-1 (1). The following is a review of some of these devices: Extension aids are devices which increase the tendency for the leg to extend. They may be kick straps, hickory sticks, or other mechanisms which during flexion of the knee are stretched or put under tension. When the limit of heel rise is reached, these devices start the knee into extension. During flexion, these devices are stretched or placed under tension, and thus restrict and slow-down flexion, so that they, therefore, have a dual function. The second major category of devices affecting the swing of the shank includes resistance mechanisms applied directly to the knee bolt. These friction-producing devices limit swing during flexion and extension. In a properly aligned mechanical friction limb, judicious selection and adjustment of extension aid and knee friction device will result in satisfactory gait at a single speed, usually the amputee’s “normal” speed. Should the amputee desire to walk at a faster or slower rate, he must, to achieve proper swing characteristics, make certain adjustments in the knee controlling mechanism. These adjustments are rarely made and it is quite obvious when one sees an amputee walking rapidly or slowly that the extension aids or friction devices are incorrectly controlling the swing characteristics. When the amputee walks too slowly, stubbing of the toe frequently occurs or the knee does not fully extend. On the other hand, should the amputee attempt to walk at a very rapid pace, we see excessive heel rise, terminal impact of the knee as it goes into full extension, usually followed by carry through or a slight goosestep effect.

The fixed rhythm of swing of the conventional knee joint frequently results in a need to vault to increase the time of stance on the sound leg in order to allow for the prosthesis to swing into full extension. This, of course, is most obvious at the faster gait speed. Mechanical friction is inde-
pendent of speed. As an individual increases or decreases his pace, the resistance mechanism exerts the same amount of braking action at the knee bolt. At fast speeds, this resistance is less effective; at slow speeds, too effective. (Therefore, what is needed is a device which is cadence responsive: a device which provides low resistance at slow speeds and high resistance at fast speeds. Such mechanisms are available in the recently introduced fluid-controlled mechanisms.)

A knowledge of the characteristics of mechanical friction devices has always guided the therapist in his training procedures; and the amputee has taken advantage of certain cues inherent in this type of device to know when he may safely bear weight on his prosthesis. Similarly, the prosthetist, knowing the limitations and characteristics of extension mechanisms such as the kick strap, the hickory stick, as well as friction producing mechanisms, aligns the prosthesis to provide stability during stance, when extension aids and resistance mechanisms are least effective. Therefore, we frequently find the knee bolt placed posterior to the weight line to provide additional stability. The audible thud of the knee as it swings into full extension late in swing phase, provides a cue to the amputee that his knee is fully extended and that he may safely bear weight on the leg. However, at the same time, the kick strap would be slack and so provide a minimum of security as the foot strikes the floor. Therefore, the amputee tends to extend his stump to increase stability at the knee. He does this because he has learned that this will insure stability at heel strike. As the amputee increases his pace, he knows that he must strongly extend his stump and then rapidly place the foot on the floor to insure full knee extension. Were he to extend his stump, causing the shank to swing into full extension and then prolong swing phase, it is quite likely that the knee would again begin to flex resulting in knee instability at heel strike. Descending a ramp usually causes the amputee concern because of the possible instability of the knee at heel strike. Since the weight line may be behind the knee bolt and create a condition of knee instability, the amputee tends to walk down a ramp at an angle to the ramp or to take very short, deliberate steps. His knowledge that the knee can buckle easily can be used to good advantage in descending stairs using the so-called jack-knife technique where the heel of the prosthetic foot is placed on the tread with the rest of the foot off the step. As the amputee bears weight on the prosthesis it buckles because of the relative position of the segments and so he can descend stairs in a step-over-step manner.

Fluid-Controlled Mechanisms

Fluid-controlled knee mechanisms provide swing characteristics in above-knee and hip-disarticulation prostheses which are significantly superior to mechanical friction devices. This improvement is due to the fact that
hydraulic resistance mechanisms are cadence-dependent. Resistance to flow (if the flow is turbulent as in available systems) varies with the square of the velocity. If velocity is doubled, resistance increases four times. This basic fact plus intelligent design of systems results in mechanisms which offer little resistance to swing at low velocities and high resistance at high velocities. Additional resistance is provided at the end of the flexion and extension ranges to prevent excessive heel rise or terminal impact. Probably the single most important characteristic of fluid-controlled mechanisms which a clinician must constantly bear in mind is that with resistance set at minimum value, there is still greater resistance to swing in fluid-controlled systems than in mechanical friction devices similarly set. While resistance ranges of fluid-controlled mechanisms vary with some having lower minimum resistances than others, the clinician should remember that these devices do not swing as loosely as conventional knees set at minimum resistances. This is due to the fact that whenever the knee angle changes, the piston moves, forcing fluid to flow through passages and orifices. When a fluid is in motion, different parts of the fluid move with different velocities. The flow occurs in layers and just as there is friction at the interface where one solid moves over another, so there is friction when one layer of a fluid flows over another. This friction in fluids is known as viscosity. As the fluid in a hydraulic knee mechanism flows through the cylindrical passages, the part of the fluid which is in contact with the walls of the tube clings to it and remains almost at rest. Those layers farthest from the walls have the highest velocity. While the viscosity of fluids vary, no fluids exist with zero viscosity. We therefore find that each time we flex or extend the knee there are two mechanisms at work which preclude true free knee (or frictionless) motion: first, the friction of the mechanical elements in the system; and second, the fluid friction or viscosity. Furthermore, the faster the piston is moved the higher the resistance to flow. It is not uncommon to find that the wearer of a “free” knee, when converted to a hydraulic mechanism, rejects this device because it requires too much effort to initiate flexion and subsequently swing. While this can be offset by changes in alignment, this characteristic cannot be eliminated. While this may be a disadvantage to a few individuals who are fitted with a hydraulic device, it is this same characteristic of fluid flow which results in the prime advantages of this type mechanism.

What are some of the basic characteristics of these mechanisms which have resulted in their widespread acceptance? While this question has been covered in much greater detail in a previous issue of the Bulletin (2), a brief review will serve to point up the significant differences between hydraulic and non-hydraulic devices.

