

Analysis of the swing phase dynamics and muscular effort of the above-knee amputee for varying prosthetic shank loads

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Abstract

The purpose of this study was to determine the effect of varying prosthetic shank mass, while maintaining the mass centre location and moment of inertia, on the swing phase kinematics, kinetics and hip muscular effort of free speed above-knee (AK) amputee gait. Six AK amputees, wearing similar prosthetic designs, had three load conditions applied to their prosthetic shank: 1) Load 0-unloaded ($X = 39.1\%$ sound shank mass), 2) Load 1- 75%, and 3) Load 2 -100% sound leg mass. Despite increases in shank mass from 1.33 to 3.37 kg the AK amputee was able to maintain a consistent swing time and walking speed. As load increased, there were significant changes in the maximum knee and hip displacements, as well as phasic shifting. The prosthetic knee Resultant Joint Moment (RJM) was negligible while the shank was accelerating (periods 1 and 2), but was a major contributor during shank deceleration (periods 3 and 4). During periods 1 and 2 the principle contributors to the shank acceleration (forces resisting excessive knee flexion) were the gravitational moment (S-G) and the moment due to thigh angular acceleration (S-AT). During the periods of shank acceleration (sections 1 and 2), there was not a significant increase in the hip muscular effort. However, during sections 3 and 4, the periods associated with shank deceleration, there were significant increases in the hip muscular effort. The hip muscular effort for the complete swing phase increased as load increased by 36.7% and 71.3% for loads 1 and 2. Despite the significant increases in hip muscular effort, four of the six subjects preferred load 1 condition.

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Introduction

Due to the loss of normal knee and ankle function in the AK amputee, changes in the joint angular patterns and compensations at the muscular level would be expected. The knee angular displacement pattern for AK amputees (Hicks *et al.*, 1985; Murray *et al.*, 1980 and 1983) is similar to that of the normal subject (Murray, 1967; Winter, 1984). However, the AK amputee hip angular displacement pattern differs by exhibiting a final phase of hip flexion after the normal flexion and extension periods (Murray *et al.*, 1980 and 1983) or an extra flexion and extension period (Frigo and Tesio, 1986; Hale and Putnam, 1987). The typical AK amputee knee moment (torque) pattern is similar to the normal subject pattern of extension followed by flexion, except the AK amputee exhibits lower peak magnitudes (Hale and Putnam, 1987; Hoy *et al.*, 1983; Judge, 1978). On the other hand, the AK hip muscular moment pattern exhibits intermediary extensor and flexor moments during the midswing stage (Hale and Putnam, 1987), between the normal pattern (Winter, 1984) of an initial flexor and final extensor moment.

Several investigators have suggested that changes in the segment inertial parameters — mass, mass centre location and moment of inertia of the prosthesis also influence the swing pattern of the AK amputee (Menkveld *et al.*, 1981; Tashman *et al.*, 1985). No significant changes were recorded in the stride length, stance time, knee angle or the rate of heel rise for six subjects when two loads were applied to the prosthetic foot (Godfrey *et al.*, 1977). Tashman *et al.*

Work was carried out at the School of Physical Education, Dalhousie University, Halifax, Nova Scotia, and the Department of Kinanthropology at the University of Ottawa, Ottawa, Ontario, Canada.

(1985) reduced the moment of inertia of a single AK prosthesis which resulted in lowering the natural swing period of the prosthesis. This, in turn, reduced the swing time and maximum knee flexion, and improved gait symmetry. Increasing the prosthetic shank mass and changing the mass centre location and moment of inertia, to near normal values, of a single AK subject resulted in increases in the stride length and walking speed, and was accompanied by a decreased hip muscular moment (Menkveld *et al.*, 1981).

The current trend in prosthetic designs is toward lightweight prostheses, as seen by the use of lightweight materials such as titanium and carbon fibre. It was expected that with the lighter prosthesis, less muscular effort would be required to advance the swing limb forward (Murphy, 1964). It was speculated that if the prosthetic limb had a similar mass and moment of inertia to that of the anatomical limb, the amputee would require three times the normal hip muscle exertion (Eberhart *et al.*, 1968). The problems encountered in the AK amputee swing phase in limiting their walking speed may be solved with proper prosthetic design, but the solution to these problems and the implication on muscular effort are not well understood.

The purpose of this study was to determine the effect of loading the prosthetic shank, (while maintaining the mass centre location and moment of inertia), on the joint and segment angular kinematics, the forces acting on the prosthetic shank, and the hip muscular effort during the swing phase of free speed AK amputee gait.

Methodology

The following criteria were used for subject selection: 1) male, 2) between the ages of 18 and 65 years, 3) unilateral above-knee (transfemoral) amputation, 4) prosthesis user for a minimum of one year, 5) body weight (without prosthesis) between 50 and 70 kg, 6) previously used the experimental prosthesis, and 7) clinically classified as a "good to excellent" walker. Six subjects were asked to participate, informed of the experimental procedure and written consents obtained. Subject descriptive information is provided in Table 1.

