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Compensatory mechanisms in below-knee amputee gait in response to increasing steady-state walking speeds

Anne K. Silverman [a], Nicholas P. Fey [a], Albert Portillo [b], Judith G. Walden [b], Gordon Bosker [c], Richard R. Neptune [a,*]

[a] Department of Mechanical Engineering, The University of Texas, Austin, TX, USA
[b] Physical Medicine and Rehabilitation Service, Audie L. Murphy VA Hospital, San Antonio, TX, USA
[c] Department of Rehabilitation Medicine, The University of Texas Health Science Center, San Antonio, TX, USA

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Abstract

Compensatory mechanisms in below-knee amputee gait are necessary due to the functional loss of the ankle muscles, especially at higher walking speeds when the mechanical energetic demands of walking are greater. The objective of this study was to examine amputee anterior/posterior (A/P) ground reaction force (GRF) impulses and joint kinetics across a wide range of steady-state walking speeds to further understand the compensatory mechanisms used by below-knee amputees. We hypothesized that amputees would rely more on their intact leg to generate greater propulsion relative to the residual leg, which would result in greater GRF asymmetry between legs as walking speed increased. Amputee and control subject kinematic and kinetic data were collected during overground walking at four different speeds. Group ($n = 14$) average amputee data showed no significant differences in braking or propulsive GRF impulse ratios, except the propulsive ratio at 0.9 m/s, indicating that the subjects maintained their initial levels of GRF asymmetry when walking faster. Therefore, our hypothesis was not supported (i.e., walking faster does not increase GRF loading asymmetry). The primary compensatory mechanism was greater positive residual leg hip joint power and work in early stance, which led to increased propulsion from the residual leg as walking speed increased. In addition, amputees had reduced residual leg positive knee work in early stance, suggesting increased output from the biarticular hamstrings. Thus, increasing residual leg hip extensor strength and output may be a useful mechanism to reduce GRF loading asymmetry between the intact and residual legs.

Keywords: Ground reaction force impulse; Joint kinetics; Propulsion; Transtibial amputee

1. Introduction

Below-knee amputation was the second most common type of amputation of US adults between 1988 and 1996, due primarily to traumatic injuries and vascular disease [1]. In addition to developing chronic pain in their residual leg [2], amputees have a higher risk of developing musculoskeletal disorders in their intact leg due to increased asymmetry in the loading and stance time of their intact leg [3–5]. Much of this asymmetry is due to the functional loss of the ankle plantar flexors, which have been shown to be critical in providing body support, forward propulsion and leg swing initiation during normal walking [6–9]. Thus, significant compensatory mechanisms are necessary to fulfill the role of the lost ankle muscles. Previous studies examining ground reaction force (GRF) (e.g., Refs. [10–13]), joint kinetic (e.g., Refs. [12,14–18]), and electromyographic (EMG) (e.g., Refs. [14,18–21]) data have provided much insight into the compensatory mechanisms used by below-knee amputees. However, it is not clear if these compensatory mechanisms remain invariant with changes in task demands, such as walking over a wide range of walking speeds.

* Corresponding author at: Department of Mechanical Engineering, The University of Texas at Austin, 1 University Station C2200, Austin, TX 78712, USA. Tel.: +1 512 471 0848; fax: +1 512 471 8727.
E-mail address: rneptune@mail.utexas.edu (R.R. Neptune).

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Previous studies have shown that the intact leg generates greater GRFs relative to the residual leg [10,12,13] and non-amputees [11] while walking at moderate speeds. However, few studies have analyzed below-knee amputee kinetics and GRFs at different walking speeds. The magnitudes of both the residual and intact leg vertical GRF peaks have been shown to increase with speed, although the residual leg peak vertical GRF increased to a lesser extent than the intact leg [13]. Similarly, across two speeds (1.2 and 1.6 m/s), the intact propulsive GRF peak was shown to increase to a greater extent than the residual leg propulsive GRF peak [12]. Since normal prosthetic foot-ankle components are passive devices, they are limited in their ability to provide increased propulsion as task demands change. Therefore, residual leg propulsion deficits may be exacerbated when the demand for forward propulsion increases at higher walking speeds, and thus require the intact leg to provide a greater portion of the necessary propulsion.

