Short and Longer Term Changes in Amputee Walking Patterns Due to Increased Prosthesis Inertia

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ABSTRACT

The purpose of this study was to quantify the time course of changes in walking patterns among unilateral, transtibial amputees whose prosthesis inertia properties were substantially altered to match with those of their intact limb. Four unilateral transtibial amputees completed three test sessions on days 1, 2, and 8 of an 8-day protocol. Walking kinetics were computed from overground trials; temporal characteristics were collected during treadmill walking. Assessments were initially performed on day 1 without additional mass on the prosthesis. Mass was then added to the distal aspect of the prosthesis such that the mass and moment of inertia of the prosthetic leg were matched with those of the intact shank and foot. This added mass remained attached to the prosthetic limb for the next 7 days. Gait assessments were completed immediately and after 5, 10, 15, 30, and 60 minutes of exposure to the altered inertia. Amputees returned to the laboratory on days 2 and 8 for additional assessments. Measures of gait symmetry between the intact and prosthetic legs changed within 5 minutes of exposure to altered prosthesis inertia and remained in this altered state until the load was removed on the eighth day, at which time symmetry indices returned to their original state. Matching the inertia properties between legs exacerbated stance time and swing time asymmetries but improved peak knee moment symmetry during swing termination. The increased joint kinetic symmetry, however, required greater muscular efforts, particularly on the prosthetic side. In conclusion, substantial alteration of the inertia properties of the prosthesis immediately altered temporal and joint kinetic symmetry between intact and prosthetic legs, both when mass was added to the prosthesis and when that mass was removed. Because of our small sample size, caution should be exercised when generalizing these outcomes to all lower limb amputees. (*J Prosthet Orthot.* 2011;23:114–123.)

KEY INDEXING TERMS: gait, rehabilitation, symmetry, prosthesis design, inertia manipulation

Effects of altering the inertia properties of lower limb prostheses on the mechanics and energetic costs of amputee locomotion have been the focus of a number of studies.[1–14] Lower limb prostheses are often much lighter than the limbs they replace, which creates an inertia asymmetry between prosthetic and intact limbs in unilateral amputees. Much of the existing research[1–14] has examined the implications of these inertial asymmetries on gait metabolic cost and symmetry and the effects of altering prosthesis mass and mass distribution on these same gait properties.

Little is known about the process by which unilateral lower limb amputees adjust to altered inertia properties of the prosthetic limb during walking. Specifically, the time course over which adaptations occur after a prosthesis inertia change is not known. Smith and Martin[15] demonstrated that nonamputees rapidly adjust their walking patterns to asymmetrical lower limb inertia conditions. Subjects wore 1.95 kg (similar to interlimb mass differences often reported for unilateral transtibial amputees) around one ankle for 1 week during which walking symmetry was periodically assessed. Changes in walking symmetry due to the asymmetrical inertia condition were complete within 5 minutes of first exposure to the load. Noble and Prentice[16] focused only on short-term (5 minutes) adaptation to asymmetrical lower limb loading (i.e., 2 kg added to one leg) in nonamputees. All kinematic and kinetic dependent variables reached a steady-state pattern within 45 to 50 strides (i.e., <1 minute) of an increase in inertia of one leg. Thus, results from both studies[15,16] suggest that nonamputees rapidly adjust to changes in lower limb inertia. Unfortunately, these results cannot be generalized to lower limb amputees, and it remains unclear whether amputees respond to altered inertia properties of their prosthesis in a similar manner. When prosthesis mass is altered experimentally, the amount of time given to the amputee to adjust to the new prosthesis mass often is not reported.[2,3,5,14] When accommodation times have been reported, these times have varied from as little as a few minutes to as long as 3 weeks.[1,4,6–8,17] Clinicians, researchers, and amputees would benefit from knowledge of the process and timing of gait adjustments to different prosthesis mass distributions.

The purpose of this study was to quantify the time course of changes in temporal and joint kinetic patterns of walking in unilateral transtibial amputees when the inertial properties of their prostheses were altered substantially and then

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returned to their normal state. Specifically, we investigated alterations in gait mechanics caused by prosthesis inertia changes over a short term (over the first hour) and a longer term (8 days). For the inertia manipulation, we chose to match the inertia properties of the prosthetic limb with the intact limb for three reasons: 1) this required a substantial change in limb mass (~2 kg), 2) previous research[8–10] has focused on this specific manipulation as a possible means of improving walking symmetry in this population, and 3) it is the reverse of the manipulation used in a previous study[15] of nonamputees in which an artificial inertia asymmetry was created by adding 1.95 kg near one ankle. Based on results of the previous study in nonamputees, we hypothesized that walking patterns change immediately after inertia manipulation and that no further changes occur after 5 minutes of exposure to the new prosthesis inertia.