All subjects wore the following AK prosthetic design: 1) Otto Block (OB) endoskeletal system, 2) either a rigid quadrilateral ischial weight-bearing, total contact socket (New York University, 1982) or an Icelandic, Swedish, New

Table 1
Subject descriptive data

Subject	Age (yrs)	Weight (kg)	Height (m)	Years Amputated (yrs)	Cause
A	33	71.1	1.72	5.5	trauma
B	22	52.3	1.75	9.0	cancer
C	61	67.7	1.66	40.0	cancer
D	26	58.5	1.71	9.5	trauma
E	35	72.3	1.76	3.5	trauma
F	40	60.3	1.85	37.0	cancer
mean	36.2	63.7	1.74	17.4	

York (ISNY) flexible socket (Kristinsson, 1983), 3) suspension by either suction, or silesian belt or a hip joint and pelvic band, 4) an OB 3R15 knee unit with swing and stance controls, 5) an extension assist unit (OB 21B30), 6) an OB 4R2 pylon, and 7) a single axis OB 1H32 ankle/foot unit.

Three experimental load conditions were selected: 1) load 0, the unweighted prosthetic shank plus unshod foot ($X = 39.1\%$ the anatomic leg mass); 2) load 1, the prosthetic shank mass was increased to 75% subject's sound shank mass; 3) load 2, the prosthetic shank and foot was increased to 100%. The sound shank plus foot mass was estimated using data presented by Dempster (1959) and corrected for the subject's height and weight. The load was applied about the mass centre location (CM) to minimize the changes in the CM and moment of inertia. The mass and moment of inertia for each load condition are presented in Table 2.

The lower limb was modelled as a rigid two link system (Fig. 1). The segment inertial parameters of the prosthetic thigh and shank were determined using a weigh scale (mass), knife-edge balance beam (mass centre location) and a specially designed pendulum (moment of inertia). The mass of the residual thigh was estimated using Dempster's cadaver data, corrected for the subject's height and weight, and taken as a percentage of the remaining thigh length. The residual thigh mass centre location and moment of inertia were mathematically determined using known equations for a right frustrum of a cone with a uniform density (Beer and Johnston, 1976). The inertial characteristics of the residual thigh and prosthetic thigh section were combined to give the total thigh characteristics.

The subjects were asked to walk at a self

Table 2
Mass (M) and moment of inertia (I) of the prosthetic shank*
for the different load conditions

Variable	Load	Subject						mean
		A	B	C	D	E	F	
Mass (kg)	0	1.75	1.62	1.64	1.54	2.01	1.92	1.75
	1	3.93	2.29	2.82	2.47	4.18	3.47	3.15
	2	5.12	2.95	3.67	3.19	5.38	4.47	4.13
I (kgm ²)	0	.034	.032	.033	.027	.042	.049	.036
	1	.040	.034	.037	.028	.049	.055	.041
	2	.040	.036	.037	.029	.054	.056	.042

*includes the foot, shoe and added loads

selected, comfortable free walking speed over a 10m long walkway. Prior to data collection, the subjects were allowed to practise walking with each load. The loads were given in a random and balanced order. A 16mm camera filmed at a rate

of 75 frames per second. A timing light operating within the camera at 100 Hz confirmed the filming rate. Markers were placed on: 1) the greater trochanter of the femur, 2) the lateral screw of the prosthetic knee joint, 3) the most postero-lateral aspect of the heel, and 4) the most antero-lateral aspect of the shoe.

The segment end points in each frame of the swing phase were digitized (Hewlett-Packard HP9874A) and transferred to a mainframe computer (Cyber 170/580). The displacement data were smoothed using zero lag, low pass, second-order Butterworth digital filter. The smoothed data were then twice differentiated using a finite difference routine. The linear and angular kinematic data were generated for each segment.

To determine the forces and moments acting on the segments, the inverse dynamics approach was used. By knowing the outcome of the forces and moments (segment and joint kinematics) and the segment inertial parameters, the forces and moments acting on the segments can be determined. Standard Newtonian equations of motion were written in vector notation for each segment as follows:

$$\text{Shank: } \vec{R}JF_k + \vec{W}_l = m_l * \vec{a}_l \quad (1)$$

$$\vec{R}JM_k + (\vec{r}^{k/CMl} \times \vec{R}JF_k) = I^{CMl} * \vec{\theta}_l \quad (2)$$

$$\text{Thigh: } \vec{R}JF_h - \vec{R}JF_k + \vec{W}_t = m_t * \vec{a}_t \quad (3)$$

$$\vec{R}JM_h - \vec{R}JM_k + (\vec{r}^{h/CMt} \times \vec{R}JF_h) - (\vec{r}^{k/CMt} \times \vec{R}JF_k) = I^{CMt} * \vec{\theta}_t \quad (4)$$

All terms are defined in the appendix.