Smoothness of flow and cadence responsiveness are two phrases which aptly describe the significance and major advantages of hydraulic mecha-
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nisms. A hydraulic mechanism swings smoothly, evenly, and consistently throughout its entire range, providing the amputee with a sense of rhythm, or continuity of movement. The amputee does not experience terminal impact or any sudden or rapid changes in the rate of swing of the leg, but rather smooth transitions from the slower portions of swing to the more rapid portions of the swing phase. Once the resistance mechanism has been adjusted to provide proper resistance at the amputee's typical normal speed, changes in walking speed do not require readjustment because of the fact that the resistance to flow is velocity-dependent.

Reliability of adjustment is a characteristic of these systems not found in mechanical friction devices. Mechanical friction braking devices are subject to wear as a consequence of use. Friction at the interface causes the "brake lining" to wear out. As this occurs readjustment becomes necessary to maintain proper resistance level. In a fluid-controlled mechanism resistance is controlled by varying the orifice diameter using a valve mechanism. For all practical purposes these mechanisms do not "wear" and therefore resistance does not decrease with use.

FLUID-CONTROLLED SYSTEMS

Three systems which are commercially available have been approved by the Veterans Administration and are being issued to eligible beneficiaries on a prescription basis. These systems are known as swing-control systems in contrast to a second type of unit, the swing- and stance-control system. While no swing- and stance-control units are currently available to veteran beneficiaries, one system will be evaluated in a Clinical Application Study in the near future based upon recommendations made by New York University (3). The three swing-control units referred to above are: Henschke-Mauch "HYDRAULIK" Swing Control System Model B, manufactured and distributed by Mauch Laboratories, 3035 Dryden Road, Dayton, Ohio; Dupaco "Hermes" Hydraulic Control Unit, manufactured and distributed by Dupaco, Inc., 205 North Second Avenue, Arcadia, California; Hydra-Cadence, manufactured and distributed by Hydra-Cadence, Inc., 623 South Central Avenue, P.O. Box 110, Glendale, California.

These mechanisms provide knee control and knee ankle coordination during the swing phase of gait. They do not provide a knee lock, nor do they appreciably increase knee stability in the stance phase. They are, therefore, no substitute for a polycentric knee or any other mechanism normally prescribed for stability of the knee.

Henschke-Mauch "HYDRAULIK" Swing Control System Model B

The Henschke-Mauch Model B System is composed of a single axis knee, a special plywood knee block, an 18 in. willow wood shank and the
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hydraulic unit. The setup is available in left and right sides based on shank shape. The knee block, however, is neuter. The unit must be installed in the special Mauch setup. It cannot be installed in conventional setups. The unit is attached to the knee block by means of two knee straps which are secured by the knee bolt. In addition, a piston rod head pin secures the piston rod to the knee block slightly behind and above the knee bolt. Attachment to the anterior aspect of the shank is by means of a shin attachment plate and two shin attachment screws. An adjustment screw between the two shin attachment screws permits changes in the flexion-extension attitude of the knee. A movable ring at the top of the unit and accessible from the rear of the shank permits independent adjustment of flexion and extension resistance.

The following is quoted from the “Manual for the Henschke-Mauch ‘HYDRAULIK’ Swing Control System for AK Prostheses,” published by Mauch Laboratories, October, 1963:

“The Henschke-Mauch ‘HYDRAULIK’ Swing Control System (Fig. 1 and 2) is designed to enable above-knee amputees to walk with various walking speeds or ‘cadences,’ and to do this in a natural appearing manner. This is achieved by providing swing forces about the knee joint of the prosthesis, which approximate closely the muscle forces acting during the swing of a natural leg at various cadences.

Figure 1. The “HYDRAULIK” unit.

“The ‘HYDRAULIK’ setup (Fig. 3) is available in three knee widths (4, 3\(\frac{3}{4}\) and 3\(\frac{1}{2}\) in. between hinges, right and left). Existing prosthesis may be converted to use of the ‘HYDRAULIK’ System by exchanging the knee and upper shank portions. Extension stop and kick strap need not be
provided, since these functions are built into the 'HYDRAULIK' System. Extension aids (kick straps) or other components located in the shank should be removed from the old prosthesis prior to installing the 'HYDRAULIK' System, since a saw cut will be made through the mid-shank.
"Figure 2 shows the basic design of the hydraulic swing control unit, comprising piston rod (C), piston (B) which is mainly a guiding means within the cylinder (A), and hydraulic fluid (D). A second smaller piston (G) moves up and down within the control bushing (H), which is tapered at the upper and lower end and is screwed into the dashpot (F), the bottom of which forms an inside taper enclosing the lower tapered end of the control bushing (H). The conical gap between these two tapers can be widened or narrowed by screwing the control bushing (H) up or down within the dashpot (F). Also screwed into the dashpot (F) is the swing adjustment screw (J), which has an inside taper facing downward and enclosing the upper tapered end of the control bushing (H). The conical gap between these two tapers can be widened or narrowed by screwing the swing adjustment screw (J) up or down within the dashpot (F). This same swing adjustment screw permits adjustment of the lower conical gap by means of a pin which engages the upper rim of control bushing (H).
"A number of channels (N) originate each from one of a group of staggered holes in the lower inside wall of the control bushing (H) and terminate within the upper conical gap. Other comparable channels (M) originate each from one of a series of staggered holes in the upper inside wall of the control bushing (H) and terminate within the lower conical gap. Lip seals (K & L) prevent escape of fluid from the lower and upper ends, respectively, of the control bushing (H) but permit fluid entrance into the bushing, thus acting as check valves. Accumulator piston (E) allows the fluid level to rise whenever the piston rod (C) is pushed into the system.

"At the beginning of the swing phase, the prosthesis is bent, the hydraulic system is compressed, and the dashpot piston (G) moves downward within the control bushing (H). The fluid pressure below the piston (G) keeps the lip seal (K) closed, and the fluid flows through the staggered holes in the lower inside of the control bushing (H) into the channels (N), then up to the upper conical gap, and from there through the lip seal (L) into the space above the piston (G), thus closing the fluid cycle. While, at the beginning of the compression stroke, all of the channels (N) serve as fluid passages, their number decreases as the piston (G) passes over more and more of the staggered holes in the lower inside wall of the control bushing (H). The progressive flow restriction thus produced depends, regarding the damping profile, on the arrangement of the staggered holes and regarding the damping degree, on the width of the upper conical gap, which can be adjusted from the outside to satisfy individual needs. Basically the same events occur during the expansion stroke, except that the functions of the lip seal (K), of the lower staggered holes, of the channels (N), and the upper conical gap are now taken over by the lip seal (L), the upper staggered holes, the channels (M), and the lower conical gap, respectively. Both sets of staggered holes are so arranged as to produce optimal simulation of the muscle actions during flexion and extension in the swing phase."