METHODS

PARTICIPANTS

Four unilateral, transtibial amputees (Table 1) participated in this study. All participants wore a prosthesis with a lock and pin suspension system and a dynamic elastic response prosthetic foot. Participants were fully ambulatory, had used a lower limb prosthesis for at least 1 year, and maintained some degree of physical activity either in their vocational and/or daily activities. Participants who reported an ability to walk continuously for 30 minutes or more were identified to ensure they were able to perform both overground and treadmill walking for prolonged periods without substantial fatigue or discomfort during the experimental protocol. All participants in this study reported they had experience with treadmill walking either for exercise purposes or during rehabilitation after being fitted with their prosthesis. The protocol was approved by the University's Institutional Review Board. Written informed consent was obtained from each participant before participation.

SEGMENT INERTIA PROPERTIES

Each participant completed three experimental sessions. Body mass, body height, and lower limb segment lengths were measured before exercise in the first experimental session. Because prosthesis mass was less than the mass of the limb it replaced, body mass was adjusted using the following equation to account for the lost mass before estimating intact segment inertia properties:

$$\mathrm{ABM} = \frac{\mathrm{MBM} - M_{\mathrm{pros}} - M_{\mathrm{residual}}}{1 - \mathrm{c}}$$

where ABM is the adjusted body mass, MBM is the measured body mass while wearing the prosthesis, M_{pros} is the mass of the prosthesis, M_{residual} is the mass of the residual limb, and c is percent of ABM accounted for by the intact shank and foot (0.057 for men; 0.061 for women).[18] Inertia properties of the thigh, shank and foot of the intact leg, and thigh of the prosthetic leg were estimated based on ABM and segment lengths.[18] Inertia properties of the residual limb were estimated by modeling this segment as the frustum of a right circular cone[8,19] and assuming 1.1 $\mathrm{g \cdot cm^{-3}}$ as the uniform tissue density.[20] Residual limb length and two circumferences, one just distal to the knee joint and the other near the distal aspect of the residual limb, were used as inputs into the model.

Inertia properties of the prosthesis were measured experimentally. Mass of the prosthesis was measured using a standard laboratory scale. A reaction board technique[21–24] was used to measure the prosthesis center of mass location, and an oscillation technique[25–27] was used to estimate prosthesis moment of inertia relative to a transverse axis through the knee.

PROSTHESIS INERTIA MANIPULATION

The mass added to the prosthesis was quantified as the difference between the combined mass of the intact shank and foot and the combined mass of the residual limb and prosthesis (Table 2). The added mass was positioned distally on the prosthesis so that the moment of inertia of the prosthetic limb about a transverse axis through the knee was matched with that of the intact limb. This mass was distributed between two concentrated and equally sized packets of lead shot that were affixed to the anterior and posterior aspects of the prosthesis. The location at which the mass was attached was defined as follows:

Table 1. Participant characteristics

Subject	Gender	Age (yrs)	Body mass (kg)	Height (m)	Time since amputation (yrs)	Preferred* walking velocity (m/s)	Cause for amputation
A	Male	44	104.1	1.77	3	1.09	Traumatic injury
B	Female	34	97.3	1.60	10	1.29	Congenital bone disease
C	Male	43	107.3	1.75	5	1.30	Traumatic injury
D	Male	51	101.8	1.83	8	1.08	Diabetes
Mean ± SD		43 ± 7	102.6 ± 4.2	1.74 ± 0.10	6 ± 3	1.19 ± 0.12	

*Preferred walking velocity was determined as the mean of five overground walking trials.

Table 2. Inertia properties of the intact and prosthetic limbs

Subject	Intact[a] mass (kg)	Pros[b] mass (kg)	Est. mass difference (kg)	$Iknee_{intact}$[c] ($kg \cdot m^2$)	$Iknee_{pros}$ ($kg \cdot m^2$)	Intact COM below the knee joint (m)	Pros COM below the knee joint (m)	d[d] (m)
A	6.03	4.27	1.76	0.604	0.325	0.268	0.215	0.396
B	6.07	3.39	2.68	0.400	0.196	0.215	0.177	0.274
C	6.23	4.22	2.01	0.585	0.287	0.260	0.192	0.384
D	6.24	4.11	2.13	0.585	0.262	0.263	0.189	0.389
Mean ± SD	6.14 ± 0.11	4.00 ± 0.41	2.15 ± 0.39	0.544 ± 0.096	0.268 ± 0.054	0.252 ± 0.025	0.193 ± 0.016	0.361 ± 0.058

[a] Intact refers to values for the combined intact shank and foot.
[b] Pros refers to values for the combined prosthesis and residual limb.
[c] Iknee refers to the moment of inertia about a transverse axis through knee.
[d] Distance below knee joint added mass was placed in order to match Iknee between intact and prosthetic limbs.

$$d = \sqrt{\frac{I_{\text{intact}} - I_{\text{pros}}}{m}}$$

where d was the distance distal to the knee, I_{intact} and I_{pros} were the moments of inertia of the intact limb (shank + foot) and prosthetic limb (residual limb + prosthesis) about a transverse axis through the knee joint, and m was the mass added to the limb. The calculated d value for each participant placed the added mass just proximal to the ankle joint (Table 2), which was consistent with the location predicted by Mattes et al.[8]

