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Gait Patterns of Above-Knee Amputees Using Constant-Friction Knee Components[a]

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ABSTRACT

Quantitative measurements of multiple simultaneous gait components for both free-speed and fast walking of 10 active above-knee amputees with constant-friction knee components are presented and compared with standards of normal variability. In addition, interrelationships between the various gait abnormalities are identified.

INTRODUCTION

Constant-friction knee components for above-knee amputees have been on the market for many years and are still prescribed and worn by many amputees. Several advances in artificial knee components have been made since the development of the constant-friction knee, and future designs of artificial limbs will most certainly improve on those available now.

Other investigators have documented selected aspects of the gait of above-knee amputees (1–10). This paper documents several previously unreported patterns of motion of the lower limbs and the trunk of above-knee amputees using prostheses with constant-friction knee components. In addition, the paper describes relationships between simultaneous displacement patterns of multiple body segments during walking with this type of prosthesis.

It is hoped that this information will contribute to a deeper understanding of the pathomechanics of amputee locomotion, and will ultimately result in improvements in the design and prescription of artificial limbs and in the rehabilitation process of the amputee.

METHODS

The gait of 10 men with unilateral above-knee amputations was studied. In all cases the amputation was the result of trauma. The average age of these men was 41 years (range 33 to 48 years); the height, 177 cm (range 169 to 187 cm); and the weight, 79 kg (range 69 to 85 kg). Each man had worn a prosthesis with a constant-friction knee component for several years and wore the limb every day. Half of the amputees had suction suspension and the other half had pelvic band suspension. Fit and alignment were checked and considered satisfactory. Six had a SACH foot, and four had a single-axis ankle and either a wooden foot or a wooden foot with compressible heel and toe.

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Six had an extension aid on the prosthetic knee and four did not. The constant-friction device in each case had been set for free speed, and was not altered. None of the amputees had any other concomitant disability.

Reflective targets were secured to specific anatomic landmarks and to the prosthesis. An interrupted-light technique was used to photograph the serial displacements of the targets in the sagittal and transverse planes as the men walked at a comfortable speed (free speed) and at a fast speed (11). The target images were photographed at a rate that varied from 15 to 30 exposures per second, depending on the walking speed of the subject.

The following measurements were made from the photographs: velocity; stride and step lengths; stride widths; foot angles; stance, swing and double-limb-support durations; patterns of sagittal rotation of the shoulder, elbow, hip, knee and ankle; patterns of vertical, lateral and forward motion at the head and neck; and patterns of vertical displacement of the heel and toe. To quantitate irregularity in forward progression, a velocity index was calculated: this index was the ratio of the forward velocity during prosthetic swing to the forward velocity during sound swing. Thus a velocity index of less than 1.00 indicated slower forward movement during prosthetic than during sound swing.

The measurements of the amputees were compared to measurements of normal men (12–14). A t-test was used to assess the significance of the differences between the measurements of the amputees and the normal men.

RESULTS

Stride dimensions and temporal components

Table I contains the means and standard deviations of the measurements of velocity, stride dimensions, and stance and swing durations for the amputees, and indicates the significance of the differences between the measurements of the amputees and the normal men. The mean walking speed of the amputees was about two-thirds that of normal men for both free-speed walking and fast walking. The amputees' subnormal velocity was mainly due to their longer cycle durations (thus slower cadences) although their stride lengths also tended to be subnormal. This is illustrated in Figure 1, which shows that the stride lengths of most of the individual amputees were within the limits of normal variability, while the individual values for velocity and cadence tended to be near

TABLE I
Velocity, stride dimensions and temporal components

	FREE (Mean ±1 S.D.)	FAST (Mean ±1 S.D.)
Walking speed (cm/sec)		
Amputees	100 ±16*	140 ±25*
Normal men	151 ±20	218 ±25
Cycle duration (sec)		
Amputees	1.38 ±.11*	1.18 ±.11*
Normal men	1.06 ±.09	.87 ±.06
Cadence (steps/min)		
Amputees	87 ± 7*	102 ± 9*
Normal men	113 ±10	138 ±10
Stride length (cm)		
Amputees	136 ±15**	164 ±22†
Normal men	156 ±13	186 ±16
Step length (cm)		
Sound	64 ±9*	81 ±12**
Prosthetic	72 ±8	83 ±11†
Normal	78 ±7	93 ± 9
Stride width (cm)		
Amputees	17.4 ±3.6*	18.3 ±4.3*
Normal men	7.7 ±3.5	9.1 ±4.1
Foot angles (deg)		
Sound	10.9 ±7.2	10.2 ±5.8†
Prosthetic	2.3 ±5.0	2.1 ±4.6
Normal	6.3 ±5.7	5.3 ±5.5
Stance Phase (sec)		
Sound	.94 ±.12*	.78 ±.09*
Prosthetic	.80 ±.07*	.62 ±.08*
Normal	.65 ±.07	.49 ±.05
Swing Phase (sec)		
Sound	.43 ±.04	.41 ±.04†
Prosthetic	.58 ±.06*	.56 ±.04*
Normal	.41 ±.04	.38 ±.03
Double-Limb Support (sec)		
1st in cycle	.20 ±.06**	.11 ±.03**
2nd in cycle	.17 ±.04**	.10 ±.04†
Normal	.12 ±.03	.06 ±.03

*The amputees were significantly different from normal ($p<.001$); **($p<.01$); †($p<.05$)

