Published in final edited form as:

J Biomech. 2016 June 14; 49(9): 1757–1764. doi:10.1016/j.jbiomech.2016.04.004.

Biomechanics of the human walk-to-run gait transition in persons with unilateral, transtibial, amputation

Tracy N. Giest and Young-Hui Chang*

School of Applied Physiology, Georgia Institute of Technology, Atlanta, GA, USA

Abstract

Propulsive force production (indicative of intrinsic force-length-velocity characteristics of the plantar flexor muscles) has been shown to be a major determinant of the human walk-to-run transition. The purpose of this work was to determine the gait transition speed of persons with unilateral, transtibial amputation donning a passive-elastic prosthesis and assess whether a mechanical limit of their intact side plantar flexor muscles is a major determinant of their walk-to-run transition. We determined each individual's gait transition speed (GTS) via an incremental protocol and assessed kinetics and kinematics during walking at speeds 50, 60, 70, 80, 90, 100, 120, and 130% of that gait transition speed (100%:GTS). Unilateral, transtibial amputees transitioned between gaits at significantly slower absolute speeds than matched able-bodied controls (1.73 ± 0.13 and 2.09 ± 0.05 m/s respectively, $p < 0.01$). Peak anterior-posterior propulsive force increased with speed in controls until 100% of the preferred gait transition speed and decreased at greater speeds. A significant decrease in anterior-posterior propulsive force production was found at 120%GTS (110%: 0.27 ± 0.04 > 120%: 0.23 ± 0.05 BW, $p < 0.05$). In contrast, amputee subjects' intact side generated significantly higher peak anterior-posterior propulsive forces while walking at speeds above their preferred gait transition speed (100%: 0.28 ± 0.04 < 110%: 0.30 ± 0.04 BW, $p < 0.05$). Changes in propulsive force production were found to be a function of changes in absolute speed, rather than relative to the walk-to-run transition speed. Therefore, the walk-to-run transition in unilateral, transtibial amputees is not likely dictated by propulsive force production or the force-length-velocity characteristics of the intact side plantar flexor muscles.

Keywords

Amputee; Below-knee; Propulsive force; High speed; Locomotion

*Corresponding Author Contact Information: Young-Hui Chang, Comparative Neuromechanics Laboratory, School of Applied Physiology, Georgia Institute of Technology, 555 14th St NW, Atlanta, GA 30332-0356, 404-894-9993 (phone), 404-894-9982 (fax), ; Email: yh.chang@ap.gatech.edu, <http://www.ap.gatech.edu/ChangLab/CNLhome.html>

Conflict of Interest Statement:

The authors do not have any conflicts of interest to report.

Publisher's Disclaimer: This is a PDF file of an unedited manuscript that has been accepted for publication. As a service to our customers we are providing this early version of the manuscript. The manuscript will undergo copyediting, typesetting, and review of the resulting proof before it is published in its final citable form. Please note that during the production process errors may be discovered which could affect the content, and all legal disclaimers that apply to the journal pertain.

INTRODUCTION

The walk-to-run gait transition has long been studied in humans, as a means to better understand the biomechanics and neural control of locomotion. Many key “determinants,” or factors that appear to drive the walk-to-run transition have been identified over the years, including: metabolic energy minimization (Minetti et al., 1994), kinetic factors (Raynor et al., 2002), critical flexor muscle activations during the swing phase of gait (Hreljac et al., 2001; Malcolm et al., 2009b; Prilutsky and Gregor, 2001; Segers et al., 2007), a critical angular velocity of the ankle to protect dorsiflexors (Hreljac, 1995), and pendular dynamics of the center of mass (Kram et al., 1997). Some have proposed the idea that a combination of these factors contributes to the walk-to-run transition (Bartlett and Kram, 2008; Malcolm et al., 2009a). More recently, work has pointed to plantar flexor force production as a major determinant of the walk-to-run transition (Arnold et al., 2013; Farris and Sawicki, 2012; Malcolm et al., 2009a; Neptune and Sasaki, 2005). Modeling of the soleus, medial gastrocnemius, and lateral gastrocnemius, and in vivo measurement of the medial gastrocnemius have demonstrated that as walking speed increases, so do the plantar flexor fascicle shortening velocities. These increases in plantar flexor fascicle shortening velocities result in a decrease in muscle force production, which then lead to decreases in propulsive force production by the legs. Furthermore, transitioning from a walk to a run at the preferred walk-to-run transition speed of ~2.0m/s greatly reduces fascicle shortening velocities in the plantar flexor muscles, and in turn increases propulsive force production (Arnold et al., 2013; Farris and Sawicki, 2012; Neptune and Sasaki, 2005). During walking, the plantar flexor muscles are the primary contributors to propulsive force production (Anderson and Pandey, 2003; Neptune et al., 2004; Silverman and Neptune, 2012; Zajac et al., 2002, 2003). Due to the plantar flexor’s intrinsic force-length-velocity characteristics, maximum propulsive force production of the anterior-posterior ground reaction forces during walking increases until the walk-to-run transition where it peaks, and then decreases during walking at speeds above transition speed (Neptune and Sasaki, 2005). As such, we sought to investigate the gait transition speed in persons with a unilateral, transtibial amputation using a passive-elastic prosthesis, to test whether intact plantar flexor propulsive force production is a major determinant of their walk-to-run transition.