The thigh interacts with the shank via the RJM_k and the RJF_k thereby influencing the angular

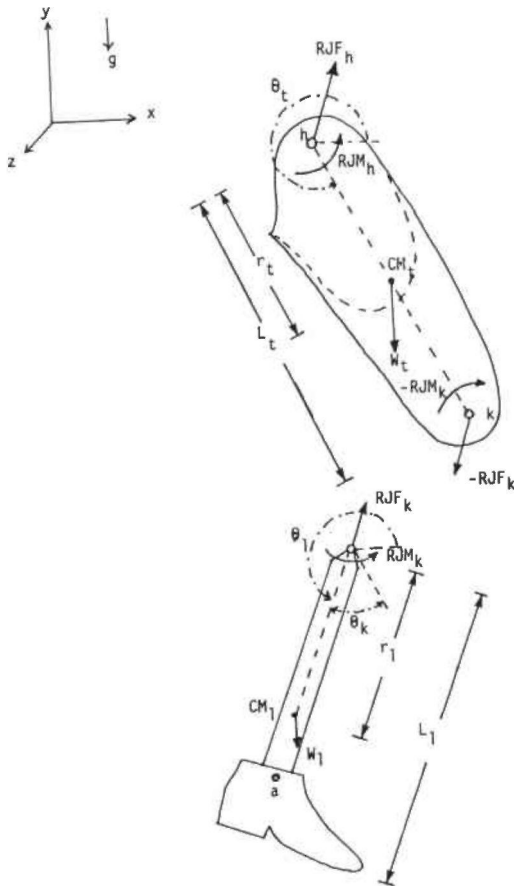


Fig. 1. Model of the prosthetic limb during the swing phase of walking.

acceleration of the shank (equation 2). The RJF_k is a function of the shank weight and linear acceleration of the shank mass centre (equation 1). The latter can be expressed as a function of the linear acceleration of the hip and the angular velocities and accelerations of the thigh and shank (Putnam, 1980). Expressing the acceleration of the shank mass centre in this manner and substituting into equation 2 yields the following equation:

$$\begin{aligned} \text{Shank: } RJM_k &= K1\cos\theta_k*\ddot{\theta}_i - K1\sin\theta_k*\dot{\theta}_i^2 + \\ &K5(\sin\theta_i)l_x - K5(\cos\theta_i)l_y - \\ &K5(\cos\theta_i)g = K3(\ddot{\theta}_i) \end{aligned} \quad (6)$$

OR

$$\begin{aligned} RJM_k &= (S-AT) - (S-VT) + (S-AH) \\ &- (L-G) = (S-NET) \end{aligned}$$

All constants are defined in the appendix. This equation describes how the thigh interacts with the shank, thereby influencing the angular acceleration of the shank in terms of the RJM_k and several shank interactive moments.

Following a similar process described above another equation can be determined which describes how the pelvis and shank interact with the thigh, thereby influencing the thigh angular acceleration in terms of the RJM_h and several thigh interactive moments.

The swing phase was divided into 4 sections based upon the direction of the hip RJM. The hip

Table 3

Group average for the stride kinematics under each load condition and the ANOVA results

Average of 18 trials for each load condition

Standard deviations in brackets

Load	*Walking Speed (m/s)	**Stride Speed (m/s)	Stride Length (m)	Stride Time (s)	Swing Time (s)
0	0.98 (.23)	1.01 (.24)	1.40 (.21)	1.41 (.15)	0.58 (.06)
1	0.99 (.22)	0.99 (.22)	1.40 (.20)	1.42 (.13)	0.58 (.05)
2	0.98 (.23)	1.00 (.23)	1.41 (.20)	1.43 (.15)	0.58 (.06)
F	0.03	0.63	0.31	1.60	0.62
p	0.744	0.554	0.740	0.250	0.557

* Walking speed - calculated over 3 to 6 strides

** Stride speed - determined by dividing the stride length by the stride time for the stride analyzed in detail

muscular effort was determined for each section of the swing phase by calculating the integral (area under the curve) of the RJM curve for each section. The overall muscular effort for the complete swing phase was also determined using the absolute hip RJM integral.

The subjects were asked to complete a questionnaire on their perception of the effort required to walk with the different load conditions and which load condition they preferred.

The statistics performed involved a two-way analysis of variance for repeated measures, load condition and trial, for selected kinematic and kinetic data. Pairwise comparisons, using Tukey's W procedure, were performed to determine where the difference occurred.

Results

The results presented are an average of the 6 subjects (18 trials) and represent the general trends. The level of significance was established at 0.05 for all comparisons.

