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Evaluation of a Prosthetic Shank with Variable Inertial Properties

**by Scott Tashman, M. Eng.
Ramona Hicks, R.P.T., M.A.
David J. Jendrzejczyk, C.P.**

Above-knee amputees walk slower than the normal population. This has been documented in adults[1,2,3] and children.[4] It has been suggested that the prolonged swing phase of the prosthesis forces a slower cadence and, therefore, a slower walking speed.[5] Since children rely on a fast cadence to obtain an adequate walking speed,[6] a prolonged swing phase can be a major obstacle to comfortable, efficient normal-speed walking.

To date, most efforts to reduce prosthetic swing phase time have been directed towards the prosthetic knee joint.[7,8] Various mechanisms have been designed to accelerate the extension of the prosthetic knee. Mechanical, hydraulic, and pneumatic systems have been developed in an effort to provide a more favorable gait.[9] Hydraulic knee units have been shown to provide a more normal cadence and walking speed for adults than simple constant friction knee units.[10]

Most of the prosthetic knee unit research has been directed towards the adult amputee population. Pediatric hydraulic knee units have been considered impractical because of size and weight limitations. Pediatric above-knee amputees are generally fitted with constant friction knee units because they are simple, light in weight, low in cost, easy to install and adjust, and require little maintenance.[11]

It has often been presumed that adjustments in the knee joint friction could be used to provide an optimum cadence for the amputee with a constant friction knee joint. A study was performed at the Newington Children's Hospital Kinesiology Laboratory to test this assumption.[12] When subjects were asked to walk at a comfortable speed, no significant changes were observed in cadence or actual prosthetic shank swing time as the knee joint friction was varied over a wide range. In all cases, the swing period of the prosthetic shank was close to the natural swing period of the shank measured off the patient. This indicates that the physical properties of the prosthetic shank play a significant role in determining the natural cadence of the above-knee amputee with a constant-friction knee joint. To force the shank to move at a frequency different from its natural frequency requires significant input of energy in the form of applied torque at the knee joint (from hip or pelvic muscle force). The test subjects, when asked to walk at a comfortable speed, did not supply the extra energy needed for a faster cadence; they instead aligned their cadence with the natural frequency of the shank.

PURPOSE

The above results led to the current project: the design and testing of a prosthetic shank with variable physical properties. The purpose of this study was to test the following hypotheses:

1. If the physical properties of the shank section of an above-knee prosthesis with a constant friction knee unit are changed in such a way as to alter the natural swing period, the swing period of the shank during gait will also be altered.

2. Reducing the swing period of the shank will increase natural cadence and walking speed.

METHODS

Design

The principal design goal for the shank was to reduce the natural swing period as much as possible. If the shank/foot is considered as a physical pendulum, it has a period T equal to: $\mathbf{T = 2\pi \sqrt{I/Mgd}}$, I is proportional to Md^2.

Where:

- T = natural swing period of shank as a pendulum
- I = rotational inertia of shank/foot above knee pivot
- g = acceleration due to gravity
- d = distance from knee pivot to center of mass
- M = mass of shank/foot

These equations indicate that changes in mass alone will not reduce the swing period of the shank; the center of mass must be shifted proximally (towards the knee joint) to significantly reduce the period.

With reducing distal weight as the primary goal, an experimental shank was constructed for the test subject, a 13 year old male knee disarticulation patient with a "good" amputee gait pattern. Since the limb was to be used for laboratory testing purposes only, some strength was sacrificed in order to obtain the maximum possible reduction in distal weight while still using readily available materials. The shank was thin and hollow, with layers of polyester resin and one layer of carbon filter cloth laminated over a plaster mold. Excess material was ground away wherever possible. In addition, the prosthesis was set in correct alignment using a heel build-up on an ultra-light SACH foot to eliminate shoes and further reduce distal weight. To enable changes in the natural swing period, a lead mass which attached to a metal rod could be placed proximally or distally inside the shank. The additional mass was chosen so that the experimental shank/foot would weigh the same as the patient's standard prosthesis.

The completed prosthesis is shown in Figure 1. With the moveable mass placed distally, the shank had a center of mass positioned similarly to the patient's original shank. Shifting the mass proximally caused the center of mass to move proximally by 13 centimeters. To deter-

Figure 1. Completed experimental prosthesis; shown during testing in the Kinesiology laboratory.

Comparison of Physical Properties: Standard vs. Experimental Prosthetic Shank			
		Experimental Shank	
	Standard Shank	Proximal mass	Distal mass
Mass of Shank (Kg)	1.90	1.03	1.03
Mass of Shank & Weight	—	1.86	1.86
Distance from Knee Joint to Center of Mass (cm)	31.0	18.7	31.7
Swing Period (sec)	1.30	1.12	1.32

Table 1.

mine the effect of changing the mass position, the pendulum swing period of the shank was measured by timing the swing of the shank, which was suspended by a metal rod through the knee joint axis. The light weight shank, with the mass placed distally, exhibited inertial properties very close to those of the patient's original shank. Shifting the mass to the proximal position reduced the pendulum swing period by 0.20 seconds or 15 percent (Table 1).