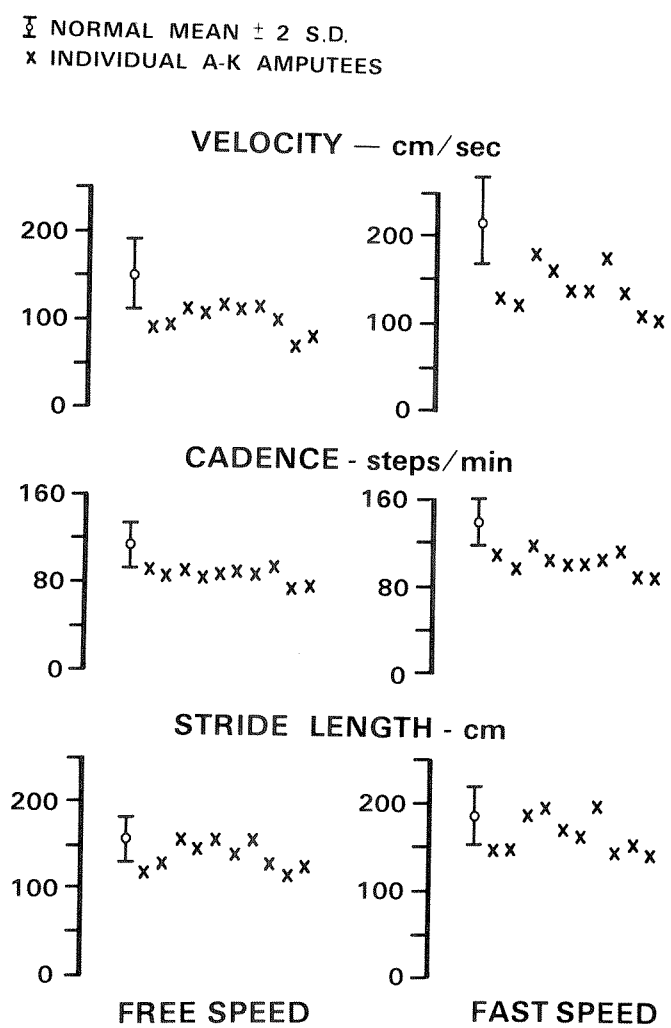

FIGURE 1.
The free-speed and fast-walking velocities, cadences, and stride lengths of 10 individual amputees are shown by "x"'s. The vertical bars bisected by a hollow circle represent the mean and 2 standard deviations above and below the mean for 30 normal men.

the lower threshold of normal variability. Stride length (expressed in percent of stature) averaged 77 percent for the amputees as compared to 89 percent for normal men during free-speed walking and averaged 93 percent for the amputees as compared to 106 percent for normal men during fast walking.

In contrast to the walking of normal men, whose successive step lengths are remarkably equal in length, the successive step lengths of the amputees were uneven during free-speed walking, and to a lesser degree during fast walking (Table I). At free speed all of the men—and at fast speed most of the men—took a longer step when the prosthetic limb was directed forward.

The stride widths of the amputees were abnormally wide. Their foot angles were different from those of normal men in that, on an average, their sound foot was in about eight degrees more outtoeing than the foot on the side of the prosthesis (Table I).

In normal men, the successive stance phases of the right and left limbs are approximately equal in duration as are the successive swing and double-limb-support phases. In the amputees at both walking speeds, the duration of the sound stance phase was longer than the duration of the prosthetic stance phase, but the duration of the sound swing phase was shorter than the duration of the prosthetic swing phase. The average inequalities in successive stance and swing phases, as expressed in percent of the walking-cycle time, are graphed in Figure 2.

Displacement patterns of the lower limbs

The mean patterns of hip flexion-extension are shown in Figure 3. Unlike normal, the pattern of motion of the hip on the prosthetic side was characterized by immediate and continuous hip extension beginning with the onset of prosthetic stance. In addition, the amputees' hip pattern on the prosthetic side was abnormal in that there was a sharp rather than gradual reversal from extension to flexion midway through the walking cycle, and this sharp reversal occurred just after sound heel-strike. These two characteristics were observed in 8 of the 10 amputees. The average total excursion of hip motion used on the prosthetic side was slightly less than that used on the sound side, and the residual hip of the amputees usually did not extend as much as the sound hip. The pattern of motion of the sound limb on figure 3 and subsequent graphs is shifted to the right compared to the pattern of normal men because the stance phase of the sound limb comprised 66 to 69 percent of the cycle duration, depending on walking speed, as compared to 57 to 61 percent for normal men.

The mean patterns of flexion-extension of the knee are shown in Figure 4. The patterns for the prosthetic knee illustrate two abnormalities characteristic of all of the amputees: the absence of the yielding knee flexion that normally occurs during the early stance phase; and secondly, continued knee extension during the last 5 percent of the cycle, particularly at fast walking, when knee ex-