**Dupaco "Hermes" Hydraulic Control System**

The Dupaco "Hermes" Hydraulic Swing Phase Unit is available in a choice of two special wooden setups, made by United States Manufacturing Company and Ohio Willow Wood Company. These setups differ from standard setups made by these companies in that they offer improved knee bearing action and reinforcement at the attachment points. Setups are available in lefts and rights and in a variety of sizes. Setups manufactured by United States Manufacturing are available in 6 sizes, measured between side joints as follows: 3 in., 3⅓ in., 3½ in., 3¾ in., 4 in., 4⅛ in. The Ohio Willow Wood setups also offers a choice of 6 calf sizes: 30 to 40 cm. in 2 cm. increments. In addition, the unit can be installed in most existing prostheses by means of a Precision Boring Fixture Set (Model No. 60880),
also made by the Dupaco Company. Screw adjustments accessible from the rear of the shank allow for relatively independent control of the flexion and extension portions of swing.

The text and the illustrations which follow were provided by the manufacturer to describe the function of the unit (4). Figure 12 shows the Dupaco setup with the unit exposed by cutting away a portion of the rear of the shank. Note the location of the flexion and extension adjustment screws.

Figure 4 shows the relative motion of shank and knee with the resulting motion of the shaft of the control unit during the flexion portion of the swing phase. Note that the foot and ankle are rotating counterclockwise (posterior) with respect to the knee bolt and knee. The positioning of the upper and lower attachment shafts in relationship to the knee bolt results in the attachment points approaching one another, thus moving the piston rod into the control unit. Note that the upper attachment shaft is fixed in the knee block and the lower attachment shaft in the shank or leg portion. Also, observe that at full extension, stance-phase stability is achieved through standard alignment procedure.

Figure 5 shows the same relative motions during the extension portion of the swing phase. Note that all relative motions are reversed. The attachment points are moving apart, and the piston rod is moving out of the control unit.

Figure 6 shows a functional schematic of the Dupaco “Hermes” Control Unit at full extension just starting the flexion portion of the cycle (corresponding to Figure 4). The flow paths outlined are used during
initial flexion, a "free swing" portion of the cycle. During this portion of the cycle, flexion resistance is minimized to match natural gait pattern. The main flow of the silicone hydraulic fluid is through the large central bypass ports, up through the extension check valve (which has been opened by the pressure differential), and through the large extension bypass ports to the upper side of the piston. Some fluid follows a parallel path through the extension needle valve and some flows through the small
piston rod volume. As in flexion, some fluid also flows between the piston and cylinder wall for temperature compensation later in the stroke. The energy storage spring aids extension at this time by extending and returning stored energy to the system. This helps achieve normal gait patterns and also reduces the energy required by the amputee in walking.

Figure 9 shows the piston approaching the end of the “free swing” portion of the extension stroke. The flow is essentially the same as in Figure 8 except that some fluid is also flowing through the throttling ports. By this time the energy storage spring has returned its stored energy to the system.

Figure 10 shows the piston at the start of extension resistance buildup. Flow through the flexion check valve has now ceased. The main flow is through the throttling ports with a bypass flow through the extension needle valve. As the piston moves upward to cover the throttling ports sequentially, the hydraulic pressure builds up above the piston to cause programmed deceleration of the prosthesis in a natural manner. The flow through the extension needle valve controls peak deceleration and, thus, the amount of energy absorbed in stopping extension travel. For this reason the flexion needle valve must be adjusted by the prosthetist after control unit installation to stop the prosthesis smoothly at full extension.

Figure 11 shows piston at full extension—with correct adjustment of control needles, the fluid resistance due to pressure buildup will stop the piston just as the extension bumper touches the end of the cylinder. At this point all flow has ceased and ready to start the next cycle as shown in Figure 6.
The Hydra-Cadence setup is composed of a single-axis knee with hydraulic resistance mechanism, with piston rod pivoted behind the knee axis, hydraulically controlled ankle, wooden foot, cosmetic cover, and hardware necessary to attach the unit to any socket. Six sizes are available, with nominal size (measured from the knee center to floor without shoe) from 16½ in. through 21½ in. at one inch increments. With the exception of the foot and the cosmetic cover, all parts are interchangeable, left or right. The Model E unit is the latest model and is the only unit manufactured. It differs from earlier models in details which do not affect overall function. The Model B requires periodic lubrication of bearings at the ankle and knee. In addition, there are sealing boots on both the knee and foot pistons. On the Model C series, periodic lubrication was eliminated, bearing material was changed, and the boots were removed and replaced with
FROM STEWART—VICKERS TO HYDRA—CADENCE MODEL D

FIGURE 13

STEWART-VICKERS with YIELDING KNEE LOCK

without YIELDING KNEE LOCK

MODEL A  MODEL B  MODEL C  MODEL D

HYDRA—CADENCE

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a more efficient wiper ring. Cosmetic cover design is the same on the B and C models. Two foot styles were available on the B and C models. The original foot was all wood requiring a toe-break while the revised foot was wood and crepe rubber and did not require a toe-break. Figures 13 and 14 illustrate some of the models from the early Stewart-Vickers up to the Model D. External appearance differences are slight between the D and E Models. For a description of how this system functions the reader is referred to “Properties of Fluid Flow Applied to Above-Knee Prosthetics” in BPR 10-1 (2).
Adjustment of the Systems

Two classes of adjustments are possible on a fluid-controlled system: those which affect alignment and made only by a prosthetist; and those which may be made by the amputee and clinic team members. This section will discuss only the second type, those adjustments which are not restricted to the prosthetist. For an amputee wearing either the Mauch or Dupaco systems, these adjustments will vary swing characteristics. Amputees wearing a Hydra-Cadence system may adjust swing characteristics and the heel height of the foot.