WALKING PROTOCOL

Each participant completed an 8-day experimental protocol (Figure 1), which included motion analysis and force platform data collection on days 1, 2, and 8. In each test session, the participants completed a walking protocol consisting of both overground and treadmill walking at their preferred velocity. Each participant's preferred walking velocity was the mean velocity of five overground trials completed shortly after the participant arrived in the laboratory on day one (Table 1). Photocells positioned 4.57 m apart were used during overground trials to monitor walking speed during the experiment. Acceptable trials were those within ±3% of the preferred velocity and had no indication of stride adjustment to ensure contact with the force plates.

On day 1, participants were first familiarized with the experimental protocol and practiced walking on the treadmill for a minimum of 10 minutes with no loads attached to their prostheses. Participants then completed three overground walking trials and 5 minutes of treadmill walking with no load added to their prostheses. Data from these trials represented a baseline condition (time 1.U where "1" refers to the test day and "U" refers to an unloaded condition; Figure 1). Participants were then seated at the beginning of the walkway. Mass equal to the estimated difference between their prosthetic and intact limbs was added to the distal aspect of their prosthesis as previously described. Participants immediately completed three acceptable overground walking trials which initiated a 60-minute overground and treadmill walking protocol during which they completed overground trials at 5, 10, 15, 30, and 60 minutes (Figure 1). On completion of the first session, participants were asked to leave the mass attached to their prosthesis as they went about their normal daily activities.

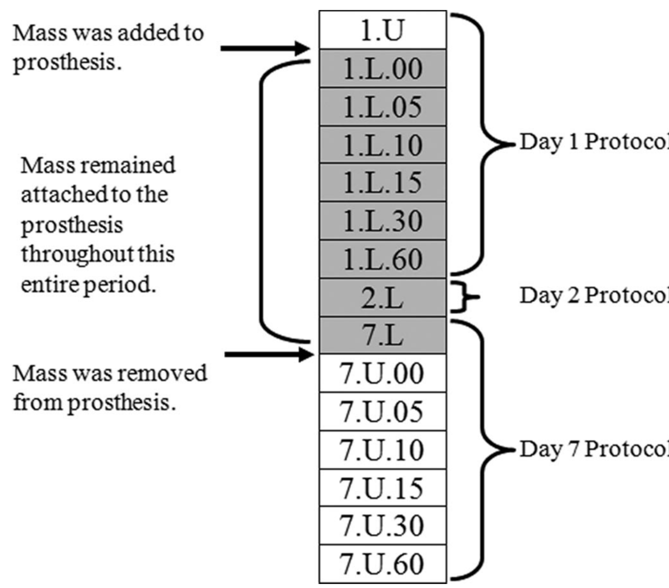

Figure 1. Schematic of the 8-day protocol. The first number in the identifier refers to the day and the next character indicates whether the condition was unloaded or loaded. Time periods with two numbers at the end were days on which the treadmill protocol was completed, where 00 through 60 refer to minutes of a 60-minute treadmill protocol. At each time period, motion and ground reaction forces data were collected during three overground walking trials and 60 seconds of foot switch data were recorded when the subjects walked on a treadmill.

Participants returned to the laboratory approximately 24 hrs later and completed three more overground walking trials and a 5-minute treadmill walking trial with the load still attached (2.L; Figure 1). Participants continued their normal daily activities with the load attached to their prosthesis for the next 6 days. On day 8, participants returned to the laboratory and completed the same walking protocol they performed on day 1, except this time the participants initially completed walking trials (8.L), while still wearing the load. The load was then removed and the participants completed the 60-minute overground and treadmill walking protocol in an unloaded state.

DATA COLLECTION

During overground walking trials, participants walked along a 30 m walkway containing two force plates integrated into the center of the floor. Lower limb motion was captured at 120 Hz during overground trials with a six-camera Motion Analysis tracking system. Retro-reflective markers were attached bilaterally over various anatomical landmarks: greater trochanters, lateral femoral condyles, lateral malleoli, lateral aspect of the heels, and heads of the fifth metatarsals. Markers were placed on the prosthetic limb by mirroring the placement of the same marker on the intact limb. To reduce variability in marker placement between sessions, distances between markers and anatomical landmarks were measured to assist with marker placement during subsequent sessions. Ground reaction forces for two sequential contacts were sampled at 1,200 Hz using the motion analysis system, which synchronized ground reaction forces and motion capture. During treadmill walking, participants walked on a Gaitway instrumented treadmill with two force plates positioned under the belt. Vertical force data during treadmill walking were sampled at 250 Hz for a period of 30 seconds.