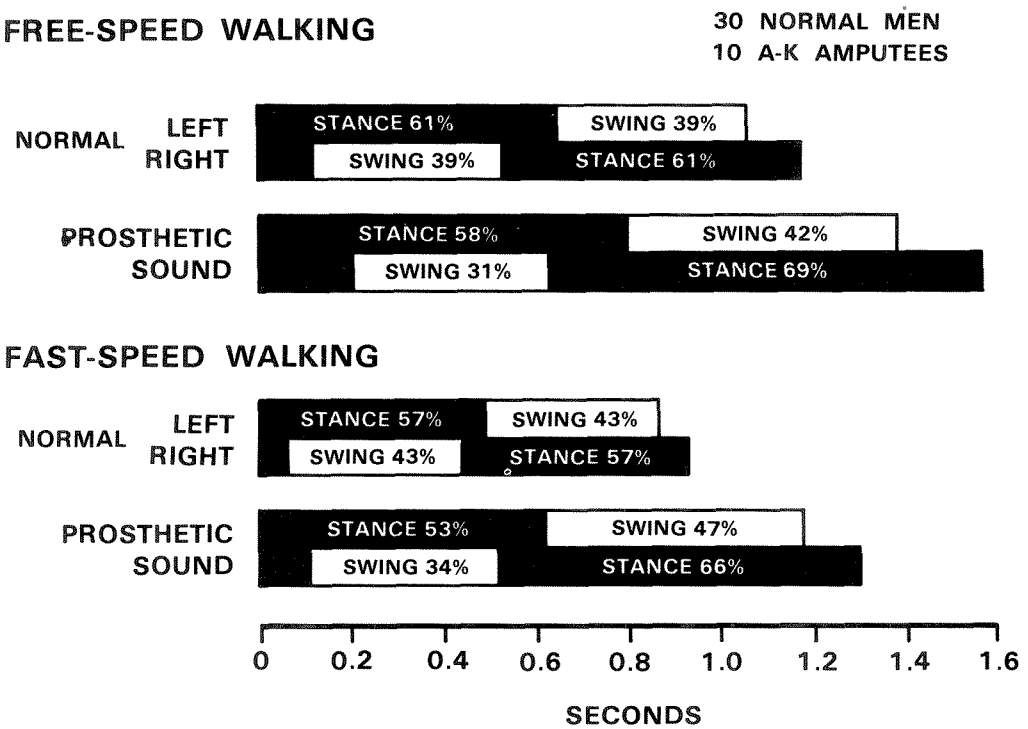

FIGURE 2.
Average stance and swing durations during free-speed and fast walking for the sound and prosthetic limbs of 10 above-knee amputees with constant-friction knee components (which are adjusted for free speed) and for the left and right limbs of 30 normal men. The numbers within the timing bars express the stance and swing durations in percent of the walking cycle durations.

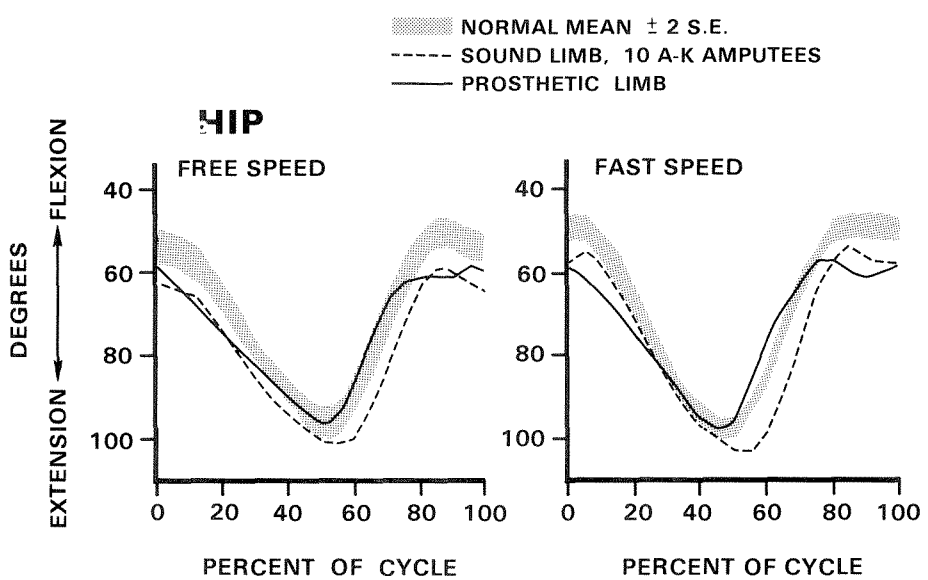

FIGURE 3.
Mean patterns of hip flexion-extension for the sound and prosthetic limbs of 10 above-knee amputees with constant-friction knee components, during free-speed and fast walking. The shaded areas represent 2 standard errors above and below the mean for 30 normal men. The ordinate scale values represent the anterior angle between the lateral thigh target (from hip center to knee center) and a lateral pelvic target (from anterior superior iliac spine to posterior superior iliac spine).

This and other graphs originate at foot contact.

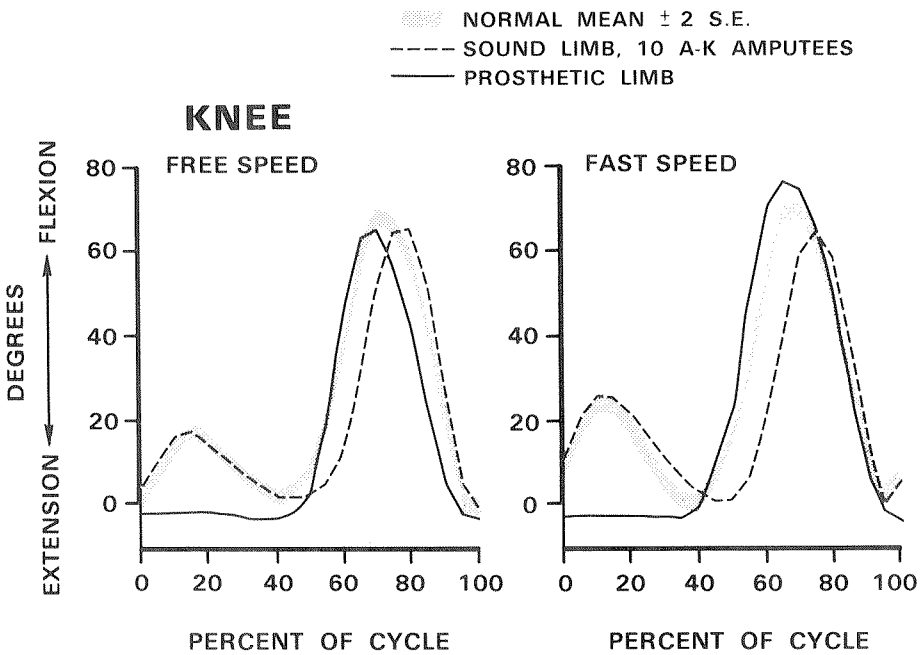

FIGURE 4.
Mean free-speed and fast walking patterns of knee flexion-extension for the sound and prosthetic limbs of ten above-knee amputees with constant-friction knee components. The shaded areas represent two standard errors above and below the mean for thirty normal men. Zero degrees represents full knee extension.

