THE SWING PHASE OF WALKING WITH ABOVE-KNEE PROSTHESSES

Eugene F. Murphy, Ph. D.
Chief, Research and Development Division, Prosthetic and Sensory Aids Service, Veterans Administration, New York, N.Y.

After many years of development, fluid control of the swing phase of walking now exists in commercially available above-knee prostheses, and other models soon should pass the experimental stage. The purpose of this review of historic and modern principles for control of artificial knee joints during swing phase is to assist the professions concerned with amputee rehabilitation to provide these new devices, not only for appropriate disabled veterans but also for all others who may be able to use them effectively. The two papers following this discuss principles of mechanical and fluid friction, specific mechanisms, and recent clinical application studies (1, 2).

Fluid-controlled mechanisms for the stance phase are also to be expected. At least one example, the Henschke-Mauch Model A "Hydraulik" Swing-and-stance Control System, is at an advance stage of evaluation. A subsequent issue of the Bulletin will consider this portion of gait.

FLUIDS

Fluid-controlled mechanisms make use in varying degrees of several key properties of fluids. Pressure at any point in a fluid is transmitted to all other points. A liquid is incompressible, for all practical purposes, within the range of pressure used in prosthetics. Thus, it can transmit energy or control signals from point to point, as in hydraulic brakes on an automobile, and can return reactions, as from a spring-loaded reservoir piston. A liquid also resists shearing forces as one layer of fluid slides over another. During oozing and slow "viscous" flow, this sliding occurs in an orderly fashion like cards in a deck; during more rapid "turbulent" flow, random whirlpool-like motions result. Another key feature of fluids is that resistance to flow increases rapidly as the flow rate increases, as we all know from experience with door closers. These properties are discussed in detail in the next paper, "Properties of Fluid Flow Applied to Above-Knee Prostheses (1)."

Air and other gases can be considered fluids if they are subjected to only small changes of pressure. Generally, however, a pneumatic
artificial knee control may be subjected to considerable changes in air pressure so that the compressed air behaves like the air springs supporting buses, yet also to such small pressure changes that leakage of air through a control valve follows to a considerable extent the rule of rapid increase in resistance for a small increase in speed of flow.

**LOCOMOTION STUDIES**

The swing phase of walking begins, of course, at toe-off of one leg and ends at the next heel contact of the same leg. Swing phase plus the subsequent stance phase from heel contact to a second toe-off make up a complete cycle or sum of the two steps of left and right legs. The cadence of walking conventionally is expressed as the number of steps per minute.

Much can be learned by careful visual and auditory observation of walking. Individuals have characteristic gaits, often noticeable at a great distance or detectable from the sound of footfalls in a corridor before the individual appears. The lateral sway of a sailor's gait is proverbial; the high knee action and great ground clearance of a football player are portrayed in countless sports pictures every autumn. Gilbert K. Chesterton's mystery story, "The Queer Feet," hinges partially upon the auditory observation by Father Brown, the hero, that someone with the same pair of boots, making a small but unmistakable creak, walked alternately with the rapid light steps of a waiter and the slow, casual, careless emphasis of a gentleman of Western Europe.

More detailed observations of motions can be made by high-speed motion pictures or by "stick diagrams" (Fig. 1 a, b, and c.) Both normal (3) and pathological (4) gaits have been studied by these and many other techniques. The Bioengineering Laboratory of the Veterans Administration Prosthetics Center routinely studies motion of amputees and users of braces, orthopedic shoes, or other devices.

The muscle properties (5), the muscular activities affecting gait (3), and the associated energy costs (6, 7) have also been studied, though perhaps less extensively. A major goal of research in the near future is the further analysis of energy costs of various segments and the better understanding of the best interrelationship among motions to conserve energy of the body as a whole or of specific, perhaps weakened, segments such as the calf group of the remaining leg.

In the normal individual, of course, muscles under both voluntary and reflex control are available to maintain balance, to provide propulsion, and to attain a smooth pattern symmetrical between left and right sides. The individual spontaneously chooses, when free to do so, a comfortable gait and cadence requiring minimal energy consumption per unit of distance walked (7). Fortunately, as indicated by the
Interrupted-light “stick diagrams” of walking. Reflective marking tapes show successive positions of joints and center lines. (a) Normal individual (with nonrestrictive, freely moving single-bar “brace”); (b) Normal leg of unilateral above-knee amputee; (c) Prosthesis (with hydraulic knee control) worn by same individual.
shallow, broad U-shaped curve of energy consumption per unit distance when plotted against speed, considerable variation of speed is possible with only a slight change of energy consumption from the minimum. Presumably at or near this most comfortable speed and cadence there is minimal use of muscles with maximal reliance on gravity and pendulum action.

At speeds and cadences considerably below, as well as above, the optimum zone, adaptations of posture and gait are needed to prevent sharp rises in energy consumption per unit of distance walked. The normal person can rather easily make these changes. Muscles apparently must both absorb and produce mechanical energy, that is, they are used as brakes and as motors. The fibrous tissues within muscles are slightly springlike, but it is unlikely that major amounts of energy can be stored for later return.

Some major muscles or groups, e.g., the rectus femoris, the hamstrings, and the gastrocnemius, cross two joints. This arrangement conserves energy by allowing the two-jointed muscle to coordinate the motions of the two joints; that is, the single muscle, like a rope, may flex one joint while simultaneously allowing the other to extend. In contrast, two individual muscles would require energy, one to flex the first joint and the second to act like a brake while controlling extension of the second joint. In general, single-joint muscles parallel the two-joint. About the hip, for example, there are several muscles which can act as hip flexors in addition to the action of the upper end of the rectus femoris; about the knee there are the vasti as well as the lower end of the rectus femoris acting on the patellar ligament as knee extensors. An analysis of the particular cadences, positions on the Blix curve of tension versus length (8), and other circumstances under which these various possibilities are used might be revealing. Reserve capacity for emergencies, like the spare parts of the steam powerplant designed on the slogan “Performance first, efficiency second,” quite possibly is only one of the reasons for the presence of several muscles.