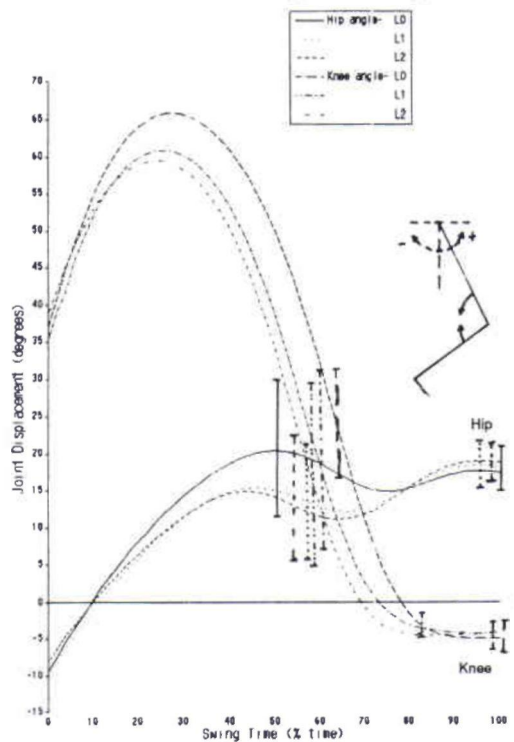


Fig. 2. The averaged knee and hip angular displacement curves for the three load conditions during the swing phase. The vertical lines represent the smallest and largest standard deviations for each load condition.

Stride Kinematics

Table 3 summarized the stride kinematics across each load condition. Increasing the prosthetic shank mass up to 100% sound shank mass had no significant effect on the stride kinematic variables.

Angular Kinematics

The knee angle was defined as the included knee angle as indicated in Figure 2. The knee angle used in equation 6 was equal to:

$$180^\circ - \text{included knee angle.}$$

The thigh angle (Fig. 2), which approximately equates to the hip angle, was expressed as:

$$\text{thigh angle} - 270^\circ.$$

The shank and thigh acceleration curves for the three load conditions during the swing phase are illustrated in Figure 3.

Angular Kinetics

Moment contributions to shank acceleration

The mean impulses of the moments acting on

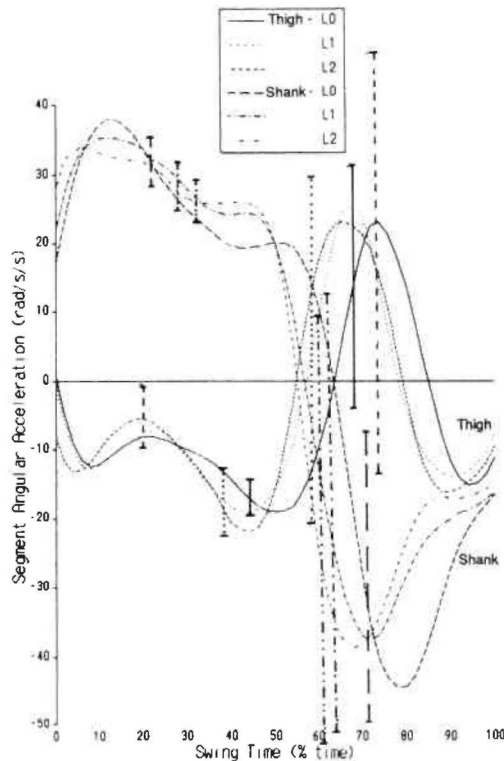


Fig. 3. The averaged shank and thigh angular acceleration curves for the three load conditions during the swing phase. The vertical lines represent the smallest and largest standard deviations for each load condition.

the shank were compared across loads and the four sections in Figure 4.

Knee and hip RJM patterns

The knee RJM pattern (Fig. 5) exhibited initial low extensor (load condition 0) or flexor moment (loads 1 and 2) followed by an increasing flexor moment. The averaged hip RJM pattern (Fig. 5) for the 3 loads, exhibited 4 sections: 1) an initial flexor, followed by 2) an extensor, then 3) a flexor and ending with 4) a final extensor moment.

Muscular Effort

The group mean hip muscular effort for each section and the complete swing phase are shown in Figure 6. In sections 1 and 2 there were no significant increases in the hip muscular effort. In sections 3 and 4 as load increased there were significant increases in the hip RJM impulse. The absolute hip RJM impulse for the complete swing phase significantly increased from 3.24 ± 1.47 Nms to 4.43 ± 1.81 and 5.55 ± 1.82 Nms for loads 1 and 2, respectively.