tension normally is decelerating preparatory to heel-strike. In addition, the mean pattern of the prosthetic knee shows near-normal flexion during the swing phase at free-speed walking—but excessive flexion during the swing phase during fast walking. That excessive knee flexion during swing at fast walking was not characteristic of all of the amputees but instead was related to the relative amount of friction of the knee component, to the speed of reversal from hip extension to flexion, or to the adjustment of the extension aid.

Figure 5 shows the mean patterns of the vertical trajectories of the heel and toe during free-speed and fast walking. During free-speed walking, peak heel rise was similar for the sound and prosthetic limbs. During fast walking, the average patterns show that prosthetic heel-off occurred earlier than normal and peak prosthetic heel rise was excessive. The toe-floor clearance distance was also excessively high for the prosthetic side during the swing phase, particularly during fast walking. In addition, the descent of the toe after heel-strike was slower than normal for the prosthetic side.

The mean patterns of dorsiflexion and plantar flexion of the sound ankle of the amputees as compared to those of normal men are shown in Figure 6. During free-speed walking the amputees used less than normal dorsiflexion during the sound stance phase. During fast walking the mean pattern was different from normal in that there was an abnormally early reversal into plantar flexion during the mid-stance phase. This type of ab-

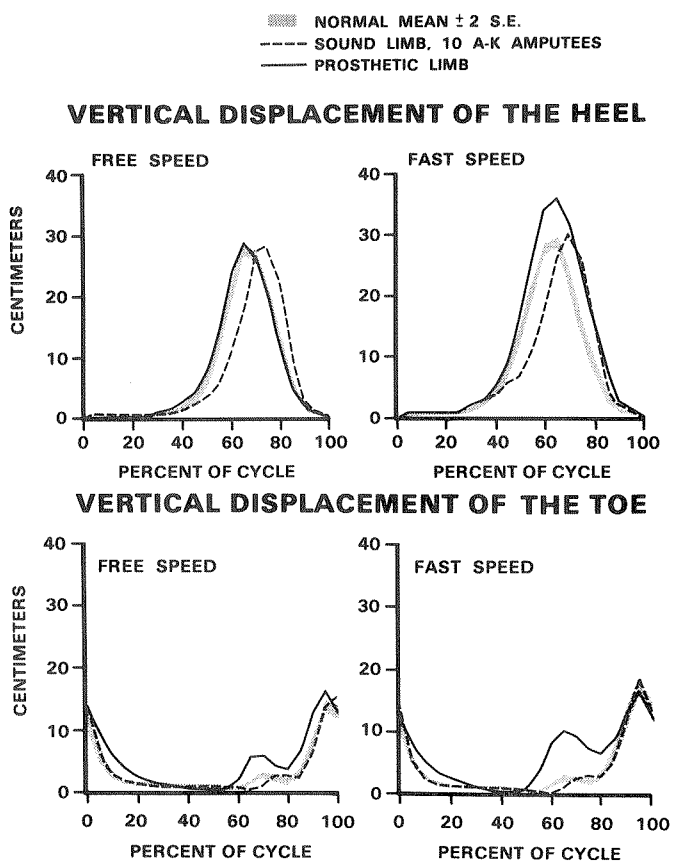

FIGURE 5.
Mean vertical pathways of the heel and toe of the prosthetic and sound limbs during free-speed and fast walking for ten above-knee amputees with constant-friction knee components. The shaded areas represent two standard errors above and below the mean for the thirty normal men. Zero centimeters represents the level of the floor.

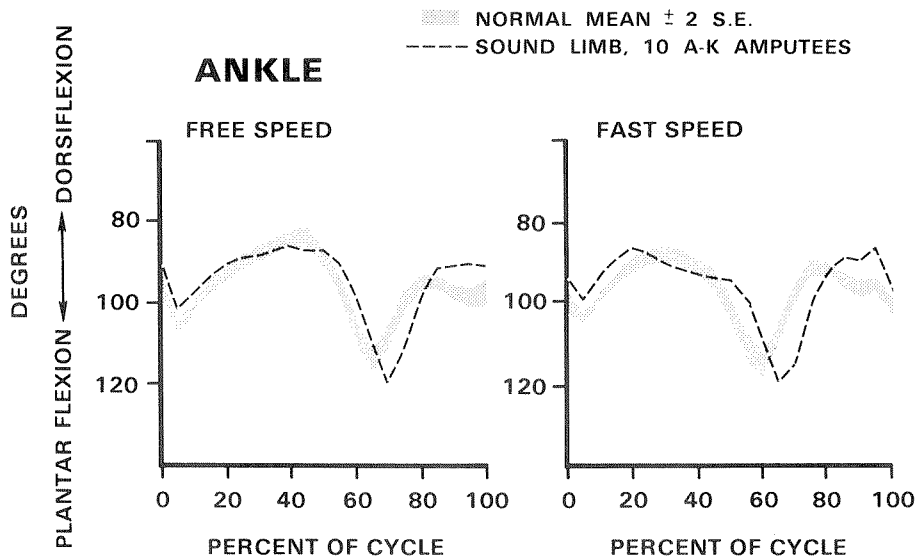

FIGURE 6.
Mean pattern of dorsiflexion and plantar flexion of the sound ankle for ten above-knee amputees during free-speed and fast walking. The shaded area represents two standard errors above and below the mean for thirty normal men. The ordinate scale values represent the anterior angle between targets on the lateral leg and heel of the shoe.

normality was observed in seven of the amputees during fast walking and in two during free-speed walking. Excessive plantar flexion at the end of the stance phase was observed at both walking speeds, and was typical of the sound ankle of all of the amputees.