The objective of this study was to further understand the compensatory mechanisms used by below-knee amputees by examining the anterior/posterior (A/P) GRF impulses and joint kinetics across a wide range of steady-state walking speeds. In particular, the braking and propulsive impulses (i.e., the time integral of the negative and positive A/P GRFs, respectively) are measures that incorporate both GRF magnitude and duration and will provide important insight into how amputees modulate propulsion in the absence of the ankle plantar flexors. Specifically, we tested the hypothesis that the intact leg will provide a greater portion of the necessary propulsion with increasing speed, and therefore asymmetry between the residual and intact leg impulses will also increase. Joint work was determined and compared to non-amputee control subjects to identify the compensatory strategies used to modulate propulsion with increasing speed.

2. Methods

2.1. Subjects

Fourteen unilateral transtibial amputee subjects (13 males, one female; 11 traumatic, three vascular; 45 ± 9 years) and 10 non-amputee, control subjects (seven males, three females; 33 ± 12 years) participated in the study. All subjects were free from musculoskeletal disorders and leg pain, did not require assistive devices, and were proficient walkers. The subjects provided informed consent approved by the University of Texas and the South Texas VA Medical Center prior to the study. Each amputee subject used his or her own prosthesis. A certified and licensed prosthetist with over 30 years of experience verified the alignment and fit of each prosthesis prior to testing.

2.2. Procedures

An eight-camera motion capture system (Vicon, Oxford Metrics) was used for three-dimensional motion analysis. Reflective markers (14-mm diameter) were placed bilaterally on the first, second, and fifth metatarsal heads, dorsal foot, heel, lateral and medial malleoli, lateral and medial femoral condyles, greater trochanter, anterior superior iliac spine, posterior superior iliac spine, iliac crest, shoulder, and C-7 vertebrae. Clusters of four markers each were also placed bilaterally on the shank and thigh. On the amputee residual leg, the first, second and fifth metatarsal head, dorsal foot, heel, and lateral and medial malleoli were placed on the prosthesis such that the markers were symmetric with the intact leg. Ground reaction forces were measured using four force plates (Advanced Mechanical Technology, Inc.) imbedded in a 10-m walkway. The force plates were concealed from the subjects with floor tiles to prevent foot-placement targeting.

Kinematic and GRF data were collected at 120 and 1200 Hz, respectively. Subjects walked steadily along the 10-m walkway at four randomly ordered speeds: 0.6, 0.9, 1.2 and 1.5 m/s. The average speed for each trial was determined using two infrared timing gates. Trials were repeated until at least five force-plate hits per foot were collected at each speed $\pm$ 0.06 m/s.

2.3. Data analysis

The experimental data were processed using Visual 3D (C-Motion, Inc.). Kinematic data were low-pass filtered using a 4th-order Butterworth filter with a cut-off frequency of 6 Hz. Kinetic data were similarly low-pass filtered at 20 Hz and normalized to the body weight of each subject. Inertial properties of the residual leg were based on Mattes et al. [22]. Propulsive and braking impulses were computed as the time integral of the positive and negative A/P GRF, respectively. To assess the degree of GRF asymmetry between legs, propulsive and braking impulse ratios were computed. In the amputees, the ratios were calculated as the residual leg impulse divided by the intact leg impulse. In the control subjects, the ratios were calculated as the left leg impulse divided by the right leg impulse. Changes in the joint kinetics were assessed by calculating the joint work (i.e., integrating the corresponding positive and negative joint power) during the stance phase (ipsilateral heel strike to toe off) in both legs. The joint power was calculated as the product of the joint moment and angular velocity. All GRF and kinetic quantities were averaged across trials for each subject at each speed, and then averaged across subjects within the amputee and control groups.

2.4. Statistical analysis

Statistical analysis was performed using SPSS 14.0 (SPSS, Inc.). Two-factor (leg and speed), repeated-measures ANOVAs were used to compare the residual, intact and control leg GRF impulses and joint work across speeds. The control leg values used in statistical comparisons were the average of the left and right legs. The ‘leg’ factor had two levels, and all three combinations of legs were compared (residual with intact, residual with control, and intact with control leg). The ‘speed’ factor had four levels, corresponding to the four walking speeds. The ANOVAs comparing the residual and intact legs had ‘within-subjects’ comparisons across all levels (leg and speed), whereas the ANOVAs comparing the control values to either the residual or intact legs had ‘within-subjects’ comparisons across speed but ‘between-subjects’ comparisons across legs. When significant differences were found, pairwise comparisons with a Bonferroni adjustment for multiple comparisons were used to determine which conditions were significantly different from each other. For the propulsive and braking impulse ratios, one two-factor (group and speed), repeated-measures ANOVA was used to identify significant differences across walking speeds and between the amputee and control subject groups. The two-factor ANOVA had four 'within-subjects' speed levels, and two 'between-subjects' group levels (amputee and control). When significant differences were found, pairwise comparisons with a Bonferroni adjustment for multiple comparisons were again used to determine which conditions were significantly different. A significance level of 0.05 was used for all statistical comparisons.

