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Walking patterns change rapidly following asymmetrical lower extremity loading

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Abstract

The purpose of this study was to determine the amount of time needed for individuals to become well accommodated to asymmetrical changes in lower extremity inertial properties. Participants walked at 1.57 m s^{-1} during four separate data collection sessions over the period of eight days. On days 1 and 7, participants completed a 60 min treadmill protocol consisting of both overground (motion and ground reaction force recorded) and treadmill (vertical ground reaction force recorded) walking. On day 1, 1.95 kg was attached distally to one shank prior to the start of the 60 min treadmill protocol and was not removed until day 7. On day 7, the load was permanently removed prior to the 60 min treadmill protocol. On days 2 and 8 participants completed three overground walking trials and walked on the treadmill for approximately 5 min. Stance and swing time asymmetries appeared immediately and were complete following initial assessment after the load was attached. Net joint moments at the knee and hip were altered and continued to change beyond initial exposure to the load, but these changes were complete within 5 min. Overall, results suggest that changes in walking symmetry due to asymmetrical lower extremity loading are immediate and complete within 5 min of exposure to the load. We recommend that at least 5 min of walking or other normal activity be used to accommodate individuals to novel asymmetrical lower extremity loading.

PsycINFO classification: 2330; 2221; 4010

Keywords: Inverse dynamics; Accommodation; Symmetry; Inertial manipulation; Amputee

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1. Introduction

Lower extremity loading has been utilized to investigate effects of inertial manipulations in both amputees (Czerniecki, Gitter, & Weaver, 1994; Donn, Porter, & Roberts, 1989; Gitter, Czerniecki, & Meinders, 1997; Hale, 1990; Hillery & Wallace, 2000; Hillery, Wallace, McIlhagger, & Watson, 1997; Mattes, Martin, & Royer, 2000; Skinner & Mote, 1989) and able-bodied individuals (Claremont & Hall, 1988; Martin, 1985; Martin & Cavanagh, 1990; Myers & Steudel, 1985; Skinner & Barrack, 1990) during locomotion. The recent impetus for these studies has stemmed from the fact that lower limb prostheses are substantially lighter than the limbs they replace. Prescribing lightweight prostheses generally centers on the argument that a lighter prosthesis will require less effort by the musculature, which ultimately reduces the metabolic costs associated with locomotion. Contrary to this practice, several investigators using computer simulations based on passive pendulum models have suggested the inertial asymmetries between intact and prosthetic limbs may contribute substantially to gait asymmetries (Bach, 1995; Mena, Mansour, & Simon, 1981; Tsai & Mansour, 1986). The effects of altering mass and moment of inertia properties of the lower extremity on locomotion are not clearly understood. Results from studies investigating these effects are often equivocal.

A limitation of many lower extremity loading studies is that little is known about the process of adapting to inertial changes during walking. Specifically, the amount of time it takes individuals to adapt to altered inertial properties of the lower extremity is not known (Selles, Bussmann, Wagenaar, & Stam, 1999). Many lower extremity loading studies often do not report how much time was provided to participants to accommodate to loads before data were collected (Donn et al., 1989; Hale, 1990; Hillery & Wallace, 2000; Martin, 1985; Martin & Cavanagh, 1990; Tashman, Hicks, & Jendrzejczyk, 1985). When accommodation times have been reported, these times have varied from as little as a few minutes to as long as three weeks (Czerniecki et al., 1994; Gitter et al., 1997; Hillery et al., 1997; Huang, Chou, & Su, 2000; Mattes et al., 2000; Skinner & Barrack, 1990). Initial use of prosthetics or orthotics and changes in footwear design are examples in which investigators would benefit from knowledge about reasonable accommodation times. In addition, investigating adaptations to novel inertial manipulations of the lower extremity would provide further insights into neural adaptations to altered limb inertia in general.

Immediately following an inertial manipulation of the lower extremity the kinematic movement patterns and/or underlying kinetics must change to meet the demands of the new mechanical constraints. In unilateral below-knee amputees, Selles and colleagues (Selles et al., 2004) recently reported that only 4 out of 22 investigated kinematic variables (all were measures of thigh kinematics) were altered when prosthesis inertia increased, whereas 13 out of 20 investigated kinetic (6 thigh; 7 knee) variables were altered during the swing phase of walking. Thus, it appears that kinetics change in an effort to accommodate the increased lower extremity inertia, while kinematic patterns remain similar to an unperturbed condition. However, it remains unclear as to whether the kinetic patterns would remain the same over a longer term because amputees in this study were only exposed to the inertial manipulation for approximately 5 min. In addition, Selles and colleagues did not report changes in temporal features of the walking pattern although others have shown that increasing lower extremity moment of inertia increases swing time for the affected leg (Mattes et al., 2000; Skinner & Barrack, 1990).