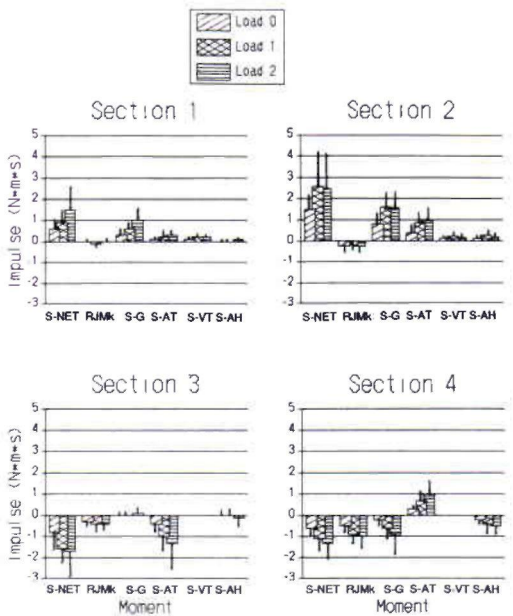


Fig. 4. The averaged impulses of the moments acting on the prosthetic shank during the four sections of the swing phase. The vertical lines represent the standard deviation for each variable.

Subjective Evaluation

Table 4 presents the subjects' preference of the load condition, taken from the questionnaire. Four of the six subjects preferred load condition 1 (75% anatomic shank mass). The remaining two subjects preferred load 0 condition (unweighted prosthetic shank).

The primary reasons given for choosing load 1 were: 1) more control, 2) less strenuous, 3) easier to swing the prosthesis, 4) smoother swing, and 5) improved consistency in walking. The reasons given for selecting load 0 by the two subjects were: 1) less weight to lift, 2) less effort required to walk, and 3) improved socket comfort.

Discussion

Despite the attempt to maintain the moment of inertia, it increased up to 28.5% (subject F), with a mean increase of 16.7%. The mass centre was

Table 4

Subject preference of the different prosthetic shank masses

(The order of preference is from the most desirable (i) to the least desirable (iii).)

Subject	Load		
	0	1	2
A	iii	i	ii
B	ii	i	iii
C	i	ii	iii
D	iii	i	ii
E	iii	i	ii
F	i	ii	iii

assumed to be unchanged since the load was applied about it. Changes in the moment of inertia could be attributed to the size of the load applied about the mass centre, the irregularity of its shape and positioning of the load.

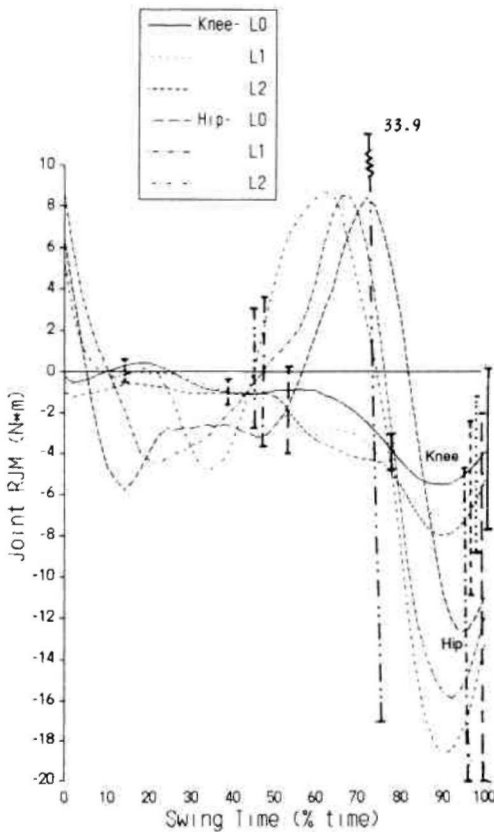


Fig. 5. The averaged knee and hip resultant joint moments (RJM) for the three load conditions during the swing phase. The vertical lines represent the smallest and largest standard deviations for each load condition.

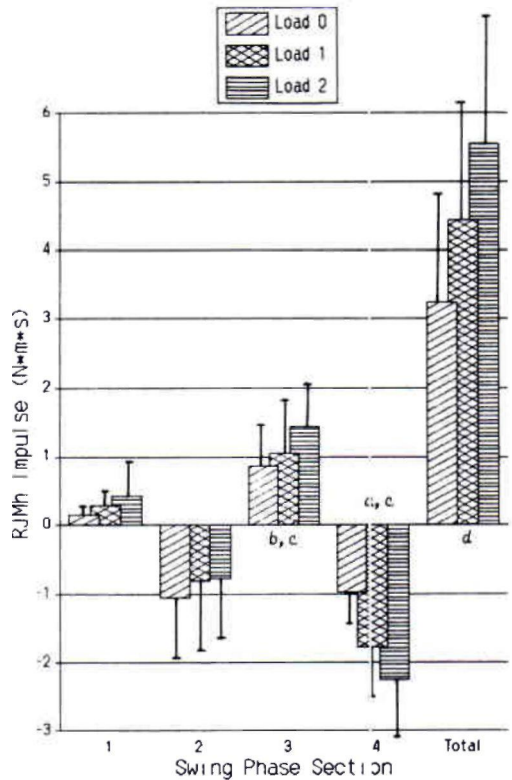


Fig. 6. The averaged hip muscular impulse (effort) for the four sections of the swing phase and for the complete swing phase for the three load conditions. The vertical lines represent the standard deviation for each load condition in each section. a) Significant difference between loads 0 and 1. b) Significant difference between loads 1 and 2. c) Significant difference between loads 0 and 2. d) Significant difference between all loads.