Evaluation

The Newington Children's Hospital Kinesiology Laboratory measured the effect on the gait of the changes made in the position of the center of mass of the experimental prosthesis. An automated video system was used to acquire three-dimensional kinematic data from 26 retroreflective markers placed at designated positions on the body.[13,14] The kinematic data were used to determine the motions of all major body segments and calculate dynamic lower extremity joint angles in three planes. Linear movement and temporal measurements, such as stride length, single stance time, swing phase time, cadence, and walking speed were also determined. Swing time was determined by measuring the time from toe-off to heel strike. The shank pendulum time was determined by measuring the time required for the prosthesis to go from full extension into flexion and back to full extension; this is equivalent to one half of the period of the shank measured as a free-swinging pendulum.

Kinematic data were acquired for two walks with the subject walking at:

a. normal speed, weight proximal
b. fast speed, weight proximal
c. normal speed, weight distal
d. fast speed, weight distal

For the normal speed walks, the subject was asked to walk at a speed that was comfortable; no further prompting was given. For the faster speed walks, the subject was instructed to walk as fast as was comfortable; again, no further instructions were given. For each mass position, the knee joint friction was set to "clinically optimal" by matching the prosthetic side heel rise to the normal side heel rise at normal speed, and the patient was allowed to walk around for a while until he seemed reasonably comfortable with the altered characteristics of the limb.

RESULTS

Stride Parameters

Stride parameters measured during the four different conditions are shown in Table 2. This data represents the first walk acquired for each condition; the variation between the first and second trials for all conditions was less than five percent. Cadence, stride length, and walking speed were all essentially the same at the "normal" walking speed with the mass placed proximally or distally. At the "fast" walking speed, the subject walked seven percent faster with the mass placed distally than with the mass placed proximally, due to both a faster cadence and a longer stride length.

Shank Swing Dynamics

At the normal walking speed, the shank pendulum time was reduced by eight percent with the weight placed proximally, resulting in an eight percent reduction in the swing phase time for the prosthetic limb (Table 3). Since the swing phase time for the normal side stayed the

Results: Stride Parameters				
	NORMAL SPEED		FAST SPEED	
	Mass Proximal	Mass Distal	Mass Proximal	Mass Distal
Cadence (steps/min)	91.6	90.0	105	109
Walking Speed (cm/sec)	94.9	94.5	126	135
Gait Cycle Time (sec)	1.33	1.33	1.15	1.10
Stride Length (cm)	126.0	125.8	144.0	149.4

Table 2.

Swing Phase Timing: Normal Speed			
	Mass Distal	Mass Proximal	% Change
Swing Time—Prosthetic Side	0.62	0.57	−8%
Swing Time—Normal Side	0.51	0.52	+2%
Swing Time Asymmetry (%)	19.5	9.1	−10%
Shank Pendulum Time	0.60	0.55	−8%

Table 3.

Swing Phase Timing: Fast Speed			
	Mass Distal	Mass Proximal	% Change
Swing Time—Prosthetic Side	0.61	0.56	−8%
Swing Time—Norm. Side	0.44	0.46	+5%
Swing Time Asymmetry (%)	32.4	19.6	−13%
Shank Pendulum Time	0.61	0.51	−16%

Table 4.

same, the swing asymmetry (prosthetic side vs. normal side) was reduced from 19.5 percent to 9.1 percent. A similar reduction in swing asymmetry was seen during the fast walk (from 32.4 percent to 19.6 percent). During fast walking with the proximal weight placement, the swing phase time was increased by five percent for the normal limb and reduced by eight percent for the prosthetic limb. The reduction in pendulum swing time was much greater (16 percent).

The dynamic knee joint motion is shown for both weight positions (Figures 2 and 3). The peak knee flexion was reduced from 64 degrees to 54 degrees at the normal walking speed and from 84 degrees to 62 degrees at the fast walking speed with the weight placed proximally. The plots also indicate delayed initiation of knee flexion and faster motion of the limb with the proximal weight placement.

DISCUSSION

As expected, proximal weight placement in the shank produced a shorter shank swing time during gait. This subsequently resulted in a shorter swing phase (toe-off to heel-strike) for the prosthetic limb. At normal speed, the decrease in swing phase was equal in time to the decrease in shank swing time (eight percent). At a faster walking speed, the same eight percent decrease in swing phase was observed, but the shank swing period was reduced by a much