In this study, we were specifically interested in the time course of changes in an individual's gait pattern produced by abrupt and asymmetrical changes in lower extremity inertial properties produced by the addition of 1.95 kg near the ankle. This mass approximated the difference between intact and prosthetic lower limbs previously reported for unilateral, transtibial amputees (Mattes et al., 2000). We defined "well accommodated" as the point at which no further statistically significant changes in walking symmetry occurred following a change in lower extremity inertia. We focused on symmetry measures in an effort to reduce the number of variables needed in our statistical analysis. In addition, previous research (Skinner & Barrack, 1990) has shown that walking symmetry is altered following an asymmetrical lower extremity inertial perturbation, but it is unclear whether symmetry changes immediately or over a period of time. It was our expectation that quantifiable changes in the symmetry of temporal and kinetic gait descriptors resulting from a lower extremity inertial asymmetry would be immediate because of the mechanical constraints of the system. More importantly, we expected the altered state of symmetry reflected after a period of short term exposure to the asymmetrical inertia condition (~5–10 min) would show no further change with continuing exposure over a 7-day period. Stated differently, we did not expect a gradual, long term adaptation process to the inertial modification.

2. Methods

2.1. Participants

Four male and two female young healthy adults (age = 24 ± 4 years, height = 1.77 ± 0.08 m, and mass = 72.9 ± 8.6 kg), free of any noticeable gait abnormalities, participated in the study. All participants had experience either walking or running on a treadmill prior to this study. Informed written consent was obtained from each participant prior to participation, and approval for the protocol was obtained from the Institutional Review Board at Pennsylvania State University.

2.2. Data acquisition

Each participant completed an eight day experimental protocol (Fig. 1), which included data collection on days 1, 2, 7 and 8. Participants wore tight-fitting shorts made of elastic material (i.e., lycra or spandex shorts) and the same footwear for all test sessions. On each day, participants completed both overground and treadmill walking trials at a speed of 1.57 m s^{-1} (3.5 mph). The walking speed used in this study was slightly higher than previously reported preferred walking speeds for young healthy adults and was chosen in an effort to present a slightly more demanding task to the participants. During overground trials, participants walked along a 30 m walkway containing two force plates (Kistler Instrument Corporation, Amherst, NY) integrated into the center of the floor. During these trials, lower extremity motion (60 Hz) and ground reaction forces (480 Hz) for two sequential foot contacts were captured simultaneously using a seven camera Vicon 370 motion analysis system (Vicon Motion Systems, Oxford, UK). Photocells, positioned 4.57 m apart, were used during overground trials to monitor walking speed. Acceptable trials were those within ±3% of the target speed and had no indication of stride adjustment to contact the force plates. During treadmill walking, participants walked on a Gait-

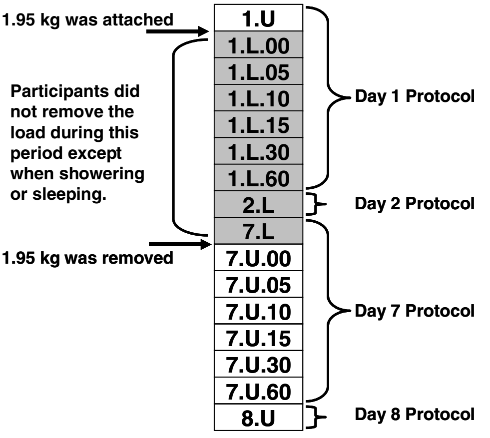

Fig. 1. Schematic of the eight day protocol. The first number in the identifier refers to the day and the next two characters refer to the time period, where 00 through 60 refer to minutes of a 60 min treadmill protocol and L and U refer to loaded and unloaded, respectively. At each time period, motion and GRF data were collected during three overground walking trials and 60 s of force data were recorded while participants walked on an instrumented treadmill.

way instrumented treadmill (Kistler Instrument Corporation, Amherst, NY) with two force plates positioned under the belt. Force data during treadmill walking were sampled at 250 Hz for a period of 60 s for each condition using Kistler's software. Temporal data were measured on the treadmill rather than overground because foot contacts could be estimated directly from the vertical force components and data could be captured during continuous walking. Joint kinetic data were not measured during treadmill walking because the treadmill force plates were only capable of measuring vertical forces and shear forces are also needed for joint kinetics analyses.