During the entire 8-day protocol, participants wore a pedometer that measured how many steps were taken each day. Participants recorded these steps in a personal activity log that was also used to monitor when subjects removed the prosthesis throughout the day and what types of activities individuals participated in during the 8 days. The purpose of having participants wear the pedometer and keep an activity log was twofold: 1) it was a way of assessing how often the mass was removed throughout the protocol, and 2) it provided a metric with which to compare activity levels of participants with reported activity levels in the literature. In other words, it provided an indication of the level of activity of the participants while the additional mass was positioned on the prosthesis.

DATA ANALYSIS

Marker coordinate data from overground walking trials were filtered using a fourth-order, zero-lag, recursive Butterworth digital filter. Cut-off frequencies (4 Hz for hip, 5 Hz for knee, 6 Hz for ankle, and 7 Hz for foot markers) were based on results of a residual analysis.[28] Each overground walking trial was analyzed using six successive foot contact and toe-off events to define one complete stride cycle for each leg. Mean stride, stance, and swing times were determined from the 30 seconds of force plate data collected during the treadmill walking trials.

Using inverse dynamics, sagittal plane resultant joint moments[28] about the ankles, knees, and hips were computed for overground trials. Inertia properties (mass, center of mass, and moment of inertia) of the prosthesis and residual limb were distributed between foot and shank segments for the prosthetic limb based on the ratio of shank and foot masses for a dismantled prosthesis with similar characteristics to the ones used by subjects in this study. During walking trials in which mass was added to the prosthesis, the added mass was modeled as a point mass and the inertia properties of the prosthetic shank were adjusted accordingly. Joint moments were normalized to body weight and height to create a dimensionless expression.[29] Ensemble averages of the three trials for each condition were computed after data were normalized with respect to time (101 points per stride).

Symmetry indices (SI) were computed for mean stance and swing times, peak knee and hip moments near swing initiation and termination, and absolute angular impulses of hip and knee moments. Absolute angular impulse provides an alternative expression to peak joint moments and quantifies the moment profile for the entire gait cycle. This variable has been used previously[9] in the literature and has been referred to as "torque effort." The SI was defined as follows:

$$\mathrm{SI} = \frac{(P - I)}{0.5 \times (P + I)} \times 100$$

where P and I refer to data for prosthetic and intact limbs and a symmetry index of zero represents perfect symmetry.[30]

STATISTICAL ANALYSIS

Eight 1-factor general linear-model analysis of variances (SAS Institute Inc., Cary, NC) with 15 repeated measures reflecting the times at which data were collected during the 8-day protocol were used with follow-up contrasts using pairwise comparisons and an emphasis on comparisons between sequential time points. Statistical significance was determined at $p < 0.05$. Because of the small number of subjects used in the statistical analysis, effect sizes (ESs) were also computed for important pairwise comparisons[31]:

$$\mathrm{ES} = \frac{(M_1 - M_2)}{s_\mathrm{p}}$$

where M_1 is the mean at one time period, and M_2 is the mean at another time period. The pooled standard deviation (s_p) was computed as follows:

$$s_\mathrm{p} = \sqrt{\frac{(s_1^2 + s_2^2)}{2}}$$

where s_1 and s_2 are the standard deviations at each time period. ESs of approximately 0.2 were considered small, those near 0.5 were considered moderate, and those equal to or greater than 0.8 were considered as large ESs.

RESULTS

Data from the daily activity logs indicated participants wore their prosthesis with the added mass attached for all daily activities. Pedometer results showed that on average participants took approximately 5,586 (±2,076) steps per day with the load attached to their prosthesis. To put these data in perspective, it has been suggested that taking 5,000 to 7,499 steps per day is indicative of normal daily activity for individuals classified as low active.[32] Amputees reported that they did not avoid any of their normal daily activities because of the extra mass added to their prostheses.

Notable asymmetries between legs for both temporal and joint kinetic measures were apparent before increasing the inertia of the prosthesis (Figures 2–4; time period 1.U). In the unloaded state, stance time of the intact leg was approximately 10 ms longer than that of the prosthetic leg, whereas swing time of the prosthetic leg was approximately 11 ms longer than that of the intact leg (Table 3). Hip and knee joint moments were greater for the intact leg than the prosthetic leg without the added mass (Table 4).

Asymmetries in stance and swing times were exacerbated when load was attached to the limb (Figure 2). This was due primarily to increased swing times of the prosthetic leg and stance times of the intact leg (Table 3).

Systematic inertia effects were limited to the peak knee moment near swing termination. Asymmetries for the peak knee moment were reduced when load was attached to the prosthesis. Changes in peak knee joint moment asymmetries near swing termination occurred within 5 minutes when prosthesis inertia was altered (Figure 3). Altered SI at the knee after a change in prosthesis inertia was due primarily to increased knee moment magnitudes of the prosthetic leg; the magnitude of the intact knee moment was not altered after a change in prosthesis inertia. Compared with the unloaded baseline condition, the magnitude of the peak knee moment near swing termination of the prosthetic leg was approximately 69% larger with the load attached, and the difference in magnitude for this same moment in the intact leg was approximately 6% (Table 4). SI measures at the hip did not show significant changes as prosthesis inertia changed (Figure 4). This was attributable primarily to similar increases in joint moment magnitudes of both the prosthetic and intact legs (Table 4).