The arm swing patterns of the amputees varied considerably from one subject to another. In general, the arm swing of the amputees was abnormal in that there was usually less shoulder and elbow motion on the prosthetic than on the sound side. There were seldom step-related reversals of the direction of motion at the elbow. The diminished excursions of arm swing by the amputees were mainly due to subnormal flexion at the shoulder and subnormal extension at the elbow for most amputees.

The patterns of vertical, lateral and forward motion at the head are illustrated in Figure 7. The patterns of vertical displacement for the amputees show asymmetrical peaks rather than the two symmetrical peaks typical of normal walking. The peak during prosthetic stance was subnormal at both walking speeds, and the peak during sound stance (prosthetic swing) was in excess of normal during fast walking. The amputees had excessive lateral deviation of the head and trunk toward the prosthetic side during prosthetic stance, and this lateral deviation was more pronounced during free-speed than fast walking. For free-speed walking the mean lateral excursion measured at the head was 11.6 ±1.0[b] cm for the amputees as compared to a mean value of 5.9 ±0.3 cm for normal men of comparable ages. For fast walking, the mean excursion of lateral motion measured at the head was 10.2 ±0.7 cm for the amputees as compared to 5.0 ±0.4 cm for normal men. For normal men, the forward displacement pattern indicates that forward motion as measured at the head and neck is remarkably uniform. In contrast the amputees moved forward more rapidly during the prosthetic single-limb-support phase than during the prosthetic swing phase.[c]

In Table 2, numerical data are given for the measured gait factors which relate to smoothness and symmetry in walking. All of the measurements of the amputees were significantly different from normal. Increases in speed from free-speed to fast walking were accompanied by improvement in the ratio of successive step lengths but by greater inequality in successive vertical excursions of the head. Both the swing ratios and the irregular forward motion of the head (velocity index) were similar for both free-speed and fast walking.

DISCUSSION

Study of the data presented suggests that the gait abnormalities of the amputees are ones which ensure safety and compensate for the problems inherent in walking with an artificial limb. Although these compensatory maneuvers seem to serve an important function, the maneuvers per se must increase the mechanical energy cost of walking for the amputee. While the exact relationship between mechanical and metabolic energy ex-

[b]One standard error of the mean.

[c]Forward displacement in Figure 7 represents variation about the average forward velocity.

TABLE 2
Measurements of smoothness and symmetry in walking (mean ±1 standard error)

Gait measurements	Free-speed		Fast walking	
	Amputees	Normal	Amputees	Normal
Swing ratio (Sound/Prosthetic)	.73 ±.02	1.03 ±.02	.73 ±.02	1.02 ±.02
Step Ratio (Prosthetic/Sound)	.88 ±.03	1.00 ±.01	.97 ±.03	.99 ±.01
Velocity Index*	.86 ±.023	1.00±.003	.87 ±.016	1.00 ±.006
Inequality in Successive Vertical Excursions of the Head (cm)**	1.9 ±.4	0.0 ±.1	3.0 ±.5	.1 ±.1

*Average forward velocity during prosthetic swing divided by average forward velocity during sound swing for amputees.
**For amputees the numbers indicate higher average vertical oscillations during prosthetic swing than prosthetic stance.

penditure has not been established, increased mechanical energy expenditure is probably related to increased metabolic energy expenditure. Waters et al. (15) found that the mean net metabolic energy cost of walking for a group of young above-knee amputees was higher (.25 ±0.05 ml / kg•m) than for normal subjects (.159 ±.032 ml / kg•m) in the same age range. The amputees in their study all had amputation as a result of trauma.

As pointed out by Saunders, Inman and Eberhart (16), the wheel provides the most economical means for translation of an object in a straight line with the least expenditure of mechanical energy. However, for bipedal locomotion, the most economical means of translation is a symmetrical gait which results in smooth gradual deflections of the center of gravity of the body through sinusoidal pathways of low amplitude. These gradual deflections of low amplitude require a minimum of accelerative and decelerative forces. Although we did not measure the pathways of the center of gravity of the body in our previous studies of normal subjects, our measurements of the lateral, vertical, and forward trajectories of the head and neck of normal subjects during locomotion demonstrate smooth sinusoidal pathways of low amplitude (12) remarkably similar to the configuration of the pathways of the center of gravity measured for a normal man by Fischer (17). In contrast, the methods of compensation used by amputees resulted in irregular lateral, vertical, and forward pathways as measured at the head and neck. These changes must impose a higher mechanical energy expenditure than normally required for locomotion, since abnormally high accelerative and decelerative forces are required to produce and restrain the motion.

According to Fischer (17), the mass center of the body deflects laterally twice during each walking cycle toward the supporting limb during single-limb-support, with a total excursion of about 2.5 cm during free-speed walking and a greater amplitude with slower walking speed. In previous studies comparing free-speed and fast walking of normal men, the total lateral excursion as measured at the head and neck averaged 5.9 ± 1.7 cm for free-speed walking and 5.0 ± 2.1 cm for fast walking (12). The amputees in this study averaged about twice this amount.