On day 1, participants were familiarized with the experimental protocol and were allowed to practice walking on the instrumented treadmill for approximately 10 min without any loads attached to either limb. Retro-reflective markers were then attached bilaterally to various anatomical landmarks: greater trochanters, lateral femoral condyles, lateral malleoli, lateral aspect of the heels, and heads of the fifth metatarsals. After all markers were attached, the distances between markers and various anatomical landmarks were measured using a standard anthropometric tape and recorded so that markers could be carefully repositioned in the same locations on subsequent test days.

Each participant then completed three overground trials and 5 min of walking on the instrumented treadmill at the experimental walking speed. Motion and GRF data from overground trials and GRF data from treadmill trials served as unloaded baseline data for the first session (time 1.U, where "1" refers to the test day and "U" refers to an unloaded condition; see Fig. 1). Treadmill force data were collected during the last 60 s of the 5 min walking bout on the treadmill.

The participant was then positioned in a chair at the beginning of the overground walkway and 1.95 kg was attached to either the right or left leg (3 participants wore the load on the left leg and 3 wore the load on the right leg) using four packets of lead shot with approximately equal masses. The 1.95 kg load added to the leg approximated the difference between the intact and prosthetic lower limbs previously reported for unilateral, transtibial amputees (Mattes et al., 2000). Packets of lead shot were equally distributed to completely encompass the lower shank just above the medial and lateral malleoli and were secured using an elastic bandage. This location was similar to that used by Mattes et al. (2000) when adding mass to the prosthetic limb to match the prosthetic and intact lower leg moments of inertia about a transverse axis through the knee. The distance between the lateral femoral condyle marker and center of the added mass was measured and recorded in order to replicate the position of the mass in subsequent sessions and for use in later analyses. Participants then immediately completed three overground walking trials without any load accommodation. All participants were successful in completing overground trials at the targeted speed within one (~5 strides) or two (~15 strides) trials. After completing the overground trials and while still wearing the load, the participant immediately returned to the treadmill. Sixty seconds of force data (time period 1.L.00) during treadmill walking were collected as soon as the belt reached the nominal speed.

From this point, the participant completed a 60 min treadmill/overground walking protocol, during which treadmill walking was stopped periodically at 5, 10, 15, 30, and 60 min and the participant completed three more overground walking trials. Treadmill data were recorded during the last 60 s of each treadmill walking bout, which immediately preceded the three overground walking trials for that time period. At the conclusion of the 60-min protocol, the added mass was removed and immediately replaced with a commercial ankle weight, which was easier for the participants to attach and remove. The commercial and experimental masses were identical (1.95 kg). Additionally, the dimensions of the experimental mass packets and the commercial weight were similar. Because both were positioned similarly on the leg (distal end), changes in center of mass location and moment of inertia about a transverse axis through the knee joint were quite similar for the two mass systems. The commercial ankle weight was not worn during data collection because it interfered with the malleolus marker. Participants were asked to wear the ankle weight at all times except when showering or sleeping.

Participants returned to the lab 24 h after the first session for a second data collection session. Reflective markers were repositioned over anatomical landmarks and the commercial ankle weight was replaced with the four lead packets used during the first session. Using the same methods employed in session one, motion and ground reaction force data were collected for three overground trials and one 5 min treadmill trial. Once all data had been captured the lead packets were again replaced with the commercial ankle weight and participants were again instructed to wear ankle weights at all times during the next six days except when showering or sleeping. On day 7, the participant returned to laboratory to complete an experimental protocol similar to the first session. The only difference between the protocols for days 1 and 7 was the loading condition. On day 7, trials were initially collected with the load attached (7.L; Fig. 1). The load was then removed with participants seated in a chair at the beginning of the walkway. The same 60 min overground/treadmill walking protocol used on day 1 was then repeated in an unloaded state. Following 24 h of normal daily activities without the added load, the participant returned for a final session on day 8. The final session was similar to the second session, but in this session the participant again walked without any added mass. During the entire eight 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 participants removed the ankle weight throughout the day and what types of activities individuals participated in during the eight days.

2.3. Data analysis

2.3.1. Treadmill walking trials: Temporal data

Treadmill force data were processed within Kistler's Gaitway software. The vertical force profile during each stride for each 60 second treadmill trial was used to determine stride and stance times for each leg within Kistler's software. These data were then exported to Microsoft Excel and swing time during each stride was computed as the difference between stride time and stance time. Participants completed approximately 50 to 55 strides during each 60 second treadmill trial. Mean stance and swing times for each leg were computed across all completed strides within each 60 second treadmill trial. Symmetry indices (SI) were then computed using these mean stance and swing times to quantify temporal differences between legs at each time period during the walking protocol (see Fig. 1). The SI was defined as

$$\mathrm{SI} = \frac{(L - U)}{0.5 * (L + U)} * 100, \tag{1}$$

where L and U refer to data for loaded and unloaded limbs and a symmetry index of zero represents perfect symmetry. SI for stance time and swing time served as two dependent variables in our statistical analyses.