Muscular activity in the normal individual, schematized in Figure 2, involves quadriceps activity beginning late in stance to allow initiation of the knee flexion shown in Figure 1, yet prevent buckling. Then activity of the quadriceps group continues in the early portion of swing phase in order to limit maximum knee flexion and maximum heel rise. The normal individual allows only a small increase in heel rise for a considerable increase in speed. Sometimes the quadriceps are also used after maximum knee flexion to accelerate return swing. The quadriceps group again begins acting just at the end of swing
phase and even more vigorously continues immediately after heel contact to allow controlled knee flexion.

![Diagram](image)

**FIGURE 2.** Phasic action of quadriceps and hamstrings muscle groups. Vertical coordinates show integrated electromyographic activity in percentage of maximum. Average of 10 adult males studied at University of California during level walking at 95 steps per minute. Redrawn from Human Limbs and Their Substitutes.

These studies, made at a fixed cadence of 95 steps per minute, showed considerable individual variation, as in Figure 3. (Because some individuals might have had their most “comfortable” cadences somewhat above or below this arbitrary cadence, one may now surmise that varying amounts of muscular activity, often acting at different timing or phasing, were needed as brake or as motor to force the individual’s normal pendulum swing to conform to the required standard. Possibly greater uniformity of intensity and of pattern of phasing of electromyographic activity would result if the tests were rerun with each subject allowed to choose his own most “comfortable” cadence.)

The hamstrings act vigorously just before full knee extension and heel contact, slowing the swinging shank during this period of terminal deceleration. Just after heel contact the hamstrings still act to stabilize the normal knee against hyperextension and perhaps against torsional and mediolateral instability.

Artificial legs with present conventional means do not simulate the active functions of muscles as motors. Fortunately, the pendulum action of swing, transmission of energy from other parts of the body,
a springlike action, and a passive braking or absorption of energy can be provided rather simply. With proper adjustments, these relatively limited means permit surprisingly effective locomotion on reasonably level ground.

**THE PROBLEM**

For many years, conventional above-knee prostheses, even though lacking connections to the remnants of the quadriceps and hamstrings, nevertheless have permitted adequate stability by means of alignment during the stance phase. Originally, only a pendulumlike swing of the shank and foot about the knee bolt was possible during the swing phase. With the geometry and distribution of masses in the typical above-knee prosthesis for an average adult, the long period of this pendulumlike action led to a single relatively slow speed or cadence of walking. Adaptation to other speeds was very difficult, and the curve of energy consumption per unit distance versus speed was more sharply V-shaped than that for the normal.
The primary goal, numerous designers realized, was to allow the above-knee amputee to walk comfortably at a speed higher than that resulting from the natural frequency of the compound pendulum. Vaulting on the good foot, by vigorous plantar flexion, allows the swinging prosthesis to cover an appreciably longer distance from toe-off to heel contact, yet costs considerable energy. By vaulting, the amputee could walk at a somewhat higher speed in feet per minute but only at the expense of higher energy consumption, uneven timing of steps with left and right feet, and uneven lengths of steps (visible, for example, in footprints in the snow). All in all, vaulting is an ungainly solution to the problem of increasing his speed. As a secondary goal, designers wished to help the amputee to vary his cadence and speed at will, without frequent mechanical adjustment of his prosthesis.

Imitations of at least passive actions of muscles and tendons would seem helpful. Tendonlike passive straps, energy-storing elastics and springs, and motion-resisting friction brakes offer possibilities. A cord, strap, or other “back check” (Fig. 4) easily imitated the actions of the hamstring group and certain ligaments in limiting extension of the knee. Cords or “artificial tendons” were used at least as early as 1800 to control the mechanical ankle only or to coordinate ankle dorsiflexion with knee flexion during swing phase. Some cords were nearly inextensible, some were used in series with rubber blocks, and others were deliberately elastic.
Cords or straps have also been used to bring to the prosthesis energy, control signals, or both from other parts of the body which remained intact. Shoulder suspenders, for example, may not only support a prosthesis to prevent it from sliding from the stump during swing phase, but they may also transmit the energy and signals of shoulder elevation to help to extend the prosthetic knee joint, which is usually uncontrolled by the quadriceps group of the amputation stump. In a few instances, a cineplastic tunnel in the remnant of the quadriceps has been tried to tense a cord serving as an artificial patellar tendon. Possibly swing phase could be improved, or a voluntary signal might be given to a knee lock or other mechanism, but the tunnel has been inadequate for stumble recovery or stair-climbing step over step.

Occasionally, transmission of control and energy from the opposite side of the body or even from a source of auxiliary power, as in a mechanical doll or automation, has been suggested. Cords, straps, gearing, or linkages could be envisioned for straight, level walking, in a scissorslike gait. Such systems would need to be disconnected for other activities like sitting.

The prosthetic knee can be controlled, if only indirectly, by a newly learned pattern of motion of the hip and stump on the amputated side. Hip extension, for example, can snap the knee into full extension against the back check near the end of swing phase and press the prosthetic heel against the ground. This motion, though, is somewhat abnormal; it must occur with an earlier phasic activity differing from normal walking, and presumably it is not responsive to a normal reflex arc capable of nearly instantaneous yet subconscious corrections of errors. These limitations as applied to stance control will be discussed in expanded fashion in a later issue of the Bulletin.