Henschke-Mauch Model "B." This system has provision for independent adjustment of flexion and extension characteristics by means of a novel adjustment mechanism at the top of the system by which the cap (Knurled Adjustment Knob) and a second but lower segment are rotated. Clockwise rotation of the segments (when viewed from the top, looking down at the cap), increases resistance; counterclockwise rotation decreases resistance. Changes in the position of the cap with respect to a vertical black line on the lower segment results in changes in flexion resistance. Rotation of the lower segment results in extension resistance. On the top of the cap is a name plate. The word "HYDRAULIK" on this plate and two brass pins (Adjustment Knob Pins) one to the left of the letter H and the other to the right of the letter K serve as reference marks with the black line to indicate flexion resistance settings (Figure 15-18). Note that in Figures 15a

![Figure 15](image)

Figure 15. (a) Minimum flexion, moderate extension, (b) minimum flexion, minimum extension.
FIGURE 16. (a) Maximum flexion, maximum extension, (b) minimum flexion, maximum extension.

FIGURE 17. (a) Minimum flexion, moderate extension (higher than Fig. 15a), (b) moderate flexion, moderate extension (same as Fig. 17a).
and b, 16b, and 17a, the pin to the left of the letter H is above the resistance marker (black line). This relative position of pin and resistance marker indicates minimum setting for flexion resistance. In each figure, the black line is in a different position in relation to the cylinder; this has no bearing on flexion resistance. Relative position between cap and resistance marker determines flexion resistance setting; while relative position of resistance marker and cylinder determines extension resistance. In Figure 16a, the pin to the right of the letter K is above the resistance marker.

This setting provides maximum flexion resistance. In Figures 17b and 18, we see two other flexion resistance settings illustrated. In Figure 17b, the flexion resistance is higher than in Figure 18. Figure 15b illustrates the position of the resistance marker (black line) when set for minimum extension resistance while Figures 16a and b illustrate the maximum extension resistance setting. Clockwise rotation of the segment causes the resistance marker (black line) to move from right to left when viewing the unit from the rear. The settings illustrated in Figure 18 are average settings for wearers of the Mauch unit. The settings in the other illustrations are of interest even though not all of them are typical.
It is uncommon for amputees to use high resistance settings regardless of how much mechanical friction they used previously. During the Veterans Administration's Clinical Application Study of the Henschke-Mauch system, a record was kept of the settings used by 31 test wearers (5). The following is quoted from the final report of that study:

“The adjustment indicators on the Mauch system cannot be converted to numerical values indicating percent of available resistance used because the relationship between adjustment settings and actual resistance is non-linear. This is, of course, true with any hydraulic system. We did, however, compute the mean flexion and extension settings based on percent of rotation of the adjustment ring. The mean flexion adjustment was set at approximately 24 percent of the range, while the mean extension setting was at 21 percent of the range. These values result in actual resistances which are considered low. In terms of actual resistance these settings provide considerably less than 20 percent of possible resistance.”

All adjustments of flexion and extension resistance are achieved by rotating the unit cap which in certain positions will also rotate the lower segment on which is the resistance marker (black line). An example should clarify the method used. Let us assume that we wish to adjust the unit to the setting shown in Figure 18. Begin by rotating the Knurled Adjustment Knob in a counterclockwise direction, until it is not possible to rotate it any further. The unit is now at a setting for minimum flexion and extension resistance and should look like Figure 15b. Rotate the adjustment knob clockwise until the adjustment knob pin to the right of the letter K is above the resistance marker. At this point, the unit is for maximum flexion and minimum extension resistance. Although not visible to the viewer, the adjustment knob pin to the right of the letter K has engaged the resistance marker segment and further rotation of the adjustment knob will result in movement of the resistance marker. Continue rotation of the knob until the resistance marker is in the position shown in Figure 18; the adjustment knob pin is still above the resistance marker indicating that flexion resistance is set at maximum value. Now rotate the adjustment knob counterclockwise (note that the resistance marker does not move) until the letter D is above the marker. The system is now adjusted as shown in Figure 18. Note that in making adjustments, changes in extension resistance occur only when the flexion resistance is at the end of the range either minimum or maximum. It is at these points that the adjustment knob pins engage the resistance marker segment. For this reason, when making changes in setting, it is easier to establish the extension value first, then the flexion setting.

 Dupaco “Hermes.” The Dupaco swing control system has provision for independent adjustment of flexion and extension resistance by means of
standard screw slots accessible at the top of the shank from the rear of the prosthesis. When viewed from the rear, the screw on the left is for extension adjustment while the screw on the right is for flexion adjustment. Clockwise rotation of the screws increases resistance while counterclockwise adjustment decreases resistance. When the adjustment screws have been turned approximately four full turns counterclockwise from full closure, resistance is set for minimum values. Additional rotation in a counterclockwise direction, although possible, will not have any effect on resistance settings. Satisfactory adjustment is achieved when heel rise is limited to a cosmetically acceptable limit and the leg reaches full extension without terminal impact. Flexion resistance adjustments should be made first.

*Hydra-Cadence.* The Hydra-Cadence system provides adjustment of swing-phase resistance by means of an adjustable valve which is accessible through a small hole in the back of the cosmetic cover. A small “wrench” is provided with each unit to make the adjustment. Adjustment of the valve affects both flexion and extension characteristics. Clockwise rotation increases resistance while counterclockwise rotation reduces resistance. While approximately six complete turns of the control knob are possible, the valve is completely opened and at minimum resistance setting when backed off 1½ to 2 turns from full closure.