DISCUSSION

Changes in temporal and joint kinetic measures caused by increasing the inertia of transtibial prostheses were investigated over a short term (over the first hour) and a longer term (8 days). Consistent with our hypothesis, changes in stance and swing time SI were apparent in the first assessment after exposure to the increased prosthesis inertia.

TEMPORAL ASYMMETRIES

The observed increases in stance and swing time asymmetries were consistent with previous findings for nonamputees who experienced an artificially created 1.95 kg asymmetry in lower leg inertia characteristics for 1 week[15] and underwent a similar assessment protocol. In both amputees and nonamputees, stance time SI and swing time SI increased (i.e., became more asymmetrical) by approximately 3% and 5%, respectively, after the loads were affixed to one leg. Stance time (SI = −0.2%) and swing time (SI = 0.3%) patterns in nonamputees were highly symmetrical before the inertia manipulation in our previous study; however, amputees in the current study exhibited modest asymmetries for stance time (SI = −2.3%) and swing time (SI = 4.2%) before inertia manipulation. The relative increases in mass and moment of inertia of the loaded leg were similar for both amputees and nonamputees in these studies because approximately 2 kg was added near one ankle in both studies. For the amputees, the added load eliminated an inertia asymmetry between legs, whereas for nonamputees the added mass created an inertia asymmetry. Higher stance time asymmetries in both studies

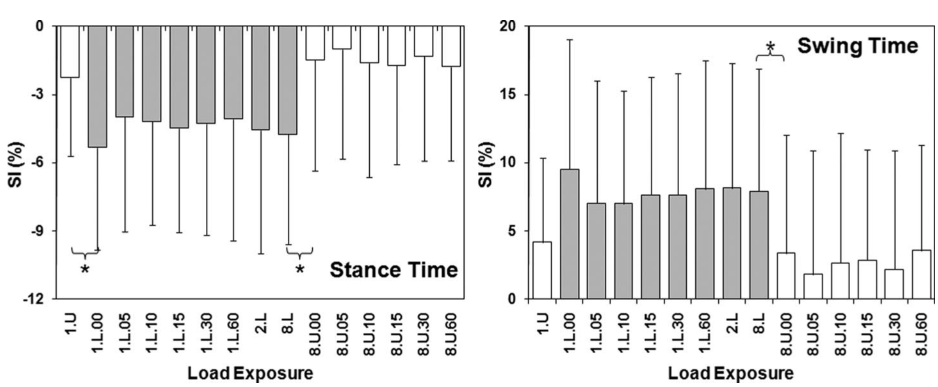

Figure 2. Mean symmetry indices for stance and swing times. Error bars indicate one standard deviation. Horizontal axis labels indicate the time period of the experimental measurement (see Figure 1). Unfilled bars refer to time points when no load was attached to the prosthesis, whereas shaded bars indicate the load was attached to the prosthesis. A negative SI indicates that the magnitude of the intact limb variable was larger than the prosthetic limb variable. *Indicates significant difference. Significant contrasts: (Stance Time: 1.U vs. 1.L.00; $p < 0.028$, ES = 0.77); (Stance Time: 8.L vs. 8.U.00; $p < 0.005$, ES = 0.67); and (Swing Time: 8.L vs. 8.U.00; $p < 0.008$, ES = 0.51). The swing time contrast for 1.U vs. 1.L.00 was not significant ($p < 0.107$, ES = 0.67). ES represents effect size.