The old peg leg with locked knee joint was under full control of the muscles controlling the normal hip joint on the amputated side, provided that the socket were fitted to the stump reasonably adequately and the suspension were sufficiently snug. Thus the amputee could point the distal tip of the peg leg almost as accurately and rapidly as he could move his stump. As Norbert Wiener (10) pointed out, the user at least enjoyed direct sensory feedback. Later, an artificial knee joint, such as that in Figure 4, free to flex and extend during swing phase, improved appearance and facilitated gait by reducing the need to vault on the good foot or to circumduct the prosthesis. This knee joint, however, deprived the amputee of the direct control, the greater ability to vary speed over a considerable range by voluntary muscular activity of the hip, and the considerable sensory feedback of the peg leg.
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It is not entirely clear from the literature as to when the possibilities became apparent of modifying the ordinary compound pendulum of the prosthesis to provide more positive control. A vertical spring in the thigh attached by a cord to a horizontal arm combined with the knee bolt was patented by Benjamin Franklin Palmer in 1849. An elastic knee-extension bias or “kicker stick” within the shank to act upon the knee joint has been available since Bly’s leg (1858), or perhaps earlier. Elastic straps on the anterior of the thigh, passing across and below the knee joint to attach to the shank, simulating the quadriceps to some extent, undoubtedly have been used for many years. Use of shoulder suspenders attached to straps acting on the shank directly or passing under a roller having a lever arm extending the knee has been mentioned repeatedly since 1866. The possibilities of deliberate friction about the knee joint seem to have been realized early in this century, perhaps more or less accidentally from previous attempts at providing adjustable bearings that could be clamped more tightly as wear occurred.

Until the time of the Civil War, though, most inventors rarely attempted to provide an elastic strap, “kicker,” or other means of imitating the action of the quadriceps muscle group to extend the artificial knee joint; indeed, they were proud of eliminating friction about the knee joint. Thus, the amputee was forced to walk with a pendulum action at close to a single and limited speed.

It is assumed that many, perhaps most, above-knee amputees used a cane or even a crutch. The Marquis of Anglesea, who lost his leg at the Battle of Waterloo and was fitted by Potts with a wooden artificial leg with dorsiflexion coordinated with knee flexion, is typically described as using a cane. Numerous amputees of the American Civil War era are portrayed with canes. Even as late as Muirhead Little’s book (11) about English experience during and immediately after World War I, the assumption was that at least one and perhaps two walking sticks or canes would be used by an above-knee amputee. Relatively slow speed in walking, use of energy from the arms, and the additional stability of tripod or quadruped gait seem to have been taken for granted.

ELASTIC KNEE EXTENSION BIAS

Elastic knee extension mechanisms or elastic knee extension bias means exist in many forms. The elastic is tightened as the knee is bent. An adjustment typically allows tightening if the elastic webbing or other spring becomes permanently stretched with use. In addition, the elastic means can be preloaded even in full knee extension, or conversely it can be adjusted to allow some range of free flexion.
before elastic resistance to further flexion is encountered; there are advantages with either extreme. Similarly, multiple steps of resistance or varying stiffnesses have been used, sometimes in relatively elaborate designs.

Such elastic means obviously simulate in a purely passive way the muscular action of the quadriceps group. There is, of course, no possibility of voluntary motor action or of a rapid reflex tightening action responding to sudden tendency to stretch the extension bias. In contrast, the patellar tendon jerk of a normal individual when struck by the neurologist’s hammer illustrates readiness not only to control normal knee flexion just after heel contact but ability to respond instantly to stumbling.

It is clear from Figure 5 that an anterior elastic strap, fastened to thigh and shank, is tightened by knee flexion and thus will tend both to resist knee flexion and to restore extension. Therefore, heel rise immediately after toe-off early in the swing phase will be limited by increasing tension in the extension bias. After maximum knee flexion and consequent heel rise have occurred, the elastic extension strap tends to force return of the shank and knee joint toward extension. To a certain extent this forward acceleration is a helpful action, because it permits the amputee to walk at higher speed than would be possible with the compound pendulum action of the prosthesis dangling from the hip joint. Merely as a free pendulum, there would be a fixed but relatively low frequency of oscillation of the shank swinging about the knee joint as the socket with the associated thigh portion pivots with the stump about the hip joint without muscular activity.

Increasing activity of hip flexors and extensors in an attempt to increase cadence would merely “crack the whip” noisily with the prosthesis with a free knee joint shown in Figure 4, leading to increased amplitude of flexion, slamming into full extension, and perhaps rebound into dangerous knee flexion before heel contact could occur. The knee extension bias of Figure 5, though, allows profitable use of some hip musculature to increase cadence to a limited extent while retaining control of the shank.

An attempt to gain still higher speed or cadence of walking, however, merely by excessive tightening of the elastic extension bias creates several difficulties. One problem is the tendency of the shank to continue to accelerate forward with greater and greater speed until it finally slams into full extension against the extension stop just before heel contact, with resulting audible noise, impact upon moving parts with consequent wear, possible rebound into unstable knee flexion, and transmission of uncomfortable shock to the stump of the amputee. A second difficulty is the risk that the toe, during
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the swing toward full extension, will be drawn forward so rapidly that it may even stub against the ground before it reaches the usual point of minimum, but appreciable, ground clearance. Toe stubbing, with the knee still bent, would be likely to lead to an accidental fall by the patient. In the extreme, a very stiff and tight extensor, by entirely blocking knee flexion, would force circumduction of the prosthesis in order to clear the ground. Because of the greater distance from hip to toe than direct line from hip to heel, the circumduction would have to exceed that required for a peg leg.

Another difficulty arises, which is due in part to the inherent construction of the artificial quadriceps-like extension strap; this difficulty is further increased by undue initial tension in the strap or an excessively stiff spring rate. When the patient tries to sit, very considerable flexion of the knee joint will create excessively high tension in the strap; this force tends to straighten the knee and move the artificial foot out into the aisle, creating a stumbling hazard to others as well as an awkward appearance of the amputee. This tendency is particularly pronounced with the suction socket prosthesis because the upper end of the strap is attached high on the anterior portion of the thigh. Thus the strap acts as a one-joint muscle, comparable to the vasti muscles. This tendency to extend the knee during sitting is less serious in the case of a user of a "conventional" type of prosthesis because the upper end of the strap can be attached to the pelvic band and thereby slackens when the hip is flexed during sitting. This strap location is somewhat analogous to the rectus femoris portion of the quadriceps originating at the pelvis.