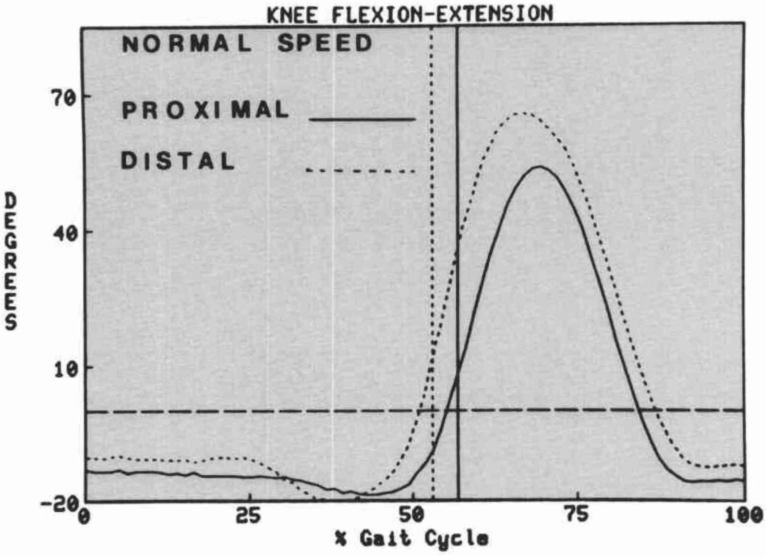

Figure 2. Knee flexion-extension angle vs. percent of gait cycle: normal walking speed, proximal and distal weight placement.

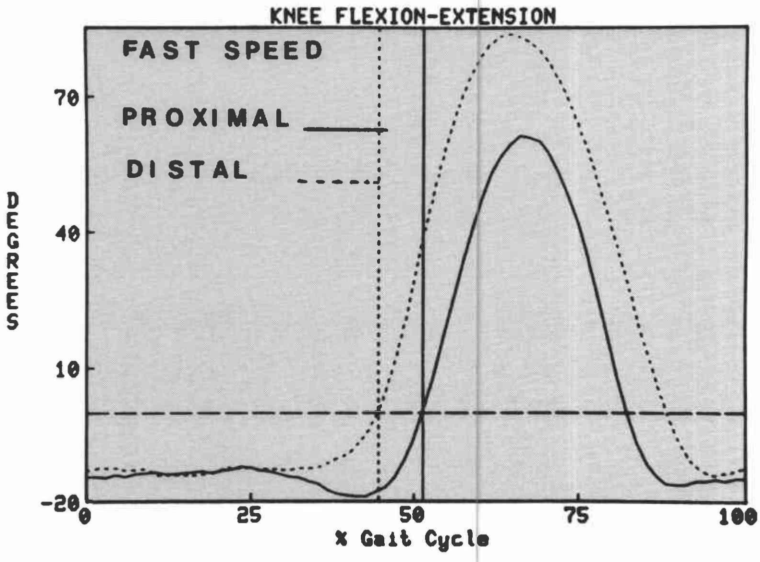

Figure 3. Knee flexion-extension angle vs. percent of gait cycle: fast walking speed, proximal and distal weight placement.

greater amount. Figure 3 illustrates the effect of this discrepancy: the limb reaches full extension well before heel strike. One explanation for this is that the subject did not have sufficient time to fully adjust to the new limb; further use should enable the subject to reduce swing phase as much as the shank swing period was reduced.

A less expected outcome was the similarity in walking speed and cadence between the two different weight placements. The reduced swing phase did not result in a reduced gait cycle time; the subject instead lengthened his stance phase to balance the decrease in swing phase. This resulted in a smoother, more symmetric gait.

CONCLUSIONS

Limited conclusions can be made based on this single-subject study. However, it appears that decreasing the natural swing period of the shank by shifting the center of mass proximally results in a faster swing phase during gait. In one subject this led to an increase in stance phase for the prosthetic side towards normal values, and considerably reduced left-right asymmetry for this subject. Improved symmetry should lead to a more energy efficient, natural appearing gait. No increase in cadence or walking speed was observed. It is possible that longer wear of the limb might have permitted the subject to naturally increase his cadence;

this could not be evaluated with the present limb design.

The outcome of this study indicates that weight distribution in the prosthetic shank/foot has a significant impact on gait. This suggests that future prostheses should be designed to minimize distal shank/foot weight.

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Kinematic and Kinetic Comparison of the Conventional and ISNY Above-Knee Socket

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Prosthetic sockets must be comfortable, but they must also be functional. Despite advances in other prosthetic components, the materials used to construct the portion of the prosthesis that most directly contributes to amputee comfort, the socket, have remained essentially unchanged since the introduction of thermosetting resin sockets in the 1950's.[1–4]

A prosthetic socket must perform at least two functions: it must contain the stump tissues, and during stance phase, it must provide a means of transferring the amputee's weight from the pelvis and residual limb to the floor. To contain the stump tissues, the socket shape encourages optimal distribution of forces and pressure, during both swing and stance phases. Body weight is transmitted primarily from the ischial tuberosity to the proximal brim of the socket, through the vertical socket walls, and finally through the knee, shank, and foot to the floor.[5]

Conventional sockets surround the entire residual limb with rigid thermosetting resins, thus requiring this single container to perform both socket functions. The rigid proximolateral socket wall has been reputed to provide stabilization for the residual limb,[3] although no em-