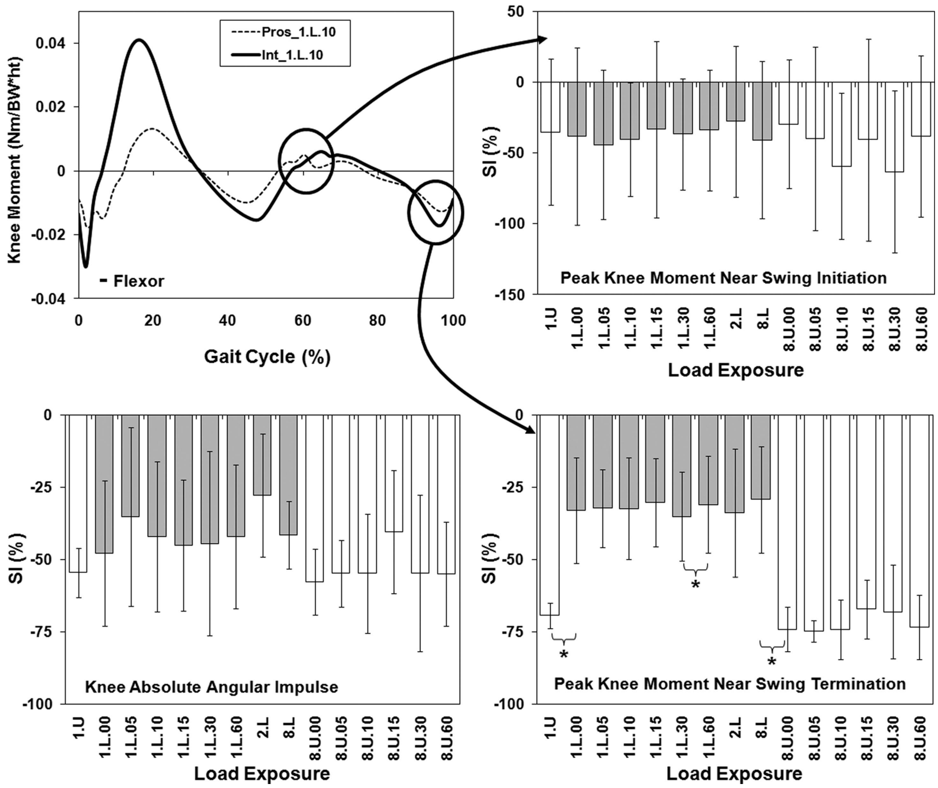

Figure 3. Mean symmetry indices across all participants for peak knee extensor moment near swing initiation, peak knee flexor moment near swing termination, and knee absolute angular impulse knee moment integral. A plot of average (across three trials and all subjects) knee moments for the prosthetic (Pros_1.L.10) and intact (Int_1.L.10) limbs is provided to clarify the focus of statistical analyses. Moment data are normalized to body weight and height. A negative SI indicates that the magnitude of the intact limb variable was larger than the prosthetic limb variable. *Indicates significant difference. Peak knee moment significant contrasts: (1.U vs. 1.L.00; $p < 0.040$; ES = 2.74); (1.L.30 vs. 1.L.60; $p < 0.01$; ES = 0.25); and (8.L vs. 8.U.00; $p < 0.029$; ES = 3.17). ES represents effect size.

were due primarily to higher stance time of the unloaded leg (unloaded leg in nonamputees increased by ~16 ms; intact leg in amputees increased by ~18 ms). Higher swing time asymmetries in both studies were due primarily to higher swing time of the loaded leg (loaded leg in nonamputees increased by ~20 ms; prosthetic leg in amputees increased by ~29 ms). Swing time of the unloaded leg and stance time of the loaded leg were minimally affected by the inertia manipulation in both studies. Therefore, regardless of whether an inertia asymmetry between legs was created or eliminated, changes in temporal patterns of walking for loaded and unloaded legs seem to be driven by the relative increases in mass and moment of inertia of the loaded leg rather than the level of inertia symmetry between legs.

Temporal changes due to higher prosthesis inertia were also consistent with predictions of a force-driven harmonic oscillator (FDHO) model[33] of walking when considered in conjunction with Huygen's law, which accounts for differences in inertia between legs.[34] An FDHO model of locomotion assumes that individuals use minimum amounts of force during locomotion and that the locomotion pattern is largely dependent on the inertia properties of the legs. In consideration of Huygen's law, the FDHO model predicted stride times would increase by approximately 34 ms. Our results showed that stride times increased by approximately 27 ms. Thus, temporal changes seemed to be consistent with the behavior of a passive pendulum model; as inertia increased, the period of oscillation increased.

The observed increases in swing time for the prosthetic leg after the inertia properties of the prosthesis were matched with those of the intact leg were larger than has previously been reported in the literature for a similar manipulation in transtibial amputees. Mattes et al.[8] found that swing time of the prosthetic leg increased on average by 19 ms (4.4%) after prosthesis inertia was increased, whereas we found that swing time increased by 29 ms (7.4%). Mattes et al. had to add approximately 1.70 kg to the prosthesis to match the inertia characteristics between legs, whereas in our study, we needed to add on average 2.15 kg. Mattes et al. found that prosthetic leg swing time increased more for larger masses, and because we added more mass at the same location as Mattes et al., it is likely that the greater mass used in our study contributed to larger increases in swing time.

Mattes et al. also reported that stance time of the prosthetic leg decreased, whereas stance time for the intact leg increased after matching inertia properties between legs. Consequently, the stance time symmetry index increased (i.e., became less symmetrical) by approximately 5% after the