Alternatively, other designs can be conceived which would provide high elastic knee extension bias at or near full knee extension but allow reduction in extension bias after the knee is substantially flexed. Such design is possible with an elastic quadriceps strap as in Figure 5, if, for example, the strap were attached to an inverted "Y" strap as in a below-knee prosthesis (Fig. 6) so that the straps at the two branches of the "Y" could slip around the knee joint, come closer to, and perhaps finally cross the knee axis as flexion increases.

A very commonly used knee extension bias consists of a roller attached to the shank but mounted ahead of the knee bolt so that tension in a leather strap passing through slots in the thigh piece and under the roller tends to extend the knee joint. This roller may be mounted, as in Figure 7, directly (or through a pivoted link) on a curved metal bracket which surrounds the knee bolt and is attached to the rear of the shank. The bracket also serves as a beamlike extension stop loaded by three forces from the knee bolt, a felt bumper in the bottom of the knee block, and the rear of the shank.
The roller-and-bracket design was widely used with shoulder suspenders (Fig. 8) both to suspend the prosthesis and to extend the knee by voluntary shoulder elevation. The user of a suspender-type limb had, at the expense of some discomfort from the suspenders and an awkward appearance of gait, at least the possibility of some sensory feedback of knee position and the additional possibility, through elevation of the shoulders, of voluntary tightening of the extension bias and thus of extension or additional stabilization of the knee joint. In recent years, though, the disadvantages of suspenders have generally been considered to outweigh the possible theoretical advantages for most amputees.

If this general design is used in a “conventional” prosthesis (Fig. 9) with pelvic band (a), a strap (d), which perhaps contains some elastic elements, may run from a posterior attachment to a pelvic band or belt down across the buttock area, down the posterior of the prosthesis, and through a slot into the knee portion to allow a flat leather strap (e) to pass under the roller, then up and out another slot in the anterior wall, and up to an adjustable elastic element (d) finally attached to the front of the pelvic band. The initial tension placed in the system can be varied, thus adjusting preload or initial free motion, if desired, and compensating for gradual stretching of the elastic webbing as necessary. An adjustment in total length of the strap also allows taking advantage of the concave upward curve of tension-
versus-length typical of rubber-based materials. Thus at relatively slack conditions, the tension increases only relatively slowly per unit of additional stretching, yielding a low spring rate. With tightening to higher preload, though, there is much greater increase in tension for an equal unit of stretching, resulting in much higher spring rate.

The posterior strap may be attached to the thigh, but the anterior end is attached to the pelvic band; thereby hip flexion during sitting slacks off the strap, allowing relaxation with the artificial knee flexed to a right angle.

**Figure 7.** "Central knee control" with upwardly loaded strap passing under roller on bracket attached to shank. Bracket also serves as extension stop.

**Figure 8.** Shoulder suspenders attached to cords (or strap) under roller of Figure 7 allow voluntary knee extension and presumably some sensory feedback of knee position. From Orthopaedic Appliances Atlas, Vol. 2, by permission.
If a suction socket is used, though, the posterior and anterior straps must be fastened to the thigh, so the strap simulates a single-joint muscle. This design is sometimes used even with a pelvic band, as in Figure 10.

Shape of the bracket and consequent relative location of the roller axis and the knee bolt will, of course, affect the stretching of elastic as the knee bends, the lever arm about the knee of the effective strap tension, and thus both the useful torques during swing phase and the inconvenient torque during sitting.
Another elastic extension bias mechanism which has frequently been used consists of a stout stick pivoted to the thigh about a point behind, and usually just below, the knee bolt with its lower end held in a leather envelope supported by an elastic strap or a tension spring within the shank. This so-called “kicker stick” or “hickory stick” (Fig. 11a) is usually also designed to serve as an extension stop. This system has the advantage that proper location of the pivot point within the thigh with respect to the knee bolt allows the upper end of the stick to pass under the knee axis (Fig. 11b) at an angle somewhat beyond the maximum knee flexion (about $60^\circ$) normally attained during swing phase. After passing dead center, the spring-loaded stick tends to hold the knee flexed rather than extended during sitting, keeping the foot safely out of the aisle.

Figure 11. “Kicker stick” provides (a) knee extension bias during swing phase but (b) passes under knee axis into sitting position. It also serves as extension stop.
Elastic extension bias about the knee joint can also be obtained with fluid-controlled mechanisms. As shown schematically in Figure 12a, a piston rod pivoted within the thigh behind the knee bolt, in a manner similar to the elastically supported "hickory stick," may be forced down into a cylinder, which is located within the shank, by flexion of the knee joint, as in Figure 12b. Thus if a compressible gas, such as air, were trapped within the cylinder, the pressure would be somewhat raised because a certain volume of the cylinder rod (represented by crosshatching) would thus be forced down into the cylinder, reducing the volume available for the gas. The gas, compressed in excess of atmospheric pressure, would then react against the net cross...
section of the piston rod, pushing it upward and thus tending to straighten the knee. In the sitting position, though, the pivot point at the top of the piston rod would have passed dead center so the upward force of the gas pressure on the piston rod would hold the knee in flexion.

If an incompressible hydraulic fluid were present, a volume equal to the intruding piston rod would have to be displaced into a reservoir or accumulator against an air chamber or more commonly a reservoir piston which in turn would compress a spring (Fig. 12c). Resistance of the spring against the reservoir piston or accumulator would thus

rod can be displaced into reservoir or accumulator against reservoir piston and spring, as symbolized by brackets. Either pneumatic or hydraulic unit exerts equal and opposite reactions (d) upward on pivot pin in thigh and downward on shank along the same line behind knee bolt, thus extending knee.
create a higher hydraulic pressure within the system, which in turn would be transmitted by the incompressible fluid back to act against the unbalanced part, the cross-sectional area of the piston rod. Thus, as in Figure 12d, a net force of pressure times cross-sectional area of the piston rod would be developed, which, for small angles of knee flexion, would react upward behind the knee bolt and thus tend to extend the knee joint. From another point of view, as shown also in Figure 12d, the bottom end of the cylinder would have an equal and opposite reaction downward against the pivot attaching it to the shank, on the same line behind the knee joint, also tending to extend the knee joint.