Near the top of the hydraulic mechanism and accessible from the rear is an adjustment knob (painted green) which is used to adjust ankle angle (dorsi-plantar-flexion) to accommodate varying heel heights. While the actual range is approximately 3 in., the practical range is closer to 1½ in. While this range allows a male to vary heel height from a sneaker to engineer boots and is therefore quite adequate, a woman may desire to wear heels in excess of 1½ in. To satisfy these needs, the manufacturer makes special feet to permit the use of heel heights from 1½ to 3 in. Clockwise rotation of the heel height adjustment knob plantar-flexes the foot; counterclockwise rotation dorsi-flexes it. A word of caution: when adjusting foot position, it will be noted that the foot will dorsi-flex as the knob is being rotated. However, when attempting to plantar-flex the foot, this immediate change does not occur. The foot must be passively plantar-flexed for the change to occur. Should the person making this adjustment neglect or forget to position the foot, it will appear when no further rotation of the knob is possible, as if the adjustment is inoperative. Should the amputee then attempt to walk on the prosthesis, he will find that at the first heel strike when the foot plantar-flexes that the foot will go into the maximum heel height position. This could result in the foot breaking as the amputee unknowingly attempts to bear weight with the foot in the position. A fall is a likely possibility.
INDICATIONS AND CONTRA-INDICATIONS

General Indications

Because a major feature of any fluid-controlled mechanism is resistance to motion, or "friction responsiveness" to cadence, the principal indication for such mechanisms is an active amputee who frequently desires to change walking speed over a wide range.

Because the shank of a conventional prosthesis with elastic extension bias and mechanical friction (constant at all speeds) tends to swing at a constant frequency, its amputee user must walk at a single cadence. To increase walking speed (in feet per second or miles per hour), he, therefore, can only lengthen his steps. To do so he must vault by vigorous plantar-flexion of his good ankle to rise on the ball of the good foot. Simultaneously he must flex his hip more vigorously to move the thigh stump and socket forward to a greater angle than usual and thus to place the knee bolt farther forward in relation to the rest of his body. Because of the constant swing characteristics, the time taken per step will remain substantially the same as at normal walking speed. Thus, the shank also will now swing farther forward before reaching the extension stop of the knee. The heel of the shoe on the prosthesis will land on the ground farther forward than usual because of a combination of the greater forward displacement of the knee joint and the increased inclination of the shank ahead of a vertical line through the knee. Obviously there is a limit to the possibility of inclining the shank without serious risk of buckling of the knee, of falling backward, or of skidding on slippery ground. In addition, such vaulting not only is conspicuous, detracting from good cosmetic appearance, but also causes excessive energy consumption at high speeds. For substantial increase in speed, hopping on the good foot may even be required.

Fluid-controlled mechanisms, in contrast, by providing resistance increasing with cadence, allow substantial increase in walking cadence and speed without vaulting and permit better voluntary control of the shank from the hip.

Also, an individual amputee may have occasion to walk at slow, almost stately, cadences at which a friction device and an elastic extension bias adjusted to his normal cadence may cause excessive risk of stubbing of the toe and stumbling, whereas the decreased resistance of a fluid device may be more appropriate. A further general indication for fluid-controlled mechanisms is strong interest in and need for essentially normal appearance of gait, particularly during swing phase.

Vocation may be an important consideration. Certain occupations may have specific requirements which are more easily fulfilled by fluid-controlled mechanisms. Walking with rapid but short steps about a bench may be
facilitated by the substantial knee extension bias characteristic of the Hydra-Cadence mechanism, particularly within 20 deg. of full extension. To some extent the Mauch and Dupaco units also offer extension bias, though more so at larger angles of knee flexion. In addition, the need in either occupation or avocation to walk frequently over rough ground would appear to be an indication for any of the fluid-controlled mechanisms with their improved control of the knee joint and, consequently, clearance of the toe over the ground. The dorsi-flexion or the toe lift featured by the Hydra-Cadence, though, decreases as the knee returns toward extension, so it is substantially reduced late in swing phase when the toe is actually closest to the ground. The ability of the Hydra-Cadence unit to plantar-flex passively during sitting with the shank extended diagonally forward or to dorsi-flex during sitting with the shank diagonally backward under the chair may have some cosmetic or even practical values in business, travel by public conveyance, or social life.

Anticipated maintenance and repair experience should be considered when selecting a system, particularly if the amputee must travel long distances to a limb facility. While it is always possible for a given unit to have an atypical service record, the VA's experience has shown that the Mauch System has the best service record followed by the Dupaco and then the Hydra-Cadence systems.

**Energy, Effort, Fatigue, and Weight**

When considering the prescription of an artificial limb, the clinician may find it important to know whether a particular system might be unnecessarily taxing on the individual or whether weight differences are significant. During the course of the VA Clinical Application Studies (5, 6, 7) of the three hydraulic systems discussed in this article, subjective data on effort and fatigue, and objective data on comparative weights of prostheses were collected on 170 cases.

Hydraulic prostheses are usually heavier than conventional limbs. A total of 133 of the 170 hydraulic limbs fabricated for the above mentioned studies were heavier than the amputees previous conventional limbs. Thirty of the hydraulic limbs were lighter than the prestudy conventional units while 7 were the same weight. The range of increased weight was from 2 ounces to 4 pounds with a mean increase of 1 lb. 4 oz. Mean weight of the conventional limb was 7 lb. 7 oz. while the mean weight of the three types of hydraulic limbs was 9 lb. 5 oz.

After they became accustomed to the characteristics of the hydraulic limbs only 31 subjects perceived of the hydraulic limbs as being heavier than the conventional limbs. The 170 subjects were queried as to whether they felt that the hydraulic limbs required more, less, or the same amount of effort to use as the conventional limb. Eighteen felt that more effort
was required to use a hydraulic limb while 110 stated that less effort was required to use the hydraulic limb. The remaining 42 did not note any differences. When the subjects' opinions were requested as to whether the hydraulic or the conventional limb was more fatiguing only 19 indicated that the hydraulic was more fatiguing, 100 stated that the conventional was more fatiguing, and the rest did not feel that there was any difference.

Unfortunately, only limited data have been gathered on the expenditure of energy on a hydraulic limb as opposed to a conventional limb (8, 9). These data do not reveal any significant difference in energy cost between hydraulic and non-hydraulic mechanisms when both are properly fitted and aligned. These findings plus the subjective findings from the Veterans Administration studies would appear to indicate that where a prosthesis is not medically contra-indicated, either type, hydraulic or conventional, can be safely used and the typically heavier weight of a hydraulic prosthesis is not a contra-indication. Indeed, what may be more important than gross weight is the distribution of the center of mass of each segment with the evidence indicating that the higher the center of mass of a segment, the less the effort. In other words, a shank having its center of mass closer to the knee will require less force to move than if the center of mass was closer to the foot.