The consistent walking and stride speeds across loads were similar to previous studies on the effect of altered segment inertial parameters on AK gait (Godfrey *et al.*, 1977; Tashman *et al.*, 1985). The findings suggest that AK amputees can attain a free and comfortable walking speed when the prosthetic shank mass is increased to 100% of the remaining shank. However, the gait may change if the patient would be required to walk for a longer period with the heavier loads.

As load increased the maximum knee flexion attained significantly decreased ($p < .0334$). The time at which this angle was reached occurred significantly earlier ($p < .0460$). The time when full extension was attained occurred significantly earlier as load increased ($p < .0006$). The knee angle patterns were similar to previous AK studies where the amputees used constant friction knee units (Frigo and Tesio, 1986; Murray *et al.*, 1980 and 1983).

The thigh (hip) displacement pattern differed from the normal pattern of flexion followed by extension (Murray, 1967; Winter, 1984), but was similar to previous AK studies (Frigo and Tesio, 1986; Hale and Putnam, 1987; Murray *et al.*, 1980 and 1983). The thigh curves exhibited a double flexion peak separated by a valley. The initial peak flexion angle significantly decreased ($p < .0215$) and occurred earlier ($p < .0007$) as prosthetic shank mass increased. The thigh then began to extend reaching a local minimum flexion angle which significantly decreased ($p < .0116$) and occurred earlier ($p < .0003$). The thigh reversed direction reaching a final peak flexion, which in turn significantly increased ($p < .0175$) as load increased. Finally, the thigh briefly extended to end the swing phase. It has been previously suggested that the nature of the knee and hip (thigh) patterns were probably related to the characteristics and function of the AK prosthesis (Frigo and Tesio, 1986) and that the prosthetic knee could be controlled by newly learned hip motion and residual thigh patterns (Murphy, 1964). The changes in magnitude and the phasic shifts in the peaks and valleys of the thigh displacement curves suggest that the AK amputee does make some adaptations at the thigh/hip to the changes in prosthetic shank mass.

The shank acceleration (Fig. 3) was fairly consistent for approximately 50% of the swing phase, remaining positive for this period. It then rapidly decreased to a peak negative acceleration. This peak was not significantly different between

loads, but the time to the peak occurred significantly earlier ($p < .0001$). The shank continued to be negatively accelerated at the end of the swing phase.

The thigh was initially negatively accelerated for approximately 50% of the swing phase (Fig. 3). The thigh rapidly increased its acceleration, reaching a peak positive acceleration. This peak significantly increased ($p < .0083$) and occurred earlier ($p < .0006$) as load increased. The thigh acceleration then decreased to a negative value and remained as such until ground contact.

The segments' angular acceleration curves for loads 1 and 2 were closer in magnitude than load 0 was to either loads 1 or 2. This may have been the result of the larger increase in the prosthetic shank load between loads 0 and 1 (an average of 1.40 kg) than between load 1 and 2 (an average of 0.98 kg). It thus appears that the degree to which the AK amputee adapts was a function of the mass applied to the prosthetic shank.

During section 1 (Fig. 7a) the knee joint continued to flex while the hip was being flexed, although it was still in an extended orientation. Load had little effect on the joint kinematics during this section. The gravitational moment (S-G), due to the mass of the prosthetic shank, was the principle impulse accelerating the shank forward and resisting continued knee flexion. As load increased the S-G impulse significantly increased ($p < .0370$). The interactive terms S-AT and S-VT, which represent the effect of the thigh angular acceleration and velocity on the shank, assisted S-G, but neither significantly increased as load increased. The knee RJM, which quantifies the role of the prosthetic knee assistive devices, was negligible during section 1 and contributed little to the shank motion.

In section 2, the knee reached maximum knee flexion, then began to extend. The hip continued to be flexed until a maximum hip flexion angle was reached. A similar kinetic pattern existed in section 2 (Fig. 7b) as that seen in section 1 where the gravitational moment was the principle contributor and was assisted by the S-AT and S-VT terms, except as load increased the term S-AT also significantly increased. The significant increase in the S-AT impulse was the result of the increase in shank mass and a change in the knee angle. It was *not* the result of a significant increase in the thigh angular acceleration. The decrease in the maximum knee flexion was the result of the larger contribution of the terms S-G, S-AT and