In the case of an incompressible fluid, as we have seen, the liquid displaced by the additional volume of piston rod entering the system must be displaced to some form of spring-loaded reservoir. The reservoir also compensates for expansion or contraction of oil volume with increase or decrease of temperature, allows preloading or static pressure even at full knee extension to give any desired initial extension bias, and stores spare fluid to make up over months or years of use for any slight leakage and for the almost inevitable loss of a minute amount by evaporation from the piston rod whenever it emerges after each wetting within the system. (Even very good seals allow some molecules to remain on the rod; sealed boots or bellows thus far seem impracticable, and conventional materials for piston rods are wetted by the usual hydraulic fluids.) Thus the reservoir has a series of diverse functions, which may require during design some compromises of size, spring rate, preload pressure, or other details.

The design details of the reservoir are irrelevant. Thus the reservoir might be on the outside of the main cylinder and piston as in Figure 12c, or might form an annular chamber surrounding the piston rod as in Figure 13. The cylinder itself might be surrounded by an annular reservoir. Even though the cylinder and the main piston may very well be much larger than the piston rod, the net pressure from the reservoir is exerted only on the unbalanced lower end of the piston rod, whereas the external cross section of the same piston rod is simply exposed to atmospheric pressure without the addition of the fluid pressure.

In some industrial compressors or pumps, an extension of the piston rod, called a tail rod, emerges to the atmosphere through an extra seal from the opposite end of the cylinder. This design eliminates the possibility of the elastic extension bias that would be useful in many prosthetic devices but possibly inconvenient in these industrial applications. (In some of the early experimental hydraulic legs for control of stance phase, developed at Hosmer and at Catranis, tail rods also were used even though extra seals were required.)
Elastic knee extension bias or "kicker" action may be further increased, if desired, by putting a spring on the exposed bottom end of the piston. If the spring were in contact with the cylinder floor at all times, however, it would simply serve to supplement the reservoir spring. If it is lifted free from the bottom of the cylinder near full knee extension and early in flexion (Fig. 14a), it will have no effect in that region. Later in knee flexion (Fig. 14b), the bottom of the spring will make contact with the floor of the cylinder and begin to resist further entry of the piston rod, further increasing the extension bias. Thus, elastic knee extension bias might be rather gentle in a first range of flexion but sharply stiffened near the point of maximum desirable heel rise, then reduced again as the shank starts to swing about the knee toward full extension. Again, the effect may be analyzed either in terms of force upon the piston rod, acting at a leverage behind the knee bolt, or through the equal and opposite reaction force upon the bottom of the cylinder acting upon the shank, again in the line of action behind the knee bolt.

Conversely, a method appropriate for hydraulic systems for supplying high extension bias only near full extension might allow open-
ing of a port after some initial piston travel, and thus permit escape of some of the oil from the reservoir; this action would allow the reservoir spring to expand, decrease its force, and consequently decrease the fluid pressure in the system. Such a system, like the fork or inverted-Y strap of Figure 6, is sometimes advocated to provide a high preload, which helps “point” the prosthesis with little knee flexion and to provide vigorous extension bias for short, nimble steps as in dancing or in moving about a work bench. At large knee flexion angles, and especially in sitting, the extension bias disappears or even becomes a flexion bias.

Figure 14. In extension (a) intermittently acting spring on bottom of piston is free and does not affect knee resistance. Approaching maximum knee flexion angle desirable for swing phase (b) spring makes contact with cylinder bottom, so its resistance to compression increases extension bias. (After upper pivot passes under knee bolt, however, both spring and reservoir pressure tend to keep knee flexed.)
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Thus far, attention has been concentrated upon the improvements in elastic knee extension bias which, of course, will limit heel rise and knee flexion early in swing phase and then encourage return of the shank and extension of the knee. Such elastic extension bias has the distinct advantage of tending to conserve energy, by storing it during a portion of the working cycle and later returning it, supplementing the effect of gravity as shown in Figure 15a and b. Excessive reliance upon knee extension bias, though, has the disadvantage of continuing to accelerate the shank forward even as the knee approaches full extension, thus tending to increase terminal impact; this results in jarring the stump of the patient and perhaps gives an audible thump. Use of the extra spring on the end of the piston rod, as in Figure 14, will reduce this tendency, whereas high knee extension bias at and near full knee extension will increase it.
THE ROLE OF FRICTION

Historically, as suggested above, legs were deliberately built with as little knee friction as possible, with gait thus dependent upon compound pendulum action. Early in the 20th century, however, some designers realized that mechanical friction might permit the amputee to walk more rapidly by voluntary muscular control of the stump and thus indirectly of the prosthesis, even though energy was obviously wasted in direct rubbing. The role of friction is discussed in some detail in Orthopaedic Appliances Atlas, volume 2 (12), from which Figures 15 and 16 are reproduced.

It is immediately apparent from Figure 16 that friction always opposes motion, first resisting knee flexion with consequent heel rise (to supplement gravity and any available elastic knee extension bias) in limiting the pendulumlike rearward swing of the shank as the amputee attempts to walk at high speed. After maximum knee flexion occurs, however, friction torque reverses to oppose extension and to reduce terminal impact, whereas extension bias would continue to act to accelerate the shank faster and faster toward full extension of the knee joint. A knee joint with properly adjusted friction might there-
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fore be used with beneficially higher elastic knee extension bias without thudding into full knee extension.