2.3.2. Overground walking trials: Joint kinetic data

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 residual analysis (Winter, 1990). Each overground walking trial was analyzed using six gait events (three events for each leg) to define one complete stride cycle for each leg, which occurred in the following order:

1. right foot contact on the first force plate,
2. left foot toe-off from the ground prior to the first force plate,
3. left foot contact on the second force plate,
4. right foot toe-off from the first force plate,
5. right foot contact on the ground after the second force plate, and
6. left foot toe-off from the second force plate.

Therefore, stride cycles were defined from event 1 to event 5 for the right leg and from event 2 to event 6 for the left leg. An algorithm with a threshold based on the mean plus one standard deviation of the vertical force for 100 samples prior to foot contact was used to determine heel strike and toe-off events when these events occurred on a force plate (events 1, 3, 4, and 6). For events 2 and 5, which did not occur on a force plate, the horizontal velocity of the toe marker was used to determine foot contact and toe-off. The horizontal velocity of the toe marker during the same event, when it occurred on a force plate, was used to determine the threshold for this algorithm.

Each lower extremity was modeled as a series of three rigid bodies connected by frictionless revolute joints. A two-dimensional inverse dynamics approach was used to compute resultant joint moments (Winter, 1990) about the ankles, knees, and hips for each overground trial. Segment inertial properties were estimated based on regression equations

Table 1
Shank mass, moment of inertia and center of mass changes due to addition of 1.95 kg

	Absolute change	Relative change (% of NL)
Increase in shank mass (kg)	1.95	60.3 (5.9)
Increase in I_{knee} for shank (kg m^2)	0.073 (0.006)	75.9 (9.5)
Distal shift in shank COM relative to knee axis (m)	0.274 (0.058)	43.9 (7.2)

Note: Numbers in parentheses represent one standard deviation.

from the literature (De Leva, 1996). During walking trials in which mass was added to the shank, the added mass was modeled as a point mass located a known distance inferior to the knee joint center. The affected shank's mass, center of mass location, and moment of inertia were adjusted to reflect the added load (Table 1). Center of pressure data were converted to a global laboratory reference frame so that coordinate data and center of pressure data were in the same frame of reference.

For each individual overground walking trial, six (3 hip and 3 knee) discreet dependent variables for joint kinetics were identified: (1) peak knee moment near swing initiation, (2) peak knee moment near swing termination, (3) peak hip moment near swing initiation, (4) peak hip moment near swing termination, (5) integral of rectified knee moment, and (6) integral of rectified hip moment. After variables had been identified in each trial, these discreet variables were averaged across the three trials for each condition, except for the trials immediately following load application on day 1 (1.L.00; Fig. 1) or removal on day 7 (7.L.00; Fig. 1). For these conditions, the first trial in which complete contact was made with both force plates was used to represent initial exposure. The reason for using only the first trial was that this trial would best capture the immediate alteration in gait mechanics as a result of the change in load condition. Symmetry indices (refer to Eq. (1)) were then computed for each of the six joint moment variables and were used as six dependent variables in our statistical analyses. Hip and knee moments were the focus of this analysis because we expected the load to primarily affect the hip and knee since the load was attached superior to the ankle joint (Martin & Cavanagh, 1990).

2.4. Statistical analysis

Eight one-factor general linear-model ANOVAs (Minitab Inc., State College, PA) with 16 repeated measures reflecting the times at which data were collected over the 8-day protocol were conducted to test the null hypothesis that SI were identical across all time periods. When the null hypothesis was rejected, follow up contrasts were conducted using Tukey's Wholly Significant Difference test. Statistical significance was determined at $p < .05$. Due to the small number of participants used in the statistical analysis, effect sizes (ES) were computed for important pairwise comparisons according to Thomas and Nelson (Thomas & Nelson, 1996)

$$\mathrm{ES} = \frac{(M_1 - M_2)}{s_{\mathrm{p}}}, \tag{2}$$

where M_1 is the mean at one time period and M_2 is the mean at another time period. Effects sizes greater than or equal to 0.8 were considered large, effect sizes around 0.5 were considerate moderate, and effect sizes equal to 0.2 or less were considered small. The pooled standard deviation (s_p) was computed as follows:

$$s_p = \sqrt{\frac{(s_1^2 + s_2^2)}{2}}, \tag{3}$$

where s_1 and s_2 are the standard deviations at each time period.