3. Results

On average, 41.0 ± 16.5% of the walking trials were not used due to incomplete force-plate hits or not meeting the walking speed requirements.

3.1. Propulsive impulses

There was no significant difference between the left and right leg propulsive impulses in the control subjects as walking speed increased, and there was no difference between the intact leg and control propulsive impulses (Fig. 1a). The residual, intact and control propulsive impulses all had significant speed effects ($p < 0.001$). There was significantly less propulsion from the residual leg at 0.6 m/s ($p \leq 0.020$) compared to all other speeds as well as a significant increase in propulsion with each increase in speed for the intact and control legs ($p \leq 0.001$). The propulsive impulse had a significant leg effect ($p < 0.001$) and leg × speed interaction effect ($p \leq 0.044$), with the residual leg propulsive impulse significantly lower than both the intact and control legs at all speeds ($p < 0.001$, Fig. 1a).

The propulsive impulse ratio had a significant group effect ($p < 0.001$) and group × speed interaction effect ($p = 0.021$). The amputee propulsive impulse ratio was significantly less than the control propulsive impulse ratio at all speeds ($p \leq 0.001$). There was no significant difference in control propulsive impulse ratio (left/right) across speeds (Fig. 1b). The amputee propulsive impulse ratio (residual/intact) at 0.9 m/s was significantly greater than the ratio at 1.5 m/s ($p = 0.036$). However, in general, the amputee propulsive impulse ratio did not significantly increase or decrease with walking speed (Fig. 1b). Thus, the hypothesis that the intact leg would provide a greater portion of the necessary propulsion as walking speed increased (i.e., the propulsive impulse ratio would decrease) was not supported.

3.2. Braking impulses

There was no significant difference between the control left versus right, control versus intact, or residual versus intact leg braking impulse at any speed, although there was significant leg × speed interaction effect between the intact and control legs ($p = 0.002$). The residual, intact and control legs all showed a significant speed effect ($p < 0.001$). The residual and intact legs significantly increased braking up to