It is not unusual to find slightly excessive lateral motion of the torso even among amputees who walk well. There may be several explanations for this. First, it is well known that there is a relationship between lateral lurching and insufficient func-

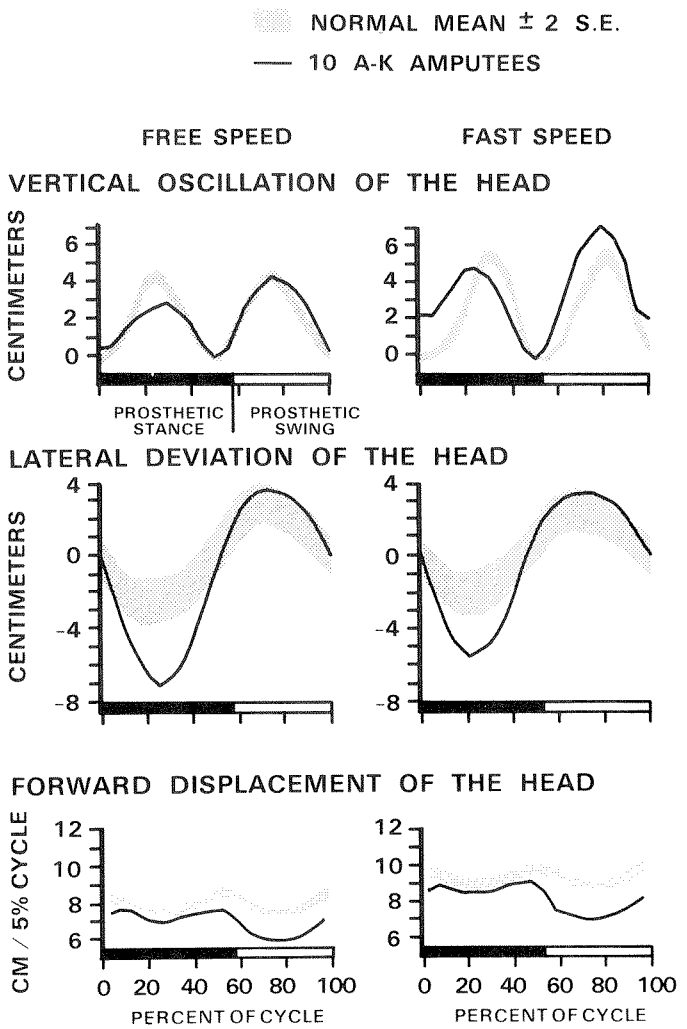

FIGURE 7.
Mean vertical, lateral, and forward pathways of the head of 10 above-knee amputees with constant-friction knee components. The shaded area depicts 2 standard errors above and below the mean for 30 normal men. The zero references on the ordinate scales of the upper two graphs represent the vertical and lateral positions of the head in the double-limb-support phase near the middle of the cycle.

tion of the hip abductor muscles (18). For the amputees, even with concentration of weightbearing on the medial aspect of the posterior brim of the socket, and with a properly-sloped lateral wall of the prosthesis, the hip abductor muscles of the residual limb are at a mechanical disadvantage as compared to the hip abductor muscles of a normal limb. In normal men this abductor muscle force is applied with the foot firmly fixed to the floor by friction, but when the amputees contract their abductor muscles during prosthetic single-limb support, the femur does not provide a fixed lever arm since it must first compress the soft tissues of the lateral thigh before purchase is obtained against the lateral wall. Secondly, the hip abductor muscles normally undergo a lengthening contraction during single-limb-support to control the descent of the pelvis on the opposite side where the limb is in the swing phase (19). If the amputees allowed normal descent of the pelvis on the sound side during the prosthetic single-limb-support phase, they would experience discomfort from compression of the soft tissue below the conjoint ramus of the pubis and ischium.

The excessive lateral trunk motion of the amputees may also relate to their abnormally wide strides, which may be caused by the presence of the medial wall of the socket. Regardless of the cause, these increased stride widths no doubt increase lateral stability. Evidently the gain in stability and comfort is worth the increased energy expenditure associated with the excessive lateral motion of the trunk during prosthetic single-limb support.

Fischer described the vertical pathways of the center of gravity as undulating through two peaks and two valleys of similar amplitude during each walking cycle, and the total excursion ranged from 3.4 to 4.4 cm, depending on speed. The valleys occur during double-limb support and the peaks occur during single-limb support. The configuration of the vertical pathways of normal men as measured at the head are identical to those calculated by Fischer, but the total excursions average 5.1 ± 1.0 cm for free-speed walking and 6.5 ± 1.5 cm for fast walking, and the successive vertical excursions within each walking cycle are of remarkably similar amplitude (12).

The amputees differed from normal in that they did not have equal successive peaks or equal successive valleys in their vertical excursions. The valley, during double-limb support when the prosthetic limb was forward, tended to be higher than