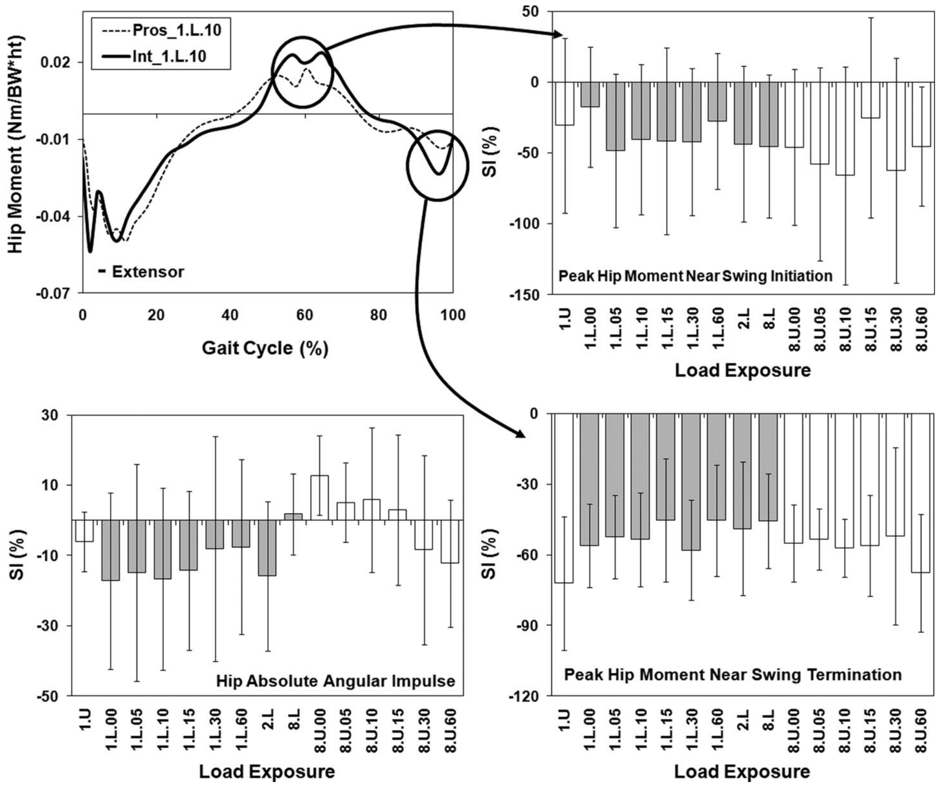

Figure 4. Mean symmetry indices for the peak hip flexor moment near swing initiation, peak hip extensor moment near swing termination, and hip absolute angular impulse. Average (across three trials and all subjects) hip moment plots are provided to clarify statistical focus for the hip. Moment data are normalized to body weight and height. A negative SI indicates that the magnitude of the intact limb variable was larger than the prosthetic limb variable. No significant changes were found for any of the hip variables.

Table 3. Mean (SD) stance and swing times for intact and prosthetic legs for loaded (1.L.00–8.L) and unloaded (1.U, 8.U.00–8.U.60) walking trials

	Stance time (ms)		Swing time (ms)	
	Prosthetic leg	Intact leg	Prosthetic leg	Intact leg
Unloaded	710 (58)	720 (32)	393 (16)	382 (28)
Loaded	708 (56)	738 (25)	422 (34)	391 (40)

These results were not statistically analyzed.

inertia manipulation. Consistent with findings of Mattes et al., we not only observed increases in stance time for the intact leg (by ~18 ms) after the inertia perturbation but also observed a slight increase (~2 ms) in stance time for the prosthetic leg. Thus, in comparison with Mattes et al., we observed a smaller (~3%) increase in the stance time symmetry index after the inertia manipulation. Nevertheless, the results of both studies clearly demonstrate that temporal characteristics of gait are altered when prosthesis inertia properties are matched with those of the intact leg and our results suggest that these changes occur immediately after the increase in inertia. Furthermore, both studies indicate that matching inertia properties of transtibial prostheses with those of the intact leg is detrimental to a symmetrical walking pattern.

JOINT KINETIC ASYMMETRIES

Joint moments at the hip and knee did not reflect the effect of prosthesis inertia as clearly as stance and swing times. The knee joint moment near swing termination became more symmetrical between prosthetic and intact limbs immediately after the increase in prosthesis inertia. The improvement (i.e., reduction) in symmetry was due to an increase in the prosthetic leg's knee moment during late swing, suggesting a greater effort required to negatively accelerate the limb just before ground contact. SI of the hip moment integral suggested hip moments during early exposure on day 1 (1.L.00 through 1.L.15) to the increased inertia differed significantly from those initially after removal of the mass on day 8 (8.U.00 through 8.U.15). These differences were primarily due to a larger extensor moment of the intact leg during the first half of stance after inertia was matched between legs. For all other joint kinetic measures, the lack of significant change in symmetry was due to compensatory actions of the intact leg. As prosthesis inertia increased, moment magnitudes of the intact leg increased. Joint moment magnitudes are indicative of muscular effort at the joint level,[9] particularly for motions such as walking that do not reach the limits of the joint's range of motion. Thus, increased moment

Table 4. Mean (SD) knee and hip moment dependent variables for intact and prosthetic legs for loaded (1.L.00–8.L) and unloaded (1.U, 8.U.00–8.U.60) walking trials