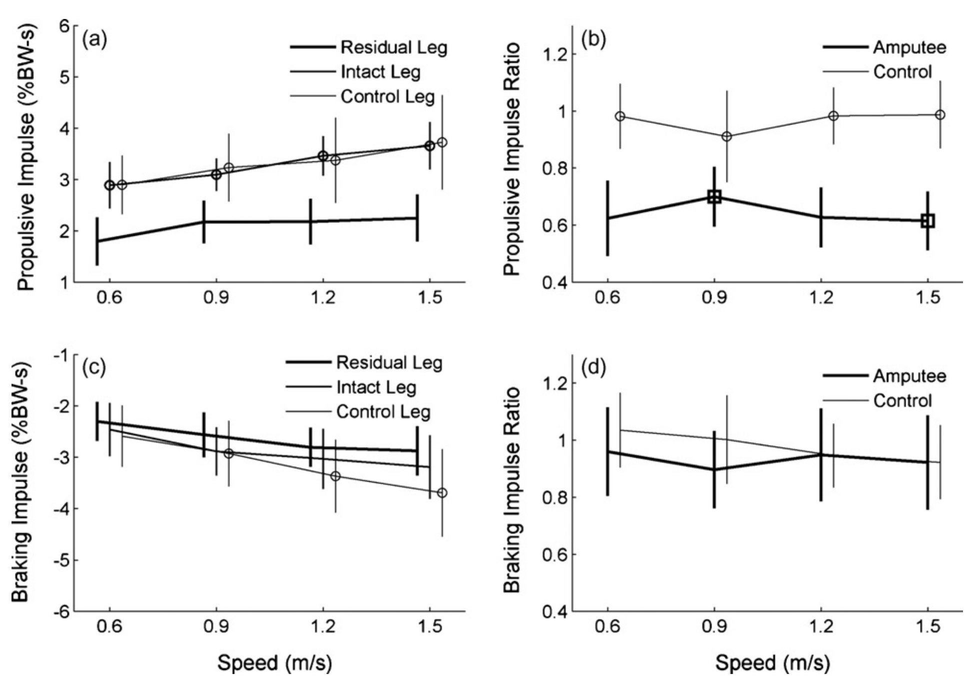

Fig. 1. Braking and propulsive impulses. (a) Propulsive impulses for the residual, intact and control legs. Significant differences with the residual leg are indicated with (○). (b) Propulsive impulse ratios for both amputee (residual/intact) and control (left/right) subjects. Significant differences between the amputee and control groups are indicated with (○). In addition, the significant difference in the amputee propulsive impulse ratio between 0.9 and 1.5 m/s is indicated with (□). (c) Braking impulses for the residual, intact and control legs. Significant differences with the residual leg are indicated with (○) and (d) braking impulse ratios for both amputee (residual/intact) and control (left/right) subjects. Vertical lines in plots indicate the standard deviations across subjects.

1.2 m/s ($p \leq 0.011$), but did not show a significant increase between 1.2 and 1.5 m/s. The control leg significantly increased braking at all speeds ($p \leq 0.004$). The residual and control leg braking impulses had a significant leg effect ($p = 0.006$) and leg $\times$ speed interaction effect ($p < 0.001$), such that the residual leg braking impulse was significantly lower than the control braking impulse at 0.9 m/s ($p = 0.050$), 1.2 m/s ($p = 0.003$) and 1.5 m/s ($p = 0.001$) (Fig. 1c).

The braking impulse ratio did not show a significant change across group or speed (Fig. 1d). Neither the amputee nor control subjects exhibited a significant change in braking impulse ratio across speeds (Fig. 1d).

3.3. Joint kinetics

Overall, both the amputee and control subjects showed systematic increases in joint power with increasing speed (Fig. 2), with several notable differences between the residual, intact and control legs at each joint. At the hip, positive work during stance had a significant speed effect ($p \leq 0.001$) for the control, intact and residual legs and all legs significantly increased with each increase in speed ($p \leq 0.002$). The residual and control legs had a significant leg effect ($p = 0.014$) and leg $\times$ speed interaction effect ($p = 0.007$), such that the residual leg was significantly greater than the control subjects at 0.9, 1.2 and 1.5 m/s ($p \leq 0.016$, Fig. 3). The intact and control legs also had a significant leg effect ($p = 0.018$), with the intact leg positive hip work significantly greater than the control subjects at the higher speeds of 1.2 and 1.5 m/s ($p \leq 0.040$, Fig. 3). Negative work at both the hip and knee had significant speed effects ($p < 0.001$), with all legs significantly increasing with each speed ($p \leq 0.010$), but there were no significant differences between the control, intact and residual legs (Fig. 3).

The control positive knee work showed a significant speed effect ($p < 0.001$), but was significantly greater only at 1.5 m/s ($p \leq 0.006$). The residual leg positive knee work also showed a significant speed effect ($p < 0.001$), and increased between the first two speeds and between the last two speeds ($p \leq 0.049$). The intact leg positive knee work had a significant speed effect ($p < 0.001$), which increased at all speeds ($p \leq 0.001$) except between 0.6 and 0.9 m/s. Positive residual leg knee work showed a significant leg effect ($p \leq 0.033$) and a significant leg $\times$ speed interaction effect ($p \leq 0.013$), such that the residual leg exhibited significantly lower positive knee work relative to the intact leg at 0.6, 1.2 and 1.5 m/s ($p \leq 0.001$, Fig. 3) and control subjects at 1.2 and 1.5 m/s ($p \leq 0.003$, Fig. 3). The control and intact legs also had a significant leg effect ($p = 0.033$) and leg $\times$ speed interaction effect ($p = 0.012$), with the intact leg having significantly more positive knee work at 1.2 and 1.5 m/s ($p \leq 0.018$).