3. Results

Subjective data from the daily activity logs confirmed that participants adhered to the protocol of removing the ankle weight only when showering or sleeping. Pedometer results also showed that participants took approximately 6574 ($\pm$2904) steps per day and took a similar number of steps on those days on which the ankle weight was worn (6651 $\pm$ 1112 steps) as they did on days without the ankle weight (6383 $\pm$ 2351 steps).

Participants initially displayed symmetrical walking patterns when first observed in the absence of an inertial asymmetry (Figs. 2–4; time period 1.U). This was true for all dependent variables being studied. Symmetry was subsequently disrupted by the application of 1.95 kg to the distal aspect of one leg. The resulting asymmetries were immediate for most dependent variables and showed little additional change with continued exposure to the load over the next six days. The removal of the load on day 7 of the protocol produced equally abrupt changes back to a symmetrical walking pattern.

More specifically, application of 1.95 kg to the distal part of the shank affected stance ($F(15,75) = 27.48$, $p < .001$) and swing time ($F(15,75) = 27.48$, $p < .001$) symmetries between legs (Fig. 2). Statistically significant changes occurred immediately following application of the load to the ankle on day 1 (1.U vs 1.L.00: $p < .001$; ES $\approx$ 3.9 for both stance and swing times) and removal of the load on day 7 (7.L vs 7.U.00: $p < .006$; ES $\approx$ 3.0 for both). These significant changes in stance time and swing time symmetry can be attributed primarily to increases in swing time for the loaded limb and stance time for the unloaded limb (Table 2).

Symmetry indices for peak knee extensor moment ($F(15,75) = 8.83$, $p < .001$), peak knee flexor moment ($F(15,75) = 198.21$, $p < .001$), and knee moment integral ($F(15,75) = 24.37$, $p < .001$) exhibited significant differences among time periods. Results for the peak knee

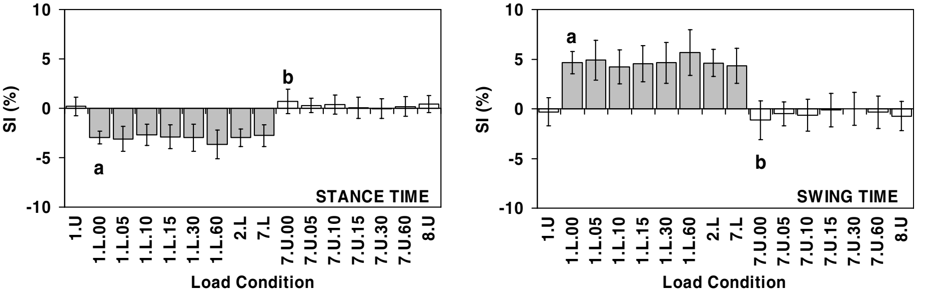

Fig. 2. Mean symmetry indices across all participants for stance and swing times. Statistically significant (a) asymmetries appeared immediately and remained throughout the period of seven days. No other significant differences were found throughout this period. A positive SI indicates that the time for the loaded limb was longer than the unloaded limb. Immediately following removal of the load, stance and swing times became symmetrical again (b) and were consistent with symmetry levels during 1.U.