The net effect of directly discarding mechanical energy in a knee friction mechanism is not yet fully understood in terms of total energy consumption of walking. It seems conceivable that a little mechanical energy wasted in friction might allow conservation and return of a substantial amount of mechanical energy by greater use of elastic knee bias. An unbalanced self-energizing brake, providing somewhat more resistance to extension than to flexion, might prove even more valuable than equal friction in both directions. An unbalanced brake might allow less frictional resistance and wasted energy during flexion with, more desirably, greater reliance on a stronger nonlinear elastic knee extension bias to limit heel rise while storing energy for later return, even though somewhat more energy would be wasted during the extension motion.

Similarly, a stiff nonlinear spring, more elastic than the usual felt pad, has occasionally been used as an extension stop. It might absorb some of the energy of the swinging shank to provide terminal deceleration, prevent excessive extension during stance, then return some stored energy to initiate knee bending just before toe-off. The effects of such energy-conserving measures deserve theoretical analysis and experimental trials.

More importantly, both friction and elastic knee extension bias enable the amputee to walk more rapidly without wastefully vaulting the entire body over the good foot, circumducting the prosthesis, tilting the torso excessively, or jerking the stump or torso vigorously. Thus the presence of suitably adjusted mechanical friction allows the above-knee amputee to walk at a selected speed with minimal adaptations involving the rest of his body. The mechanical energy wasted in friction is probably small compared with the savings made possible in general metabolic energy consumption, although more analysis certainly would be desirable.

Any possible saving in general metabolic cost for the entire body would be welcome for any individual, particularly for prolonged walking, but this economy would be particularly important for patients with cardiac problems. Reduction of energy demanded even from specific muscle groups, like the calf of the remaining leg, might be crucial for an individual who already has either weakness or vascular involvement of the part.

As discussed in the next paper (1), in the static condition there is relatively high mechanical friction, sometimes called “stiction,” but then after breakaway the mechanical friction during sliding drops to a constant value, which is independent of the relative velocity of the sliding surfaces. The higher “stiction” aids high initial pre-
load of knee extension bias in moving the stiffly extended prosthesis short distances as an extension of the stump. The difference between high static friction and lower sliding friction enables a violin bow to set the string into vibration. If by chance this difference also transmits subtle vibrations from the swinging knee to the stump to provide some limited sensory feedback, no special effort seems to have been made to design the flexibility or impedance of prosthetic components to match the faint signals or to train the amputee to recognize and use the cues.

A simple mechanical clamp, as in Figure 17, can be added to an above-knee artificial-limb knee joint, with or without knee extension bias, to allow walking at a somewhat higher speed than would be possible with pendulum action alone. Unfortunately, any single setting of the friction will be most appropriate only for a single walking speed, so if the amputee wishes to walk still more rapidly, he must either tighten the friction clamp further or must tolerate unsymmetrical and wasteful irregularities in his gait, such as vaulting upon the good foot and waiting for the relatively slow pendulum swing of the
shank returning from excessively high heel rise (related to undue knee flexion), followed by unusually large forward swing into full extension, leading to heel contact at an unusually long step forward with the prosthesis.

As in mere pendulum action, a single value of mechanical friction is best suited for only a single (even though higher) speed; therefore adjustments of friction typically are provided. Generally, these adjusting means are screws accessible from the back of the knee block or even through the bottom of the knee cap in the interest of cosmetic appearance. These clamping screws may be adjusted by the prosthetist during initial trial in the limb shop or on occasional return visits for routine maintenance, but they can hardly be adjusted by the amputee himself. In fact, many prosthetists have believed professional judgment is needed to attain the best adjustment.

Figure 18. "Variable-friction" knee unit with one friction clamp (Section A-A) constantly engaged, but a second intermittently engaged, while approaching maximum heel rise and during terminal deceleration. From Human Limbs and Their Substitutes, by permission.
SHANK ANGULAR ACCELERATIONS:

- **NORMAL SUBJECT**
- **A.K. SUBJECT**
  - "FREE KNEE" VERY SMALL CONSTANT FRICTION
  - CONVENTIONAL KNEE MODERATE CONSTANT FRICTION
  - U.C. KNEE OPTIMUM ADJUSTMENT
  - NAVAL VARIABLE FRICTION KNEE OPTIMUM ADJUSTMENT

*Figure 19. Angular acceleration of shank versus time observed for normal subject and for amputees with various prostheses. Relatively smooth patterns of normal contrast with sharper peaks with prostheses. (Courtesy of Univ. California, Berkeley.)*
Several designs have been developed for adjusting the mechanical friction through the clothing by moderately inconspicuous motions while the amputee is on the street. Perhaps the best known of these designs is that of the Webb patent assigned to the British Desoutter firm (13). A notched thumb wheel in the front of the thigh was operable through the trouser leg.

Other attempts have been made to introduce a series of friction surfaces, engaged in two or more steps as knee motion increases. Thus knee flexion and heel rise might be increased only a little at higher speed, as in the other leg of the amputee and somewhat as in the normal person. A small amount of friction might be used during the initiation of knee swing, followed by an increasing amount at a larger knee angle approaching the maximum desired, so that this additional friction might help to decelerate the shank. Then, during the beginning of the return swing toward extension, the friction again would be low, followed by engagement of the additional amount shortly before full extension so as to aid terminal deceleration and thereby reduce terminal impact. One such design, developed by Oliver while at the Navy Prosthetics Research Laboratory, is shown in Figure 18. In recent years this basic principle has been applied experimentally by Northwestern University with a larger number of smaller steps of increasing friction, relying upon washers of different shapes.

The effects of normal muscles and some various approximations are shown in Figure 19. The top curve represents angular accelerations (upward) and deceleration or slowing (downward) of the shank of a normal subject, displayed conventionally against a horizontal axis representing steady travel of oscillograph recording paper, and hence time. It is relatively smooth. There are only a few tremors at toe-off and a rapidly damped oscillation or shock at heel contact. The gentle hill at maximum knee flexion and the broad valley at terminal deceleration imply smooth, graceful, economical motions under excellent control.