The residual, intact and control legs also showed significant speed effects in positive ankle work ($p < 0.001$),

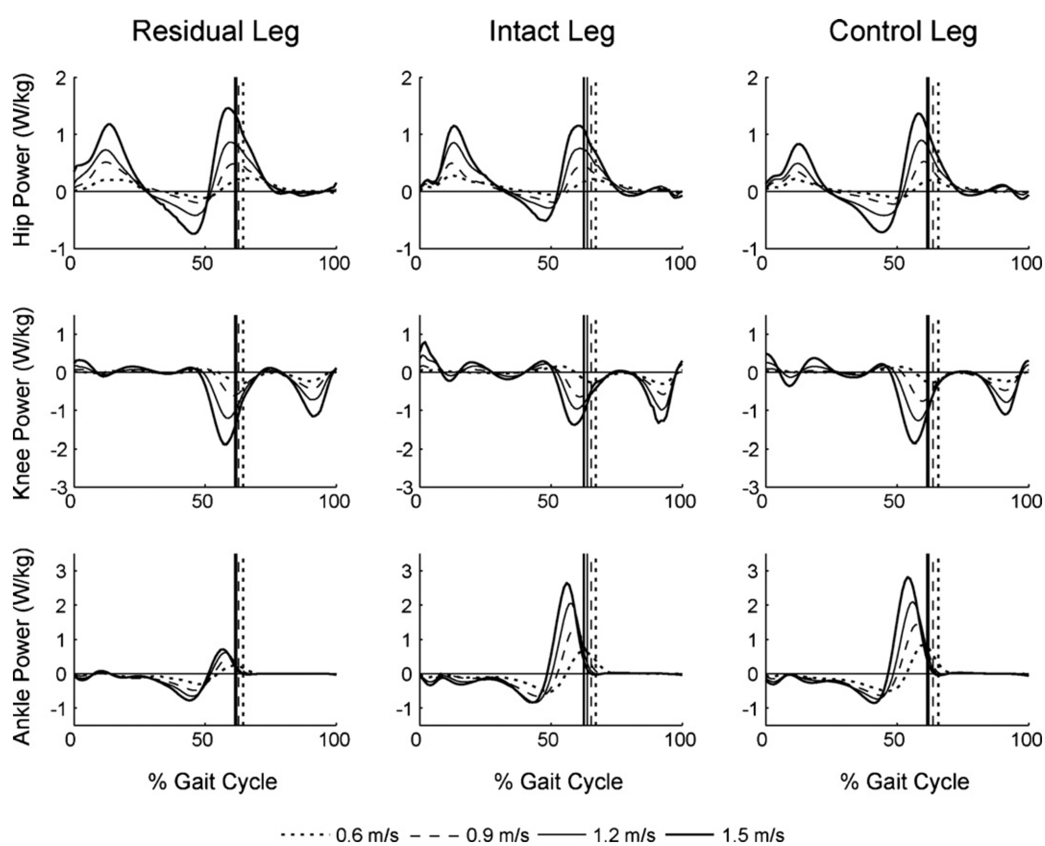

Fig. 2. Average residual, intact and control leg net joint powers at the hip, knee and ankle normalized to the gait cycle. Vertical lines indicate toe-off.

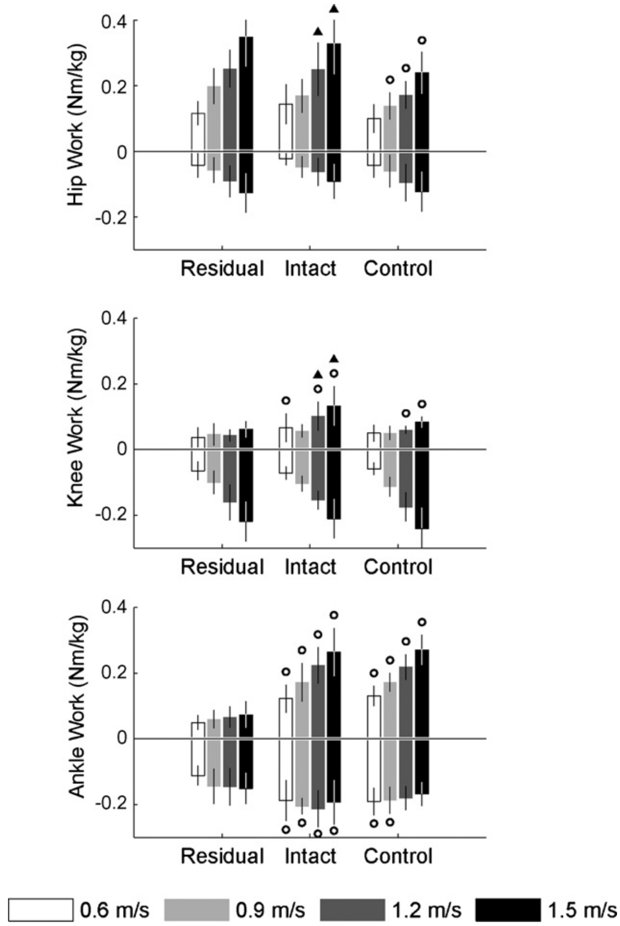

Fig. 3. Average residual, intact and control leg positive and negative joint work at the hip, knee and ankle during stance. Significant differences with the residual leg are indicated with (○). Significant differences between the intact and control legs are indicated with (▲). Vertical lines indicate the standard deviations across subjects.