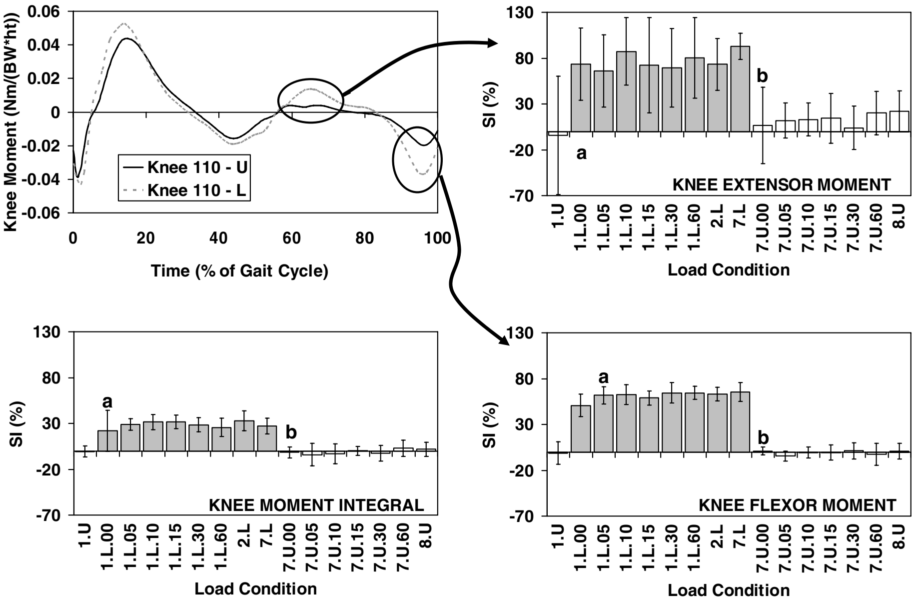

Fig. 3. Mean symmetry indices across all participants for peak knee extensor moment near swing initiation, peak knee flexor moment near swing termination, and rectified knee moment integral. A plot of average (across three trials and all participants) knee moments for unloaded (U) and loaded (L) limbs is provided to clarify statistical focus for the knee. Moment data are normalized to body weight and height. A positive SI indicates that the magnitude of the loaded limb variable was larger than the unloaded limb variable. With application of the load, noticeable asymmetries appeared immediately. For the peak knee extensor and knee moment integral, no further changes in symmetry occurred until the load was removed. However, the peak knee flexor moment continued to change beyond initial exposure to the load, but changes were complete within 5 min. Immediately following load removal, all variables returned to baseline symmetry levels (1.U) and did not change significantly over the next 24 h. a – indicates last time period in which significant changes occurred following load application. b – indicates last time period in which significant changes occurred following load removal.

extensor moment SIs and knee moment integral SIs were identical to results for stance and swing times with similar p-values and effect sizes. Peak knee flexor moments changed immediately (1.U vs 1.L.00: $p < .001$, ES = 4.3). The knee flexor moment during 1.L.00 differed significantly from all from all other loaded conditions, except for 1.L.05 ($p = .08$; ES = 1.00) and 1.L.15 ($p = .48$; ES = 0.80). All effect sizes for these comparisons were greater than 0.8, which suggests a meaningful difference was present. However, 1.L.05 was not significantly different from any of the loaded time periods that followed (1.L.10 through 7.L), which suggests that changes in this variable were complete within the first 5 min of exposure to the new load. Immediately following load removal, symmetry returned to levels consistent with baseline (i.e., 1.U). In absolute terms, knee moments of the unloaded limb were not altered after the mass was added to the contralateral limb. Thus, changes in the symmetry indices for knee moment variables can be attributed to increases in magnitude of the loaded limb's knee moment.

At the hip, significant differences were only found for the peak hip extensor moment near swing termination ($F(15, 75) = 7.48$, $p < .001$). However, changes in symmetry for this variable did not occur immediately after loading or unloading. Rather, the first notable change in symmetry occurred 5 min after load application (1.L.00 vs 1.L.05: $p = .01$;

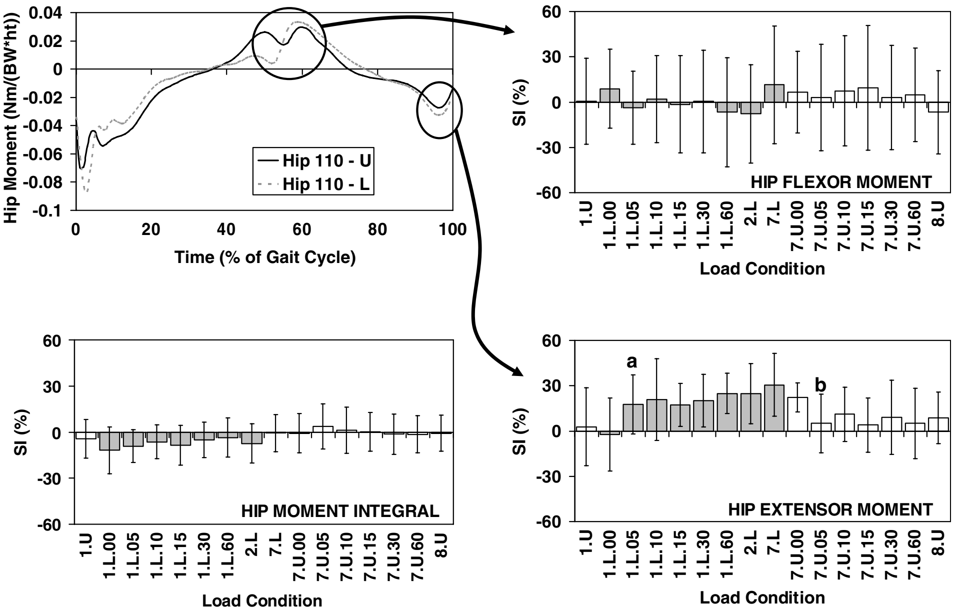

Fig. 4. Mean symmetry indices across all participants for the peak hip flexor moment near swing initiation, peak hip extensor moment near swing termination, and hip moment integral. Average (across three trials and all participants) hip moment plots are provided to clarify statistical focus for the hip. Moment data are normalized to body weight and height. Changes in symmetry at the hip due to asymmetrical loading were only observed in the peak hip extensor moment near swing termination. Results were similar to the results for the peak knee flexor moment near swing termination. However, changes occurred not only during the first 5 min following application of the load but also during the first 5 min following removal of the load. a – indicates last time period in which significant changes occurred following load application. b – indicates last time period in which significant changes occurred following load removal.