**Contra-Indications or Cautions**

While it is obvious that any of the usual contra-indications against actual use of a prosthesis for above-knee amputees, will be equally applicable to consideration of the prescription of a fluid-controlled system, such cases comprise only a small percentage of the typical clinic case load. The majority of cases seen at amputee centers may not derive sufficient benefit from a hydraulic device to justify the added expense or be suited for these systems for less obvious reasons.

There are numerous cautions which should be carefully considered before prescribing a fluid-controlled mechanism. Presently available devices are chiefly appropriate for average adult sizes with reasonable length of mid-thigh stumps. Height of the total individual, length of the opposite shank, and stump length should be considered, individually and in combination. Relatively short individuals cannot as yet be satisfactorily fitted with any of the available mechanisms. None is yet appropriate for small children, although tall teenagers may well be able to use some of the available devices. It is hoped that the growing popularity of such devices, plus increasing prosthetics services to juvenile amputees, will eventually lead to development of smaller sizes in rugged construction.

Correspondingly, a combination of a long thigh stump (or especially a knee disarticulation) in a relatively short individual would be at least a caution and probably a strong contra-indication, even if the amputee would
C FOR SUCTION SOCKET ALLOWS FOR SEALING PLATE AND DISTAL SOCKET CLEARANCE VOLUME FOR VALVE FUNCTION.

C FOR OTHER A/K SUSPENSION ALLOWS FOR DISTAL STUMP CLEARANCE AND END PLATE.

FIGURE 19
accept lowering of the knee bolt compared with the level of the opposite knee. The skill and the patience of the prosthetist concerned are important factors in predicting whether such limitations can be overcome.

As shown in Figure 19, the Hydra-Cadence unit requires at least $1\frac{3}{8}$ in. above the knee center to the top of the plate. Further space, depending upon the type of socket to be used, is also needed below the bottom of the stump. In this connection, either the hard closed-end total-contact socket or the conventional pelvic band type of socket will permit fitting of longer stumps than could be fitted with a foam-type soft closed-end total-contact socket or an "open-end" type of suction socket requiring approximately 1 in. more of clear space between the end of the stump and the bottom of the socket.

A serious flexion contracture of hip, bringing the distal end of a long stump far forward of the knee axis, creates a cosmetic problem in any prosthesis. Such a condition should cause particular concern before prescribing a Hydra-Cadence prosthesis because its metal shank frame and plastic cosmetic cover must be substantially vertical over the ankle and because the shank cannot be shaped or faired as can wooden shanks.

In the case of the Mauch Model B, $\frac{3}{4}$ in. of wood in the knee block must be allowed above and toward the posterior slots. Forward flexion of the hip and thus forward displacement of the distal end of the stump may permit conservation of this amount of wood posteriorly even with relatively long stumps, though possibly at the expense of cosmetic appearance of the anterior portion of the thigh unless the stump is quite conical in form.

The Dupaco unit requires a minimum of $\frac{1}{2}$ in. between the knee bolt center and the end of the stump, provided an intimate contact between the end of the stump and the top of the knee bolt can be tolerated. More clearance is likely to be needed.

The overall height limitations, again related to the length of the shank of the opposite leg in the case of the unilateral amputee or to the desired total height of the amputee in the case of a bilateral, vary somewhat among units. The shortest appears to be the Dupaco unit in a wooden shank, needing only 8 in. below the knee before narrowing of the shank for the shaping above the ankle can be tolerated. Other units need at least 2 in. more than the Dupaco below the knee but above the narrowing portion in the shank. Because an important consideration for use of hydraulic mechanisms is an interest in appearance of gait, it would seem undesirable to sacrifice cosmetic appearance of the static shank by tolerating an exceptionally thick ankle region in order to allow room for the fluid-controlled mechanism.

Caution and careful consideration should be given to relatively low motivation of the patient to improve appearance of his gait. If, for example,
he has been long accustomed to walking with a free knee joint with no mechanical friction and slack in the extension bias, it is very questionable whether he would gain from the additional friction and extension bias provided by the available fluid-controlled mechanisms. The major interest and goals of the amputee should be carefully analyzed by the clinic team.

Similarly, some caution should be used before prescribing a relatively expensive and complicated fluid-controlled mechanism for an amputee who is careless about the appearance of his gait even at his normal cadence. Some amputees do not seem to object to extreme knee flexion and heel rise early in swing phase, undue vaulting upon the good foot in mid-swing of the prosthetic shank, or slamming of the knee into full extension at the end of swing phase, faults easily overcome at a single cadence by proper adjustment of the prosthesis, by gait training, and especially by adequate motivation.

The other extreme, finicky concern by an amputee about static appearance of the prosthesis or worry about acoustic cues to his status as an amputee (for example, foot slap), presumably would not be a factor in the designs using conventional ankles and feet but would create caution in prescribing the Hydra-Cadence mechanism. Its hydraulic ankle joint permits relatively free plantar-flexion over a very large range and, therefore, permits some degree of foot slap immediately after heel contact, as found in a high fraction of those amputees who have been systematically studied. Thus, it seems that there is to some extent a dilemma between providing improved appearance of the overall pattern of gait while introducing foot slap which calls attention to the gait.

Certain occupations may either create indications, as noted above, or cause concern and possibly great caution before prescription. Hobbies may have similar effects.

Caution should be observed in prescribing the Hydra-Cadence type of fluid-controlled mechanisms, with its large range of both plantar- and dorsi-flexion, for individuals who have occasion to wear rubber boots, high leather boots (such as farmers or surveyors use), or similar restrictive footwear. Such boots may interfere appreciably with the action of the ankle joint and reduce the effectiveness of the entire unit.

During the VA Clinical Application Study (6) of the Hydra-Cadence prosthesis there was some slight indication that this mechanism was not appropriate for use by drivers of tractors. Physical damage to the unit could occur while climbing on and off the tractor seat by contact against parts of the tractor, though a curved metal shin guard added to a later model offers some protection. The large range of ankle motion may perhaps cause some difficulties.
Corrosion by salt water, whether from external exposure or from accumulation of perspiration because of leakage in the socket, likewise caused severe corrosion damage to Hydra-Cadence mechanisms in some rare cases during the clinical application study and subsequent use. Probably none of the units is appropriate for use by wearers in serious risk of immersion in water, especially in salt water. (Such exposure, of course, is deleterious to any of the ordinary conventional artificial limbs as well, so special attention should be given to the use of stainless steel or other non-corrosive metals and to widespread application of nylon bushings and plastic laminate construction for individuals who necessarily risk such exposure.)