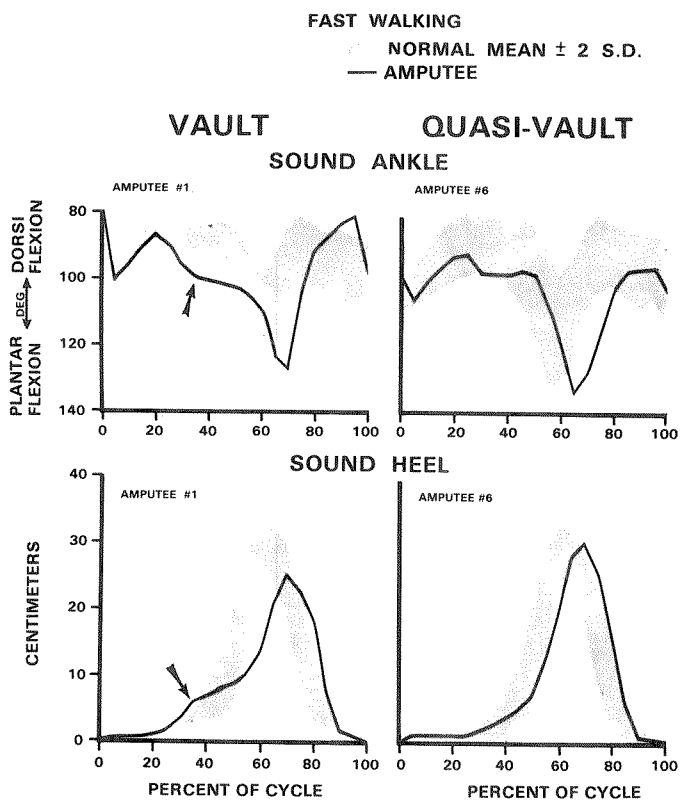

FIGURE 8.
Patterns of ankle motion and of the vertical pathway of the heel of the sound limb for above-knee amputees #1 and #6 (solid lines) and the mean pathways ±2 standard deviations for 30 normal men (shaded area) during fast walking. The patterns of amputee #1 (on the left) show marked premature plantar flexion and heel rise (arrows) which would be easily recognized as vaulting by a trained observer watching for abnormal vertical oscillation of the body and premature heel rise. The premature plantar flexion and heel rise of #6 was mild and could be called quasi-vaulting.

the valley when the sound limb was forward—particularly during fast walking. This may have related to the fact that the prosthetic knee must be fully extended at heel-strike, while normal knees during fast walking are flexed at heel-strike. The relatively lower peak during prosthetic single-limb support may relate to the fact that the amputees tended to incline their trunks laterally and forward at that time, or to the fact that the prosthetic limb was often slightly shorter than the sound limb. The relatively higher peak during the sound stance phase (prosthetic swing phase), particularly for

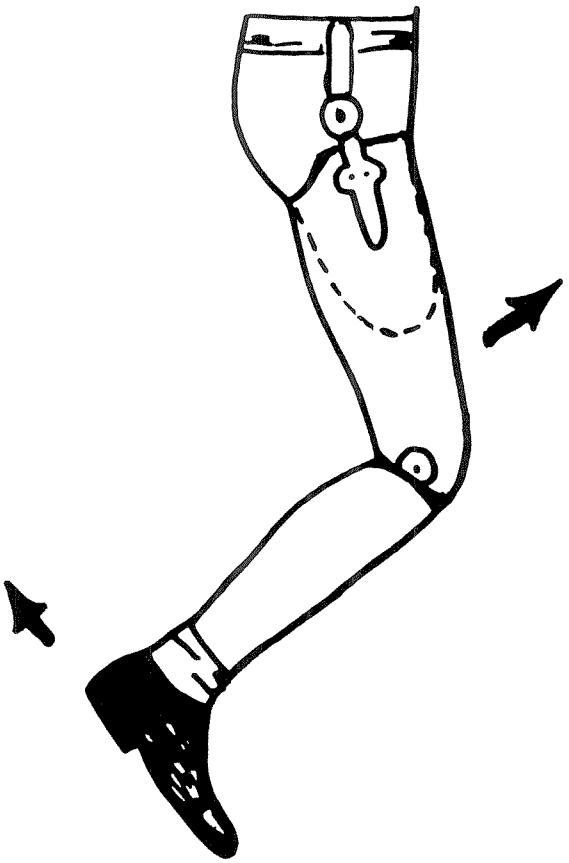

FIGURE 9.
The diagram seeks to depict in simplified form the mechanism of knee flexion during prosthetic swing. Knee flexion is produced here by the application of force moving the thigh segment in the counterclockwise direction as a result of flexion of the hip of the prosthetic limb. The resulting force applied to the knee is eccentric to the mass center of the shank-foot segment and initially produces clockwise rotation of that segment about the knee. (Among the factors which affect the extent and speed of such shank-foot rotation are the amplitude and speed of the applied force; another is the amount of freedom-to-rotate allowed to the knee joint.)

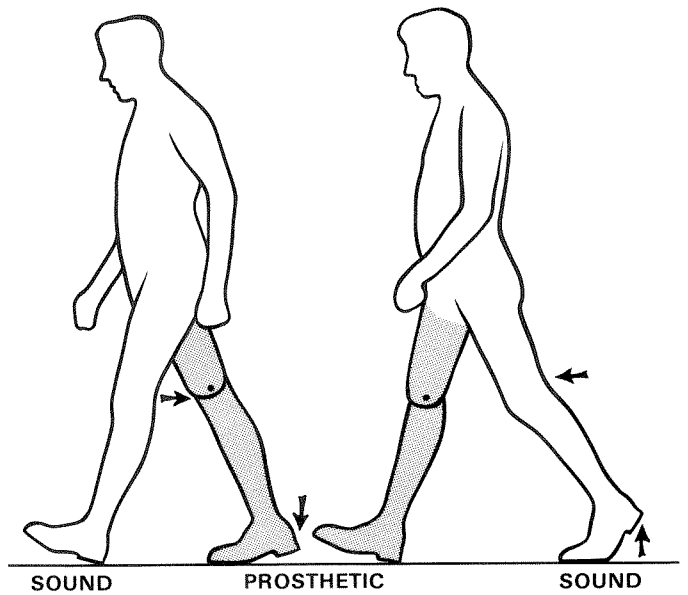

FIGURE 10.
Diagrams of an amputee at sound and prosthetic heel strike illustrating that the relatively longer step is taken when the prosthetic limb is forward. The length of the step taken by the sound limb (left diagram) is limited by the inability of the amputee to actively plantar flex the prosthetic ankle and safely flex the prosthetic knee.

fast walking, related mainly to a vaulting or quasi-vaulting pattern.