Table 2
Mean stance and swing times across all participants during symmetrically (no load on either leg) and asymmetrically loaded (1.95 kg added to one leg) walking conditions

	Stance time (ms)		Swing time (ms)	
	Loaded leg	Unloaded leg	Loaded leg	Unloaded leg
Loaded (1.95 kg asym)	608 (3)	626 (4)	404 (4)	385 (3)
No load (sym)	612 (2)	610 (4)	384 (5)	386 (4)

Note: Numbers in parentheses represent one standard deviation.

ES = 0.9) and 5 min after load removal (7.L vs 7.U.05: $p < .001$; ES = 1.3). In absolute terms, hip moments of the unloaded limb were not altered after the mass was added to the contralateral limb. Thus, changes in the symmetry indices for hip moment variables can be attributed to increases in magnitude of the loaded limb's hip moment.

4. Discussion

Our results were consistent with our hypothesis that changes in walking symmetry, reflected by temporal and kinetic asymmetries, due to asymmetrical lower extremity inertial conditions occur within the first 5–10 min of exposure to the load. Asymmetries in stance and swing time symmetries were apparent from the first treadmill assessment following application of the 1.95 kg load to one ankle, and did not change with further exposure to the load. However, hip and knee moments exhibited further changes between the initial exposure (1.L.00) and after 5 min of exposure (1.L.05). No further changes occurred in the hip or knee moment after time period 1.L.05 until the load was removed on day seven (7.U.00). Upon removal of the load, walking symmetry returned to baseline levels within 5 min. Thus, our data do not reflect the presence of a long term, gradual adaptation in gait mechanics to a change in lower extremity inertia.

The observed increases for stance time of the unloaded leg and swing time of the loaded leg as a result of asymmetrical loading were similar to results from Skinner and Barrack (1990) for similar loading conditions. When 1.82 kg was added to one ankle, Skinner and Barrack observed increases of approximately 20 ms in stance time for the unloaded leg and swing time for the loaded leg. Estimated symmetry indices for stance and swing time were −5.0% and 6.4%, which are consistent with, although slightly greater than our computed symmetry indices (Fig. 2). A force-driven harmonic oscillator (FDHO) model has been used for predicting stride times during walking (Holt, Hamill, & Andres, 1990) and in consideration of Huygen's law has been used for predicting stride times following asymmetrical mass manipulations in the lower extremity (Lin-Chan, Bilodeau, Yack, & Nielsen, 2004). Huygen's law implies the actual stride time should be intermediate to the predicted stride times for the loaded and unloaded legs (Lin-Chan et al., 2004). For our participants, the FDHO model predicted the added mass would cause the stride time of the loaded limb to increase by approximately 57 ms when compared to the prediction for the unloaded leg. Based on Huygen's law, stride time for both legs was expected to increase by approximately 32 ms. The increase in stride time we observed was in the predicted direction, but was a more modest 16 ms (Table 2). The increase in stride time for the loaded limb was limited to an increase in swing time, whereas the increase in stride time for the unloaded limb was due to a longer stance time. These data illustrate the dynamically coupled nature of the lower extremities during walking. That is, any change in the temporal pattern of one leg ultimately affects the temporal pattern of the other leg, with the underlying constraint that stride time must remain identical for both legs when walking in a straight line.

Neuromuscular changes, as reflected by net joint moments, were more pronounced at the knee than the hip and were reflected by increased peak flexor and extensor moments during swing for the loaded limb. It was not surprising that the response at the knee was clearer than the response at the hip since hip moments are typically more variable than knee and ankle moments (Winter, 1987). In addition, additional mass used in this study increased the moment of inertia about a transverse axis through the knee by approximately 76%, whereas the moment of inertia about a transverse axis through the hip joint increased by a more modest 44%. Thus, it is possible that the greater sensitivity of the knee moment dependent variables is due in part to the larger relative increase in moment of inertia at the knee.