	Knee		Hip	
	Prosthetic leg	Intact leg	Prosthetic leg	Intact leg
Peak moment near swing initiation ($N \cdot m \cdot BW^{-1} \cdot ht^{-1}$)				
Unloaded	0.0049 (0.0034)	0.0077 (0.0039)	0.0162 (0.0061)	0.0258 (0.0073)
Loaded	0.0069 (0.0053)	0.0100 (0.0061)	0.0200 (0.0056)	0.0296 (0.0082)
Peak moment near swing termination ($N \cdot m \cdot BW^{-1} \cdot ht^{-1}$)				
Unloaded	−0.0075 (0.0007)	−0.0160 (0.0021)	−0.0110 (0.0025)	−0.0208 (0.0071)
Loaded	−0.0125 (0.0009)	−0.0172 (0.0009)	−0.0135 (0.0021)	−0.0230 (0.0054)
Integral of the rectified moment ($[N \cdot m \cdot BW^{-1} \cdot ht^{-1}] \times s$)				
Unloaded	0.7109 (0.0946)	1.2335 (0.2054)	1.6034 (0.3352)	1.5905 (0.2378)
Loaded	0.8212 (0.1633)	1.2390 (0.1621)	1.6109 (0.3606)	1.8005 (0.3006)

These results were not statistically analyzed.

magnitudes found in this study suggest a greater muscular effort of walking with increased prosthesis inertia.

The lack of significant changes in all but two joint kinetic variables was surprising because Smith and Martin[15] observed in nonamputees that four of the six joint moment variables (three knee, one hip) changed significantly within 5 minutes of exposure to an increased leg inertia. Variability (i.e., standard deviations) in joint moment SI for amputees was approximately twice that of nonamputees in the previous study.[15] In addition, the sample size was small in our study. These two factors likely contributed to the lack of significant changes in kinetic variables found for the amputees in this study. For example, we were unable to detect statistically significant changes in the hip moment near swing termination immediately after addition (1.U vs. 1.L.00) or removal (8.L vs. 8.U.00) of the additional of the mass, although qualitatively this variable seemed to be more symmetrical when amputees were walking with the additional mass (cf., Figure 4). In addition, effects sizes for these contrasts were 0.68 and 0.52, respectively, which were moderate ESs, suggesting a meaningful difference. The greater variability of the amputee walking patterns has been noted previously by investigators.[35,36] Selles et al.[9,10] have also reported that over the short term (<5 minutes) adding as much as 1 kg distally to a transtibial prosthesis significantly increases the magnitudes of hip and knee joint moments of the prosthetic leg during the swing phase of walking. Therefore, our results are consistent with those of Selles et al.[9,10] and suggest that although temporal asymmetries are exacerbated by distal loading, joint kinetic asymmetries are reduced to some extent. Unfortunately, the improved joint kinetic symmetry would likely result in an increased metabolic cost for the amputee during walking as others have reported for a similar inertia manipulation.[8]

Noble and Prentice[16] found that adaptations to increases in leg inertia occurred rapidly, which is consistent with our results. Their results indicated nonamputees adapted their mechanics to accommodate an increase in leg inertia within 45 to 50 strides, but it took as long as 70 strides for some variables to return to baseline values after the load was removed. Given that their participants walked at $1.56 \text{ m} \cdot \text{s}^{-1}$, this suggests that accommodation to a change in leg inertia took approximately 1 minute, which is within the window between our first assessment (1.L.00) and our second assessment 5 minutes later (1.L.05).

Several limitations of our study should be noted. First, because of our small sample size, caution should be exercised when generalizing these outcomes to all lower limb amputees. Second, because temporal measures were assessed during treadmill walking and joint kinetic measures were assessed during overground walking, these estimates did not occur at the same time. Temporal assessments were performed after at least some exposure to the load because overground walking trials occurred before treadmill trials. Because most of the effects due to load are exhibited during swing, future studies could limit kinetic analyses to the swing phase, which would make it easier to estimate temporal and joint kinetic variables at the same time as was done by Noble and Prentice. Finally, the results of this study are limited to conclusions based on the point at which accommodation to the load was complete (i.e., 5 minutes), because we were not able to capture every single stride in our assessments. Further insights into the adaptation process itself can be gained by investigating every single stride. Future studies focusing only on the swing phase or using a treadmill with multidimensional force plates should be able to address this limitation.

CONCLUSION

In conclusion, matching the inertia properties of the prosthesis with those of the intact leg immediately altered temporal and joint kinetic symmetry between legs after the addition of the mass to the prosthesis. Removal of the additional mass from the prosthesis resulted in an equally rapid return to baseline levels of symmetry for both temporal and joint kinetic measures. In addition, matching inertia properties of the prosthesis does not seem to benefit unilateral transtibial amputees. Our inertia manipulation created a more asymmetrical walking pattern in terms of temporal characteristics, but a more symmetrical joint kinetic pattern during the swing phase. Temporal changes were consistent with predictions from a passive pendulum model, although loading effects on net joint moments suggest the response to the inertia manipulation was not purely passive. As inertia increased, greater muscular efforts were required to control the motion of the higher limb inertia during walking.

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