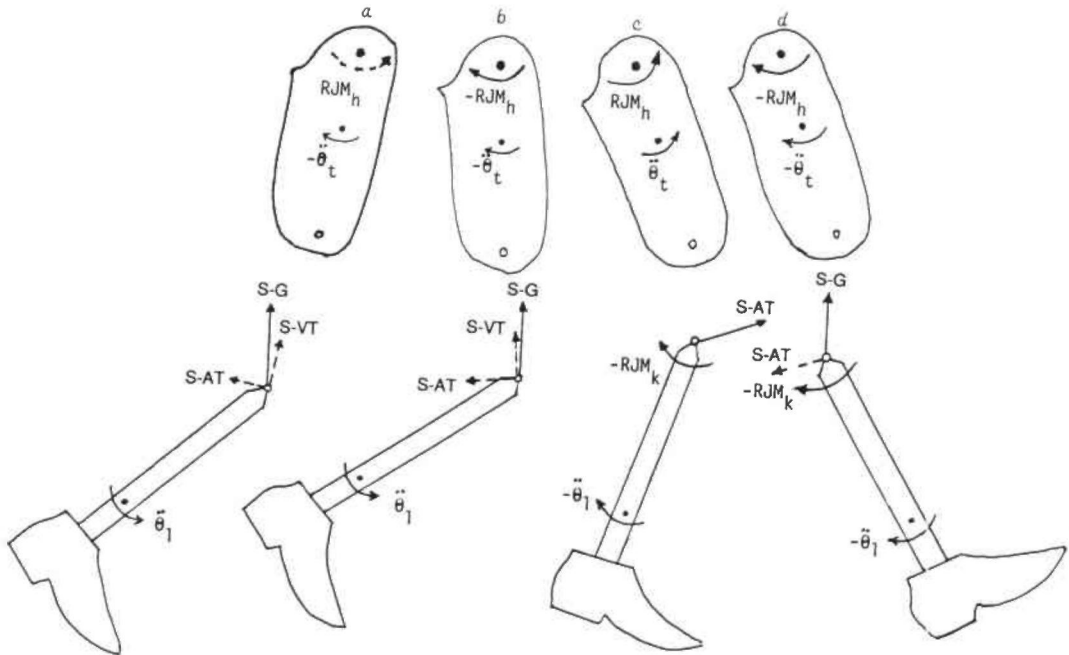


Fig. 7. The direction of the shank and thigh acceleration, the hip and knee RJM, and the principle moments contributing to the shank acceleration during the four sections of the swing phase for the three load conditions.

S-VT to the shank acceleration. The knee RJM continued to be negligible and contributed the least to the shank acceleration. Because of the reduced role of the knee RJM, the AK amputee was more dependent on the gravitational moment and the interactive moments, S-AT and S-VT. The AK amputee appeared to compensate for the reduced knee RJM by using thigh motion. During this period the thigh continued to be decelerated and the hip extensor moment contributed to this. Since the term S-AT described the effect of thigh acceleration on the shank motion, the contribution of the hip extensor moment to the thigh deceleration illustrated how the AK amputee attempted to control the motion of the knee.

In section 3 (Fig. 7c) the knee continued to extend and ended with the knee reaching full extension (hyperextension). The hip reversed direction and began to extend until a minimum hip flexion angle was attained. During this period a peak negative shank and positive thigh acceleration occurred. The shank deceleration was primarily the result of the moment S-AT. Its contribution significantly increased as load increased. Again, the thigh motion was important in influencing the prosthetic shank swing motion. Although the thigh acceleration did not

significantly increase as load increased there was a significant increase in the hip flexor RJM. The AK amputee flexed the hip to accelerate the thigh forward; the thigh interacted with the shank, through the term S-AT, to begin slowing down the swing of the shank in preparation for the next footstrike. The knee RJM assisted S-AT in slowing down the shank.

Section 4 was characterized by the knee being maintained in hyperextension and the hip flexing and briefly extending prior to the next footstrike. The knee must be maintained in full extension for a longer period because it reached full extension earlier for the heavier loads. During this section (Fig. 7d) the shank was decelerated primarily due to the knee RJM, which was assisted by the gravitational moment. Both terms significantly increased as load increased. These two terms not only slowed the shank down but also attempted to flex the knee. Flexion of the knee leads to knee instability at ground contact and may result in stumbling. The hip extensor RJM not only decelerated the thigh but aided in stabilizing the knee. The result of the thigh deceleration was that the term S-AT countered the knee RJM and S-G term and attempted to maintain the extended knee until ground contact.

Muscular Effort

The hip muscular effort (Fig. 6) significantly increased as load increased. This was the result of the significant increase in the hip RJM impulses for sections 3 and 4. The increase in the hip RJM impulse in section 3 was related to the poor response of the knee RJM to the increased mass. To attain similar shank velocity for section 3 across loads S-AT significantly increased. S-AT was dependent on shank mass and thigh acceleration, to which the hip RJM positively contributed. In section 4 the hip extensor RJM impulse significantly increased and performed a dual function. First, it slowed the thigh down and secondly, it maintained the knee in full extension, by contributing to the interactive moment S-AT, for a significantly longer period of time as load increased.