In general, increased joint moment magnitudes for the loaded limb during swing were consistent with greater inertia of this limb. Neuromuscular changes were not complete immediately following initial exposure to the load. Instead several joint moment variables exhibited additional changes up to 5 min after load application. Beyond this point all joint moment variables did not change until the load was removed. These results are consistent with previous findings. For example, when participants were asked to walk on a split-belt treadmill and velocity of one of the belts was slowed relative to the other, participants adjusted lower extremity muscle excitations over approximately 20 strides to account for the differences in belt speeds (Dietz, Zijlstra, & Duysens, 1994; Prokop, Berger, Zijlstra, & Dietz, 1995). Participants walking with an exoskeleton strapped to their lower extremities, which provided a velocity dependent resistance during stepping, adjusted lower extremity muscle excitations over about 10 strides to accommodate the additional load imposed by the exoskeleton (Lam, Anderschitz, & Dietz, 2006). Despite differences between our perturbation and those of split-belt walking and walking with an exoskeleton strapped to the legs, it appears that adjustment to lower extremity loading occurs over the course of several strides. Thus, the neuromuscular system may need time, albeit brief, to adapt to altered inertial properties of both the lower extremities.

It is worth noting that an apparent after-effect was observed in the peak hip extensor moment near swing termination following load removal. On an individual participant basis, we found that three out of our six participants showed a clear after-effect for the peak hip extensor moment following load removal and two other participants exhibited a slight after-effect for this variable. Previous researchers (Lackner & Dizio, 1994; Lam et al., 2006; Sainburg, Ghez, & Kalakanis, 1999; Shadmehr & Mussa-Ivaldi, 1994) investigating load manipulations have used the existence of an after-effect following load removal to suggest that the internal model of the task itself was altered when adjusting to the new load. Thus, the presence of the after-effect in the hip moment following load removal in our study potentially suggests that the process of accommodating to the altered lower extremity inertia during walking was not purely a mechanical response, but also involved altering the internal model. It should be noted, however, that evidence of an after-effect was only seen for this one variable so caution should be used when interpreting this result. To better understand whether neural adaptation occurs as the result of altering lower extremity inertia, electromyography of the lower extremity muscles should also be investigated in conjunction with inverse dynamics to provide a more direct interpretation of muscle responses following load manipulation.

While our results indicate at first glance that temporal data adapted more abruptly than net joint moment data there are several methodological limitations to our study that could have contributed to this outcome. First, this outcome may be due in part to our experimental design, particularly the timing of overground walking, the source of our net joint moment analyses, and treadmill walking, the source of our temporal analyses. Since three overground walking trials were collected immediately following application of the load to one ankle and prior to any treadmill walking, joint moment data for 1.L.00 actually reflect less exposure to the inertia change than temporal data for 1.L.00. When temporal data were collected for 1.L.00, participants would have already completed three overground walking (~50–60 strides), traversed the distance between the overground walkway and treadmill area (~10–15 strides), and walked (~20–30 strides) on the treadmill while the treadmill belt speed increased to the experimental speed. Had we been able to capture both temporal and kinetic data continuously following a unilateral change in leg inertia, accommodation profiles for temporal and joint moment data may have been more similar. Secondly, our joint moment data were averaged over three strides compared to 50–55 strides for our temporal data suggesting that our temporal data were influenced less by a spurious stride. Given the above limitations to this study, however, the results of this study clearly show that no further changes occurred in temporal or joint moment data after 5 min of walking under altered inertia conditions. Future studies in this area should focus on the first 5 min of accommodation in an effort to gain further insights into potential neural adaptations to the additional lower extremity loads. This assessment would probably be accomplished more effectively during treadmill walking. Although it may be difficult to analyze the stance phase of walking without an instrumented treadmill capable of measuring both vertical and shear forces, swing phase analyses which do not require ground reaction force information would be valuable since many of the changes due to the load are reflected in swing phase dynamics.

In conclusion, walking symmetry was perturbed immediately following initial exposure to asymmetrical loading. Temporal data reflected an immediate and stable adaptation to the inertial change, both when load was added to the limb and when it was removed. Both hip and knee moments exhibited further changes beyond initial exposure to the load, but these changes were complete within 5 min. Whether temporal data reflect an immediate adaptation to the load or the observed response is related to the timing at which the temporal data were assessed remains unclear. From a more practical perspective, our results clearly suggest that individuals should be given at least 5 min of walking or other normal activity to adapt to altered inertial conditions before gait assessment is conducted. The neuromuscular system, as reflected in our project by joint moment profiles, appears to adapt over several minutes of exposure to altered inertial conditions. Our recommendation of 5 min is consistent with the shortest accommodation times reported in the literature, suggesting that data from these studies (Czerniecki et al., 1994; Gitter et al., 1997; Hillery et al., 1997; Huang et al., 2000; Mattes et al., 2000; Skinner & Barrack, 1990) most likely reflect a fully accommodated response.

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