None of the artificial knees attains these smooth accelerations and decelerations of modest values provided in the normal individual by normal muscles under reflex control. All the prosthetic shanks sustained almost equally large and abnormal deceleration values. The University of California early experimental unit and the Navy variable friction knee joint, however, opened out the sharp canyonlike notch at terminal deceleration, which was characteristic of slamming into full extension with very small or moderate constant friction that occurred in conventional designs of constant friction about single-axis knee joints.
HYDRAULIC RESISTANCE

A basic characteristic of a fluid is rapid increase in resistance to flow with only moderate increase in velocity of flow. Mathematically, resistance is directly proportional to velocity for orderly layerlike viscous flow and to the square of the velocity for the more common turbulent flow. This latter characteristic would seem ideally suited to the problem of providing additional friction at higher speeds of walking. In fact, Mauch (14) showed that, for certain assumptions, knee friction or resistance proportional to the square of the speed was exactly what was needed to allow the amputee to vary cadence.

All types of fluid tend to resist flow, whether through sharp-edged orifices, through valves or faucets, or through long ducts or passages, so a pressure drop is required to cause flow. Resistance to flow depends upon a number of factors: properties of the fluid itself (often dependent on temperature), the channel through which the fluid flows, and the speed. These factors are considered in more detail in the subsequent paper (1). Intuitively, we all realize that thick lubricating oil flowing through a long, narrow channel with rough walls at high speed will require far more pressure drop than thinner kerosene flowing through a shorter, wide channel with smooth walls at lower speed. The designer therefore can manipulate numerous design details.

A relatively simple type of fluid mechanism can be illustrated schematically in Figure 20. As the knee is flexed, the piston rod is forced down into the cylinder, with the additional volume of the entering piston rod compensated for by compression of the reservoir spring with consequent increase in elastic extension bias, as we have already seen. In addition, though, the downward motion of the piston displaces the operating fluid from below the piston through a bypass line around to the top of the piston. An adjustable valve in this line allows initial adjustment of resistance to flow during trials on the individual amputee. In addition, this basic resistance automatically varies with the square of the speed of the turbulent flow typically encountered at practicable walking speeds, so that double the flow produces four times the resistance, triple the speed of flow produces nine times the initial resistance, etc.

We are all familiar with this basic principle in connection with hydraulic doorclosers. After the door is opened, the elastic extension bias from a strong spring tends to close the door at a rate that has been adjusted to minimize drafts yet to avoid slamming of the door. If we attempt to assist the spring to push the door closed even more rapidly, we encounter greatly increased resistance.
With the design shown in Figure 20, resistance to flexion and to extension is interrelated because the same bypass valve is used to regulate flow in both directions. During extension of the knee, the piston is pulled upward so that the fluid in the upper chamber (and any temporarily stored in the reservoir) must be displaced back through the same valve to the bottom chamber. Under many circumstances such an interrelationship between flexion and extension may seem adequate, because it is such a marked improvement over constant mechanical friction. After all, more rapid walking tends not only to produce increased heel rise and knee flexion but also to produce violent slamming into extension, both of which could be resisted by spontaneously increased friction. On the contrary, during slow, stately walking as in a procession, or for sauntering or window shopping, very little resistance to heel rise is desirable so that, immediately after maximum heel rise, adequate clearance by the swinging foot will be obtained at the point of closest approach to the ground. Likewise,
at slow cadence there should also be very low resistance to extension so that full extension will be attained just before heel contact to allow a conventional prosthesis to attain adequate alignment stability immediately at the beginning of the stance phase. Again, the automatically diminished hydraulic resistance at very slow speed is at least in the correct direction to improve both flexion and extension parts of the swing phase compared with constant mechanical friction.

A somewhat more sophisticated design, however, can be obtained by providing separate bypass tubes or passages and independently adjustable valves for controlling flexion and extension as in Figure 21. In each bypass some form of check valve, shown schematically as a ball, is closed against flow in the opposite direction so that for either motion only one line is operating. The action of these check valves may be compared to the hinging of entrance and exit doors, each permitting easy opening in the direction for proper flow of traffic yet closing under pressure from the opposite direction. Ball-and-cone, flapper, or other valves can be used, or various types of flexible cup seals can be either deformed to allow flow in one direction or expanded by fluid pressure to close the passage against flow in the opposite direction.

![Figure 22. Fluid mechanism with multiple channels in each direction allows high resistance near end of motion even though fluid is flowing slowly, because moving piston successively cuts off entrances to escape channels. Basic principle is shown, highly schematized, in enlarged diagram on right (with reservoir and check valves omitted for clarity). In early flexion, descending piston can force oil out all five bypass channels; later the bottom edge cuts off channels until only the lowest remains open. Relative sizes, numbers, and locations of channels can be assigned by the designer. Valve allows adjustment during field use.]

An inherent limitation of the relatively simple systems shown in Figures 20 and 21 is the provision of greatest friction during the period of most rapid fluid flow, i.e., approximately during the middle of the flexion motion toward heel rise and again during the middle of the
return swing of the shank toward extension. (The exact relationship between piston motion within the cylinder and knee flexion is somewhat affected by the relative positions of the knee bolt, the pin supporting the top of the piston rod, and the pivot which allows the cylinder to rock slightly within the shank. With designs which have thus far seemed practicable, there is a close correlation between speed of knee flexion and speed of motion of the piston.)

With the design of Figures 20 and 21, the oil would be flowing quite slowly when the prosthesis would be approaching desirable maximum heel rise and again approaching full extension, and the regulating valve would thus impose relatively low resistance when resistance is most needed.