A few of the amputees had markedly excessive plantar flexion of the sound ankle during the early sound stance phase, resulting in vaulting consisting of visually obvious premature heel rise on the sound side, and visually obvious uneven successive vertical oscillations of the body (Figure 8, amputee #1). Precise measurements of the sound ankle patterns of the individuals, however, showed that most of the amputees had a quasi-vaulting pattern which would not have been evident to even a trained observer. That measured pattern consisted of premature but less-marked ankle plantar flexion on the sound side during the sound stance phase, accompanied by mild elevation of the heel (often less than one centimeter) (Figure 8, amputee #6). These measured vaulting or quasi-vaulting patterns did not relate to the amount of prosthetic toe-floor clearance during swing, nor did they relate to the type of suspension, as one might expect. It could be that this vaulting is related to the amputee's perception of where the prosthetic foot is in space. Indeed, the patterns of excessive prosthetic toe-floor clearance during swing (Fig. 5) suggest that the amputee's perception of where the foot is in space is not adequate to permit safe walking if toe-floor clearance is reduced to normal levels—the amputee needs a larger margin of safety.

In normal men, velocity of forward progression is remarkably uniform, as we have demonstrated with calculations of a velocity index. The amputees in this study had a mild stop-start type of forward progression, with relatively more-rapid forward movement of the head and trunk during prosthetic single-limb support than during sound

single-limb support. The rate of forward movement tends to speed up during early prosthetic stance because the amputees incline the trunk forward slightly to ensure that their line of gravity is anterior to the prosthetic knee axis, thus avoiding the flexion moment. This forward trunk movement occurs despite the fact that the residual hip was extending forcefully throughout the first half of the walking cycle, as seen in Figure 3. Conversely, the rate of forward movement tends to slow down during sound single-limb support, to allow time for prosthetic swing to occur. The mechanism which produces knee flexion during prosthetic swing is illustrated in Figure 9. Abrupt reversal of the residual hip from extension to flexion produces forward rotation of the thigh segment with rotation of the shank-foot segment in the opposite direction, similar to the action of a double pendulum. In absence of hydraulic or pneumatic swing control or other damping forces, excessive prosthetic knee flexion and excessive heel rise occur, thus prolonging the prosthetic swing phase and temporarily slowing down forward progression during walking. This set of conditions was observed only during fast walking.

The swing phase during free-speed walking was also prolonged, but this was not related to excessive prosthetic knee flexion or heel rise. Indeed, the absence of excessive knee flexion and heel rise during free-speed walking (Figures 4 and 5) seems to indicate that the friction settings of the knee components were optimal for the amputees' free-speed walking. The prolonged prosthetic swing phase during free-speed walking may have related to an increased sense of security and comfort during sound (as compared to prosthetic) single-limb support.

Another factor contributing to irregular forward progression—and also to the asymmetry in the sound and prosthetic swing phases—was the amputees' tendency to take a longer step with the prosthetic than with the sound limb. The mechanism for the typically uneven step lengths of amputees is illustrated in Figure 10. Normally, at the instant of heel strike the limb in the rear position contributes to the length of a step by a combination of hip extension, slight knee flexion, and slight ankle plantar flexion. For the amputees, however, the relatively shortened sound step length resulted from the inability to actively plantarflex the prosthetic ankle to raise the heel, and the inability to flex the prosthetic knee until weight is transferred to the forward limb. In order to achieve more equal successive step lengths, the amputee should be instructed to extend his residual hip as much as possible at the end of stance, and if feasible (depending upon the type of prosthesis), to increase transverse rotation of the pelvis. Beyond that, without active plantar flexion force in the prosthesis, the only practical way to ensure step length equality would be to take a shorter step with the prosthesis—but this would of course compromise the amputee's already slow velocity.

The amputees in this study were functionally independent and active persons who participated in outdoor recreational activities and had overcome the problems associated with uneven terrain, inclines, and stairs. They walked remarkably well compared to other groups of physically disabled persons whom we have studied, such as patients with hip pain, knee pain, or Parkinsonism. The gait abnormalities measured in this group of amputees are ones which must increase the mechanical energy cost of walking, but which appear to provide safety and to compensate for the problems related to walking with prostheses having constant-friction knee components.

CONCLUSIONS

1. The walking speeds of the amputees were subnormal as a result of both longer than normal cycle durations (slower cadences) and subnormal step lengths.
2. The stride widths of the amputees were wider than normal and their successive step lengths tended to be uneven, with the prosthetic-step length usually longer than the sound-step length.
3. The successive stance and swing phases were uneven, with longer stance on the sound side but longer swing on the prosthetic side.
4. The forward, vertical, and lateral trajectories of the trunk lacked the characteristic symmetry and smoothness of normal gait.
5. Both the sound and prosthetic side arm-swing patterns were abnormal, with less than normal flexion at the shoulder and subnormal extension at the elbow.
6. The amputees had an abrupt reversal of the residual hip from flexion to extension just after sound heel-strike, at both walking speeds.
7. The prosthetic knee did not yield into flexion in the early stance phase. The maximum knee flexion during swing was similar for the sound and

prosthetic limbs during free-speed walking, but the prosthetic knee had excessive flexion during the swing phase for fast walking.

8. The prosthetic heel rise was excessive when prosthetic knee flexion was excessive.

9. The toe-floor clearance distance of the prosthetic foot was in excess of normal, particularly during fast walking.

10. During fast walking, the sound ankle of most amputees had a premature reversal into plantar flexion during mid-stance.

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