Loading the prosthesis to 75% and 100% anatomic shank mass resulted respectively in a 36.7% and 71.3% increase in the hip muscular effort. This differs from the suggested threefold increase if the prosthetic shank had the same mass, mass centre location and moment of inertia (Eberhart *et al.*, 1968). The fact that the mass centre location and moment of inertia of the prosthetic shank were minimally changed in this study may account for some of this difference.

A statistically significant increase in the hip muscular effort does not mean that it was practically significant and the subjective evaluation data bears this out. Four of the six subjects preferred load 1 over loads 0 and 2 (Table 4). This suggests that the hip RJM impulse may not be the only factor to consider when designing the prosthesis. Subject preference may have been influenced by the subject's familiarity with the weight of the prosthesis and the function of the knee unit. Some subjects may not prefer load 0 because it was lighter than the prosthesis they used for everyday activities. Subjects who have sophisticated knee units may have rejected heavier loads because the simple knee unit did not have the dampening qualities; hence, terminal impact was more pronounced with increased loads.

Many other factors exist when muscular effort is considered: 1) prosthetic alignment (Ishai *et al.*, 1983), 2) large and asymmetrical trunk rotations (Cappozzo *et al.*, 1982), 3) components of the AK prosthesis, particularly the knee (Hicks *et al.*, 1985; Ishai *et al.*, 1983; Tsai and Mansour, 1986; Zarrugh and Radcliffe, 1976), 4) moment of

inertia and the location of the mass centre of the prosthetic shank (Tashman *et al.*, 1985; Tsai and Mansour, 1986), 5) the amount of residual muscle mass and residual thigh length (Tsai and Mansour, 1986). With the introduction of the energy storing ankle/foot, there may be less dependency on the hip musculature to initiate swing, because of the potential capacity to accelerate the prosthetic shank upward and forward during terminal stance. Further research should be carried out to investigate the role of the energy storing feet in the initiation of swing and the effect on the role of the hip musculature during the swing phase.

Conclusions

- 1) Despite increases in the prosthetic shank mass up to 100% of the anatomic shank mass, by adding loads ranging between 1.33 to 3.37 kg, the AK amputee could still maintain a consistent free walking speed.
- 2) As prosthetic shank mass increased the maximum knee flexion attained decreased and full knee extension occurred prematurely.
- 3) The gravitational moment (S-G) and the shank moment due to thigh acceleration (S-AT) were the principle contributors to the shank acceleration and resisting knee flexion for all load conditions.
- 4) The hip muscular effort of the complete swing phase significantly increased as load increased. This increase was primarily due to significant increases in the hip flexor and extensor RJM impulses for sections 3 and 4, the period when the shank was decelerated in preparation for the next footstrike.
- 5) Despite significant increases in the hip muscular effort four of the six subjects preferred load 1, or 75% sound shank mass condition. This suggests that the hip muscular effort may not be appropriate as a sole determinant of prosthetic design. An increase in the hip muscular effort does not imply an inefficient gait.

Acknowledgements

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Appendix

- $K1 = r_1 * L_t * m_1$
 $K3 = I^{CM} + ((r_1)^2 * m_1)$
 $K5 = r_1 * m_1$
- × denotes the cross product
 * denotes multiplication
 s denotes shank
 t denotes thigh
 h denotes hip
 k denotes knee
 CM denotes mass centre of segment
 m_s denotes mass of the segment
 W_s denotes weight of the segment
 I^{CMs} denotes the moment of inertia of the segment about its mass centre
 a_s denotes linear acceleration of the segment
 θ_x angular displacement of segment or joint
 $\dot{\theta}_x$ denotes angular velocity of segment or joint
 $\ddot{\theta}_x$ denotes the angular acceleration of the segment or joint
 $r^{k/j}$ represents vector directed from point k to point j
 RJM_j denotes the resultant joint moment about the joint j
 RJF_j denotes the resultant joint force at joint j
 l_x, l_y denotes the linear acceleration of the hip joint in forwards or backwards and up or down directions.
- L represents the segment length
 r represents distance between mass centre of segment to proximal joint
 g represents the acceleration due to gravity (9.8 m/sec.)
- S-NET — the net effect of all moments acting on the shank and is equal to the shank angular acceleration multiplied by the moment of inertia of the shank measured about the transverse axis passing through the knee.
- RJM_k — resultant joint moment of forces exerted by the assistive devices about the prosthetic knee joint.
- S-AT — shank interactive moment which is a function of the thigh angular acceleration.
- S-VT — shank interactive moment which is a function of the thigh angular velocity.
- S-AH — shank interactive moment which is a function of the linear acceleration of the hip joint.

- S-G — shank interactive moment which is a function of the gravitational forces acting on the leg.
- RJM_h — resultant joint moment of muscle forces acting about the hip joint.

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