although the residual leg did not increase with speed as much as the intact and control legs (residual $p \leq 0.028$, intact and control $p < 0.001$). The residual leg positive ankle work also had a significant leg effect ($p < 0.001$) and a significant leg × speed interaction effect ($p < 0.001$), with the positive residual leg work being significantly lower than both the intact and control legs at all speeds ($p < 0.001$, Fig. 3). The residual leg negative ankle work had a significant leg effect ($p = 0.009$) and leg × speed interaction effect with the control leg ($p < 0.001$), and was significantly lower than the control negative ankle work at the first two speeds ($p \leq 0.026$). The residual leg negative ankle work also had a significant leg effect with the intact leg ($p < 0.001$), and was significantly less than the intact leg negative ankle work at all speeds ($p \leq 0.001$).

4. Discussion

The objective of this study was to investigate compensatory mechanisms used by amputees as they walked at increasing steady-state walking speeds by examining changes in the A/P GRF impulses and joint kinetics. While the intact leg consistently generated significantly more propulsion than the residual leg, the ratio between the two legs remained constant with increasing speed. Therefore, the hypothesis that the intact leg would provide a greater portion of the necessary propulsion as walking speed increased was not supported. Similarly, the amputee braking impulse ratio did not change with speed, although the residual leg did exhibit significantly less braking than control subjects at all speeds except 0.6 m/s. These results were consistent with previous studies that have observed significantly lower GRFs on the residual leg relative to the intact leg [10,12,13] and control subjects [12,13]. Previous studies examining amputee GRFs at different walking speeds have found the intact leg peak propulsive [12] and vertical [12,13] GRFs increase to a greater extent than the residual leg. These studies appear to conflict with our results. However, consistent with these studies [12,13], we found a significant leg by speed effect with the residual and intact propulsive impulses, with the intact leg increasing at a higher rate than the residual leg at the higher speeds (Fig. 1a). Yet, we found that the impulse ratio did not significantly change, as the portion of each leg's contribution to the total GRF impulse remained invariant with walking speed. Thus, our results are consistent with these previous studies [12,13].

The primary compensatory mechanism was increased positive residual leg hip joint work, especially in early stance (Figs. 2 and 3). Increased residual leg hip extensor moments and powers in early stance have been observed in previous studies of amputee walking [18,23,24] and are consistent with the gluteus maximus (GMAX) and the biarticular hamstrings providing propulsion in early stance in able-bodied walking [6,8]. Increased residual leg GMAX activity has been observed in amputee walking [18,25] and a modeling and simulation analysis of symmetric amputee walking showed that the residual leg GMAX increased its excitation to deliver more energy to the trunk to provide forward propulsion [26]. Increased and prolonged residual leg biarticular hamstring activity has also been observed in previous studies [14,18–21,25,27] and would not only provide increased propulsion [6,8], but would correspond with the reduced residual leg positive knee work shown in early stance (Figs. 2 and 3; see also Ref. [12]). Increased co-contraction of the biarticular hamstrings with the knee extensors acts to reduce the net knee extensor moment and corresponding positive knee work. This highlights an important potential limitation of our study in that we analyzed net joint moment-based quantities, which are sensitive to the levels of co-contraction. Thus, changes in these quantities should be interpreted accordingly and future studies should be performed at the individual muscle level.

The positive work at the intact hip joint was also significantly greater than the controls at the two highest walking speeds (Fig. 3). This is especially evident in early intact leg stance (Fig. 2), which corresponds to the residual leg propulsion phase. A greater intact leg hip extensor moment in early stance has been previously observed [23] and may provide additional propulsion. A simulation analysis of symmetric amputee walking showed the intact leg GMAX transfers more energy from the leg to the trunk to provide forward propulsion compared to non-amputee walking [26]. Future modeling and simulation work should assess whether this mechanism similarly occurs in asymmetric amputee walking.