A still more sophisticated design (Fig. 22) may be attained by allowing the moving piston to close off, one by one, the escape ports through which the oil can flow to the other side of the piston. A great bulk of oil escapes through a number of ports operating in parallel with relatively little resistance while the piston is moving at high speed, but as the piston slows toward the end of its motion, fewer and fewer ports are available for the remaining oil and thus greater resistance per unit of piston motion is obtained. The overall resistance can be further adjusted by some form of valve. In addition, the size, number, and location of the escape ports along the cylinder wall can be different in the flexion and extension directions, and the exact arrangement can be adjusted to match the designer’s concept of the most desirable friction characteristics in the various portions of the swing phase. All the existing commercial or experimental types of fluid-controlled knees based on the principles shown in Figures 20, 21, and 22 emphasize the primary role of hydraulic friction. These designs also provide elastic extension bias, but with somewhat lesser concern for its role.

**PNEUMATIC UNIT**

A somewhat different concept, emphasizing elastic bias with lesser concern over velocity-dependent frictional resistance, can be applied in a pneumatic unit. The basic design (Fig. 23) is similar to Figure 21. Since air is compressible, though, no special reservoir is needed to compensate for the entry of the piston rod. Air dashpots have been suggested since 1901.

The action may be studied from a graph of moment or torque about the knee axis in relation to knee angle, as in Figure 24. If the flexion valves were tightly shut, downward motion of the piston would simply compress the air trapped in the bottom chamber, as in ordinary tire pumps, commercial air compressors, or air springs. Air pressure, at practicable walking speeds, would rise considerably more rapidly than directly proportionally to piston motion. The relative geometrical lo-
cations of the knee axis and the piston and cylinder pivot points, of course, will affect both the piston motion and the changing lever arm for air pressure on the piston, with line of action along the piston rod, to react on the knee block to develop torque about the knee axis. In general, though, with the flexion valve entirely closed, the torque would rise in a concave upward curve as indicated by the top curve (A) of Figure 24, and the return to extension would approximately follow the same curve. If the flexion valve were tightly shut, then, there would simply be an air spring representing a considerable amount of elastic extension bias, sharply limiting heel rise but tending toward a considerable thud at full extension, without terminal deceleration.

![Figure 23. Pneumatic unit with independent regulation of flexion and extension leakages and hence air-spring resistances. No reservoir is needed because air is compressible.](image)

If, however, the valve regulating flexion resistance is partially opened, there is a slow leak of gas to the other side of the piston. It happens that resistance to airflow increases more rapidly than increase in speed, though possibly not as rapidly as the square of the speed as would happen with turbulent liquid flow.

Thus at slow speeds of flexion and a partially open valve, there is only modest resistance to airflow; therefore, much of the displaced
air escapes to the other side of the piston without much compression, so the torque reacting on the knee is less than that of an air compressor. As the piston moves faster, however, less of the air can escape, more resistance is encountered, air pressure builds up, and the air spring or elastic extension bias action becomes more noticeable, increasing resistance to flexion and thus to heel rise. Eventually, though, near maximum desirable heel rise, the piston moves less rapidly, finally stops, and then begins to reverse. Meanwhile air under relatively high pressure has a considerable opportunity to leak away to the other side of the piston, reducing the pressure of the compressed air and thus automatically reducing the elastic extension bias.

![Theoretical Knee Moment vs. Angle Relationships](image)

**Figure 24.** Theoretical relationships of torque or moment about knee axis to knee angle for pneumatic units similar to Figure 23.

As the knee begins to return toward extension, the extension bias rapidly decreases as the compressed air in the lower chamber not only continues to leak through the regulating valve but also, with upward piston motion, re-expands. Soon the air pressures above and below the piston are equal; there is no further upward flow through the passage on the right side of Figure 23, so its check valve shuts. At that instant of balanced pressures there is no further elastic extension bias, or, in Figure 24, the heavy curve crosses the horizontal axis.
Then, with continued swing of the shank toward extension and consequent upward motion of the piston, pressure in the upper chamber increases over that in the lower chamber. If the extension-regulating valve in the bypass on the left side of Figure 23 now were shut tightly, pressure in the upper chamber would increase as in an air compressor (and that in the lower chamber would decrease as in a vacuum pump if, as indicated in Figure 23, there were no communication with the outside atmosphere).

In general, though, the extension-regulating valve will also be partially open, so air pressure in the upper chamber in excess of the lower will blow open the second check valve and cause air leakage through the regulating valve to the lower chamber. Again an air spring action is built up, with increasing compression of air in the upper chamber but also with increasing tendency for air to leak away. As the piston slows during the terminal deceleration until it approaches the extension stop and the knee approaches full extension, the compressed air once again has an increasing opportunity to escape until the pressures on both sides of the piston again are equalized.

If the amputee attempts to walk more rapidly, these effects increase, giving him automatically some increase in resistance to heel rise and to thudding into full extension. Because resistance to airflow increases at an exponent somewhat less than the square of the velocity, increase of resistance with cadence presumably is not ideal, but perhaps the relationship is sufficiently close to meet the needs of many individuals over a reasonably wide range of practical cadences.

ADJUSTMENT OF EXTENSION STOP

Termination of swing phase and alignment stability during stance phase depend to a considerable extent upon appropriate adjustment of the extension stop. Basically there is a triangle composed of thigh axis, shank axis, and a third side which is adjustable in direct or indirect manner to permit initial alignment and to compensate for any changes during wear.

Various designs have been used. Cords may be adjusted in length. The thickness of the padding of the contact point between the knee block and a “kicker stick” may be altered. In the usual fluid-controlled mechanisms the piston can only be pulled upward to a certain point, so the knee extension stop may be adjusted either by varying the effective length of the nearly vertical piston rod or by moving anteriorly or posteriorly in the shank the attachment point of the lower end of the cylinder.
MURPHY: Swing Phase of A/K Walking

DEVELOPMENT OF FLUID-CONTROLLED MECHANISMS

The obvious and considerable virtues of fluid-controlled mechanisms for providing smooth control of the artificial knee joint over a wide range of cadences have evolved only through a long history of development. Although leaks, hydraulic swishing or gurgling noises, and clicks of mechanical parts have long plagued designers, several units have reached widespread use. A later paper in this issue discusses clinical application studies of some of the actual mechanisms built on the principles outlined above (2).

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