Location of the patient with respect to his prescribing clinic team and to the prosthetics facility must be carefully considered in the prescription of any fluid-controlled mechanism. Training of the amputee to use his newly found agility is particularly important during the early weeks, perhaps months, of use of the fluid-controlled mechanism. Thus, the patient should be located reasonably close to the training facility. Occasional follow-up and possibly a greater amount of servicing than would be required with conventional mechanisms may well be necessary, so he should be located also within a reasonable distance of the prosthetics facility, and he should have the possibility of obtaining adequate time from his work and appropriate travel funds (personal or sponsored) to permit reporting to the prosthetics facility as needed. Mere distance alone cannot be specified because in certain parts of the country travel of scores or even hundreds of miles may seem less of a barrier to frequent visits to the therapist or the prosthetist than would a bus or subway ride in a congested metropolitan area. Again, psychological motivation of the patient and recognition by the patient, his family and his employer of the importance of training and servicing may be far more important than mere numerical values of distance or travel time.

Economic aspects may work in either direction. Certainly the initial price of any of the present fluid-controlled mechanisms is much higher than that of a conventional knee with elastic extension bias and mechanical friction. Some people claim, though, that ultimately the total cost of prosthetics services over a period of years may actually prove less for the fluid-controlled mechanisms because of the ease of replacing individual standardized components for complete subassemblies, while preserving other portions of the prosthesis. At the present time, with only a few years of experience for such devices, no firm determination can be made on the validity of this argument. So far, the high initial price of hydraulic mechanisms would appear to present a serious caution in the cases of elderly individuals, private purchasers, or sponsoring agencies with limited bud-
It may well be that the initial price will decrease appreciably as such devices are used in much greater volume.

**Bilaterals.** Bilateral above-knee amputees pose an unsolved problem. No experience was obtained from the VA Clinical Application Studies because bilaterals were deliberately excluded from the initial sample. It is expected that cautious use will be made now that these devices are available on contract. The relatively few cases fitted with bilateral Hydra-Cadence prostheses at the Prosthetics Education Project at the University of California at Los Angeles, appear to have functioned quite satisfactorily.

It is claimed, apparently logically, that these individuals particularly benefit from the large range of plantar-flexion with very low resistance through increased knee stability at heel contact and improved ability to descend slopes. The considerable knee extension bias, particularly when the leg is bent less than 20 deg., is also considered to be favorable for these bilateral above-knee amputees.

Certainly the bilateral cannot vault on one artificial foot to allow a longer step with the other prostheses; thus, to vary speed he must change cadence. Even mild saving of energy or increase in range of cadences through any of the fluid-controlled mechanisms may be very welcome to such severely handicapped patients.

Useful experience on bilaterals apparently is lacking on any of the other fluid-controlled mechanisms. All others, of course, provide only knee control with no special provision for ankle function. Certainly bilateral cases should be fitted with even more consideration for alignment than is accorded to unilateral amputees. Adjustable couplings allowing trial with the actual fluid-controlled systems should be used routinely. Apparent hip flexion of the socket bore should compensate not only for any true hip flexion but for cutting of the anteriorly bowed femur at a midpoint leaving the remaining segment in an apparent state of flexion. Not enough is yet known of the possible distortion of biomechanics as a result of the pelvis rocking about the ischial seat as well as the femur rotating about the anatomical axis in the acetabulum. It is to be expected that this false location of an effective hip joint will be influenced by end-weight-bearing through the femur or at least total contact with the end of the stump.

**Elderly Amputees.** Geriatric amputees, whose special problems have been excluded thus far from the VA Clinical Application Studies and probably from most other experiences, should be fitted with fluid-controlled mechanisms only with considerable caution. Physiological conditions, of course, should be considered rather than mere chronological age because some individuals are spry and active at advanced chronological ages whereas others may be quite decrepit at a relatively early age. The clinical application studies largely ruled out geriatric amputees in the usual sense by
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requiring that the amputee subjects be active, vigorous individuals, as well as service-connected and otherwise eligible for a new prosthesis.

In the special circumstances of the Veterans Administration, most so-called geriatric amputees as a result of cardio-vascular disease are likely to be non-service-connected and, therefore, each becomes financially responsible for maintenance of his prosthesis after initial issuance. By definition they are in a VA hospital because they have stated they have been unable to afford the cost of their own hospitalization, medical care, and prosthesis. Pending substantial further experience on maintenance and economic factors, issuance of a fluid-controlled mechanism which is potentially susceptible to unusually high maintenance costs would thus seem unwise and scarcely helpful to the patient. Further, true geriatric amputees (in the usual sense of the term, with emphasis on physiological deterioration) will not be likely to have occasion to walk at a wide variety of speeds or to be especially concerned about appearance of gait.

As yet, there is no appreciable experience in the issuance of fluid-controlled mechanisms to individuals long ago amputated as young and active men but who have now attained advanced age while retaining considerable levels of activity and high aspirations to function and appearance.

Female Amputees. The VA Clinical Application Studies likewise had no experience with women amputees. The Prosthetics Education Program at the University of California at Los Angeles has fitted a few Hydra-Cadence prostheses, reportedly quite successfully. There has been very little reported experience with other units on female patients.

The ability to vary heel height with a Hydra-Cadence prosthesis is a feature which may be especially appropriate for women. In the other units, of course, interchangeable SACH (Solid Ankle Cushion Heel) feet should be considered to provide possibilities for wearing high or low heels at will. The extension bias of several of the units, enabling rapid short steps about a workbench, might be especially useful in kitchen work.

In fitting women, the size problem discussed above should be especially carefully considered. The limitations on both circumference of calf and ankle joint of the standard Hydra-Cadence shank should be especially noted. Some of the other fluid-controlled mechanisms, fitted within more conventional wooden shanks, might be more appropriate for individualized carving or sculpturing provided that the limitations on minimum shank length were not severe.