Analysis of the data provided much insight into how amputees adapt to faster walking speeds. However, some individual subjects exhibited strategies not captured in the group average analysis. For example, Subject A generated similar kinetic patterns as the group average and had a relatively constant propulsive impulse ratio with increasing walking speed (Fig. 4b). The intact leg generated only slightly greater propulsion than the residual leg, which resulted in an impulse ratio that was the highest of all amputee subjects, and most similar to the control subjects (Fig. 4a and b). However, contrary to the group average and previous studies (e.g., Ref. [28]), Subject A had an increased residual leg stance time relative to the intact leg. The prolonged stance time along with a systematic increase in GRF magnitude resulted in a residual leg propulsive impulse that was similar in magnitude to the intact leg.

In contrast, Subject B produced a propulsive impulse ratio that decreased with increased walking speed (Fig. 5b), which was due to the intact leg propulsive impulse increasing with speed while the residual leg propulsive impulse decreased beyond 0.9 m/s (Fig. 5a). This subject was consistent with the hypothesis that the propulsive impulse ratio would decrease with speed due to increased dependence on the intact leg for propulsion. Increased dependence on the intact leg occurred primarily at the ankle, with the ankle power and work increasing with speed, which was consistent with previous studies [15,23]. In addition, this subject displayed increased output from the residual leg hip flexors in late stance. In non-amputee simulations, the hip flexors act to accelerate the leg into swing, but also decelerate the trunk forward such that its net effect on the A/P GRF is negative [8]. Therefore, the increased residual leg hip flexor output may be helping to accelerate the residual leg into swing in the absence of the gastrocnemius [7–9,26], but would appear to further reduce residual leg propulsion.

These two subjects used different compensatory mechanisms, even though both were proficient walkers. An interesting note is that Subject A used a solid-ankle, cushioned-heel (SACH) foot, while Subject B used an energy storage and return foot (ESAR). While ESAR feet do not provide active propulsion, they have been shown to improve residual leg propulsion [29]. Thus, it was surprising that Subject A, with the SACH foot, had better GRF symmetry than Subject B, using an ESAR foot, which highlights the effectiveness of neuromotor compensations alone in achieving symmetric gait patterns. In addition, Subject A was a traumatic amputee, while Subject B was a vascular amputee. In general, vascular amputees have decreased self-selected walking speed and stride length compared to traumatic amputees [29], and reduced propulsive GRFs relative to both traumatic amputees and control subjects [30]. To assess whether the mixed

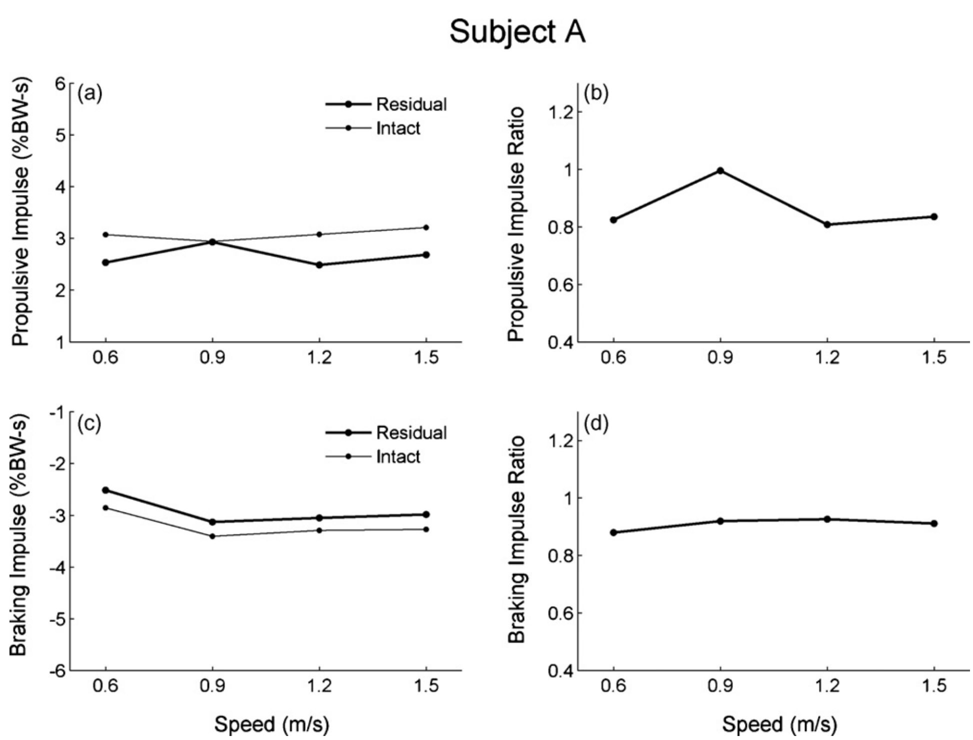

Fig. 4. Subject A (a) propulsive impulses for the residual and intact legs, (b) propulsive impulse ratio (residual/intact), (c) braking impulses for the residual and intact legs, and (d) braking impulse ratio (residual/intact).

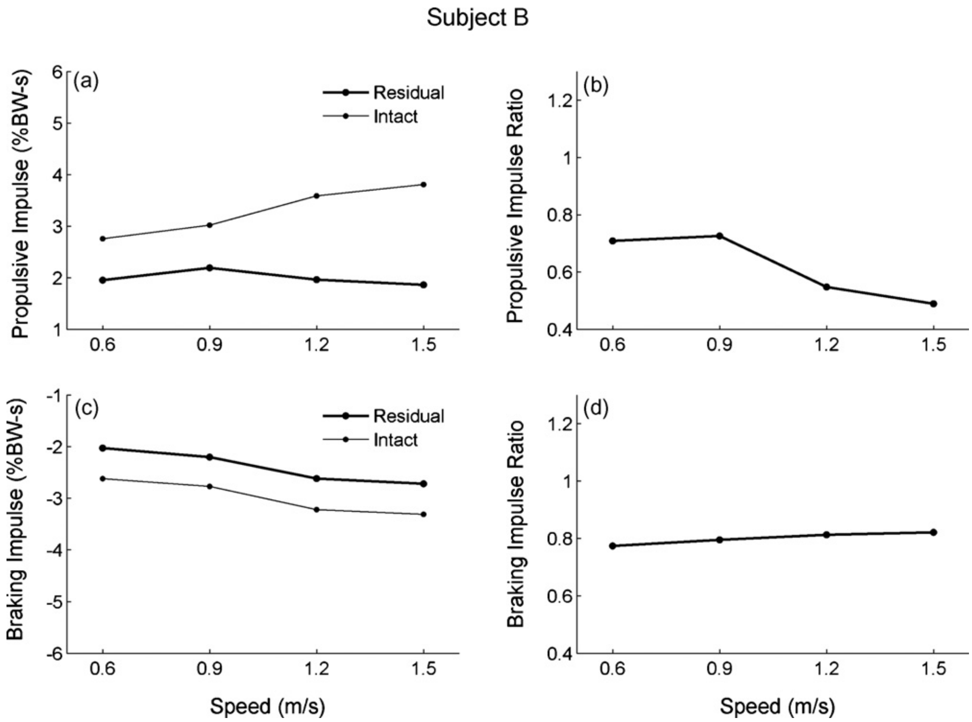

Fig. 5. Subject B (a) propulsive impulses for the residual and intact legs, (b) propulsive impulse ratio (residual/intact), (c) braking impulses for the residual and intact legs, and (d) braking impulse ratio (residual/intact).

etiology influenced our results, we performed the same statistical analysis with only the traumatic subjects ($n = 11$). Although there were some slight differences in the statistical results, the overall conclusions remained the same. Similarly, a post hoc analysis was performed to assess the effect of foot type on the results. Only subjects with an ESAR prosthesis ($n = 9$) were included in the analysis. Again, the results largely matched the overall analysis with only slight differences in the statistical results. Thus, the etiology of the amputation and prosthetic foot type did not influence the conclusions of the study and including all the subjects only strengthened the statistical results.

These results have important implications for improving GRF loading symmetry for lower-limb amputees. Amputees have an increased risk of developing joint disorders on the intact leg due to increased dependence on the intact leg [3–5]. In the present study, the amputees did not display greater GRF asymmetry as walking speed increased, suggesting that the residual leg can effectively compensate for the lost ankle muscles, and that greater dependence on the intact leg may not be necessary. In addition, it appears loading asymmetry is not likely a reason amputees have a slower self-selected walking speed compared to control subjects, as asymmetry was not influenced by walking speed. The primary compensatory mechanism was increased output from the residual leg hip extensors in early stance, which appears sufficient to prevent the GRF asymmetry from increasing. Thus, increasing residual leg hip extensor strength and output may be a useful mechanism to reduce reliance on the intact leg and delay the onset of joint disorders. Future work using modeling and simulation techniques should analyze different strategies used by individual subjects to see if one strategy produces lower joint loads than another.

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Conflict of interest

None.

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