TRAINING CONSIDERATIONS

Should an amputee receive training if he is being fitted with a prosthesis incorporating a hydraulic system? I am sure that if this were the amputee's first prosthesis, we would all agree that he should. I am not so sure
that everyone would agree that training is equally important for an amputee who, after wearing a conventional prosthesis for many years, is receiving a hydraulic prosthesis for the first time. While there are exceptions, we have found that amputees who do not receive instruction in the proper use of a hydraulic system, do not derive the fullest potential from the hydraulic function. Training on a hydraulic prosthesis, which includes instruction in the making of adjustments, is strongly advocated in all cases. Such training, or "orientation" (if more acceptable to the amputee who may feel he is a good walker and not in need of training) should stress the changes in walking habits which the wearer must make.

A sizable majority of above-knee amputees habitually extend their stump (using hip extensors) late in swing to insure full extension of the shank at heel strike. While many therapists teach this maneuver, it is unnecessary and undesirable. A properly adjusted prosthesis, especially if it has an extension aid, will always fully extend. The knee of the prosthesis is kept in full extension until the weight line passes in front of the knee by pressing back against the socket with the stump after heel strike. This is almost automatic as peak activity of the gluteus maximum, the principal hip extensor, normally occurs immediately after heel strike. When walking on a hydraulic prosthesis, it is particularly important for the amputee to avoid this gait deviation as it will result in an unnecessary expenditure of energy. A hydraulic system, by its very nature, resists changes in velocity; therefore, the amputee's attempt to accelerate forward swing of the shank by a hip extension movement will result in higher resistance to swing. Amputees should be instructed to decelerate slowly and smoothly the forward swing of the thigh, ending with the stump pointing to the anticipated heel strike point; at that moment the knee will be fully extended and ready for heel strike which is accomplished without hesitation. The entire walking process should be a smooth translation of the body forward without noticeable acceleration and deceleration.

Step lengths should be of equal length. Typically, amputees take a longer step with the prosthesis than with the sound leg. This usually is the result of a habit pattern which developed because of the amputee discomfort or mistrust of the prosthesis during training on his first limb. As a result, the amputee attempts to decrease stance-phase on the prosthetic side. When converting an amputee to the use of a hydraulic limb, it is usually easier to teach the amputee to shorten the prosthetic step rather than increase the length of the sound leg step. While equal step length is desirable for all amputees, those amputees who are wearing a Hydra-Cadence prosthesis will experience a secondary deviation if he takes a long prosthetic step. The longer the prosthetic step, the greater will be the angle between foot and floor at heel strike. Since this device permits relatively free plantar-flexion, the greater the angle of travel, the greater
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the velocity of the foot at foot flat and therefore the louder the foot slap.

For a more detailed discussion of training consideration for amputees wearing a Hydra-Cadence prosthesis, the reader is referred to Hydra-Cadence Guidebook (10).

SUMMARY

The Dupaco “Hermes,” the Henschke-Mauch “HYDRAULIK” System Model B, and the Hydra-Cadence System are hydraulic knee mechanisms for above-knee and hip-disarticulation amputees, which provide marked improvement in prosthetic performance over the conventional non-hydraulic knee mechanisms. Because swing-phase characteristics of hydraulic mechanisms are superior to mechanical friction systems most amputees who are properly fitted and trained in the use of hydraulic mechanisms will demonstrate an improvement in gait when compared to a non-hydraulic system. However, the degree of improvement may not be sufficient to warrant the increased expenditure to purchase and maintain a hydraulic prosthesis.

In the main, selection of a specific hydraulic mechanism will be influenced by the individual biases of clinic team members, and quite possibly, of the amputees. Some of the characteristics of the three systems are summarized in Table 1.

<table>
<thead>
<tr>
<th>Characteristic</th>
<th>Mauch “B”</th>
<th>Dupaco</th>
<th>Hydra-Cadence</th>
</tr>
</thead>
<tbody>
<tr>
<td>Independent control of flexion and extension resistance</td>
<td>Yes</td>
<td>Yes</td>
<td>No</td>
</tr>
<tr>
<td>Temperature compensation</td>
<td>Yes</td>
<td>Yes</td>
<td>No</td>
</tr>
<tr>
<td>Integrated knee and ankle function</td>
<td>No</td>
<td>No</td>
<td>No</td>
</tr>
<tr>
<td>Wood setup</td>
<td>Yes</td>
<td>Yes</td>
<td>No</td>
</tr>
<tr>
<td>Permits choice of foot</td>
<td>Yes</td>
<td>Yes</td>
<td>No</td>
</tr>
<tr>
<td>Foot provided</td>
<td>No</td>
<td>No</td>
<td>No</td>
</tr>
<tr>
<td>Adjustable heel height</td>
<td>No</td>
<td>No</td>
<td>Yes</td>
</tr>
<tr>
<td>Provides extension bias</td>
<td>Yes</td>
<td>Yes</td>
<td>No</td>
</tr>
<tr>
<td>Can be installed in existing prostheses</td>
<td>Measurement between side bars</td>
<td>Measurement between side bars or calf size</td>
<td>Measurement from knee-center to floor</td>
</tr>
<tr>
<td>Sizing</td>
<td>Yes</td>
<td>Yes</td>
<td>Yes</td>
</tr>
<tr>
<td>Training of new wearers indicated</td>
<td>Yes</td>
<td>Yes</td>
<td>No</td>
</tr>
<tr>
<td>Increase in weight of prosthesis a problem</td>
<td>No</td>
<td>No</td>
<td>No</td>
</tr>
</tbody>
</table>
Cautious, yet realistic, use of the various fluid-controlled mechanisms should make their distinct benefits available to active adult above-knee amputees having appropriate vocational and avocational backgrounds who are concerned with appearance of gait at a variety of speeds. Use of fluid-controlled mechanisms on bilateral amputees or women unilateral amputees, because of limited experience, should be regarded as somewhat experimental thus far. Geriatric amputees, for a number of reasons, are probably poor risks for such devices.

ACKNOWLEDGMENT

The author is indebted to Dr. Eugene F. Murphy, Chief, Research and Development Division, Prosthetic and Sensory Aids Service, Veterans Administration for his generous assistance in the preparation of this article.

REFERENCES