The effective removal of plantar flexors in a unilateral, transtibial amputation in conjunction with the use of a passive-elastic prosthesis results in a diminished capacity to generate propulsive forces on the amputated side (Sanderson and Martin, 1997; Silverman et al., 2008). Amputated side peak anterior-posterior propulsive forces are consistently, significantly lower than able-bodied subjects and the intact side during walking at speeds ranging from 0.75–1.75m/s. (D’Andrea et al., 2014; Sanderson and Martin, 1997). Intact side peak anterior-posterior forces are not different from able-bodied subjects at speeds of 0.75, 1.50 or 1.75m/s (significantly higher than controls at 1.0 and 1.25m/s), however to our knowledge have never been investigated at walking speeds higher than 1.75m/s (D’Andrea et al., 2014). Furthermore, the walk-to-run gait transition has not previously been characterized in unilateral, transtibial amputees. Previous work has shown that amputee subjects using passive prostheses have slower preferred walking speeds than able-bodied controls (Gailey et al., 1994; Snyder et al., 1995; Waters et al., 1976; Winter and Sienko, 1988). Unilateral,

transtibial amputee subjects also have slower maximum sustainable walking speeds than control subjects, with ranges of ~1.5–1.6m/s and ~1.9–2.0m/s, respectively (Genin et al., 2008; Nolan et al., 2003). Consequently, otherwise healthy unilateral, transtibial amputee subjects provide an excellent experimental model to assess the effects of intact plantar flexor propulsive force production on the walk-to-run transition speed.

The purpose of this work was to determine the gait transition speed of unilateral, transtibial amputee subjects and assess if an intact side mechanical limit of the plantar flexor muscles (as indicated by peak propulsive force production) is a major determinant of the walk-to-run transition for amputee subjects. We hypothesized that unilateral, transtibial amputees would transition between gaits at a slower absolute speed than matched, able-bodied control subjects, as they have slower maximum sustainable walking speeds than control subjects. We further hypothesized that peak propulsive force production in amputee subjects' intact side would not increase at speeds higher than the gait transition speed (peak propulsive force production at 100% > 110–130%), thus supporting the idea that a mechanical limit of the plantar flexors (dictated by force-length-velocity characteristics) is a major determinant in amputee subjects' walk-to-run transitions. Alternatively, if amputees' peak propulsive force production continued to increase at speeds above the gait transition speed, then a mechanical limit of the plantar flexor muscles is most likely not a major determinant of the walk-to-run transition in unilateral, transtibial amputee subjects.

MATERIALS & METHODS

Subjects

All subjects gave informed written consent prior to completing the Georgia Institute of Technology IRB approved protocol. Subjects included 10 healthy, unilateral, transtibial amputee (AMP) subjects (5 males, 5 females; amputation: 3 elective due to congenital deformity, 7 traumatic; mean age $\pm$ SD: 26.7 $\pm$ 4.5 years; mass: 67.4 $\pm$ 14.6 kg; sound leg length: 91.5 $\pm$ 5.6 cm) and 10 healthy, able-bodied matched control (CON) subjects (5 males, 5 females; age: 29.6 $\pm$ 6.9 years; mass: 67.2 $\pm$ 10.0 kg; leg length: 91.1 $\pm$ 5.4cm). We took great care in matching each CON subject to one of the AMP subjects for all listed characteristics. There were no significant differences in age, weight, and leg length between the two groups; in addition, subjects were also qualitatively matched on activity level, training type, and gender (Table 1). All subjects participated in running for exercise or training at least once per week. Persons that were pregnant, diabetic, sedentary, or suffering from any cardiovascular or neurological pathologies, such as peripheral neuropathy, were excluded from the study. All CON subjects were free from musculoskeletal and neurological trauma. AMP subjects wore their own custom made, well fitting, prosthesis with a commercially available passive-elastic ankle-foot component. Passive-elastic ankle-foot components were chosen to maintain homogeneity of the prosthetic ankle function in amputee subject group. AMP persons using Solid-Ankle Cushion Heel (SACH) or powered ankle-foot devices were excluded from participating in the study, since SACH feet do not provide energy storage and return and powered ankle-foot devices provide active propulsive assistance (Herr and Grabowski, 2012; Torburn et al., 1990). To prevent fatigue, subjects completed testing over two days (no more than one week apart) and refrained from exercise on testing days.

Experimental Protocol

Day 1: Determine Gait Transition Speed—Subjects habituated to walking and running on a single belt of a custom dual-belt treadmill at self-selected speeds for three minutes with two-minutes rest. Determining each individuals' gait transition speed (GTS) consisted of two conditions: 1. incrementally increasing the treadmill speed; and, 2. incrementally decreasing treadmill speed, similar to what has been previously described in the literature (Hreljac, 1993a, b; Kram et al., 1997; Prilutsky and Gregor, 2001; Raynor et al., 2002). Once the treadmill belt reached the appropriate speed, subjects stepped onto the moving belt. Subjects were given up to 30 seconds to both walk and run before verbally indicating which gait they preferred. Subjects rested while the moving belt's speed was either increased or decreased by 0.1m/s, and then the task was repeated. During the increasing portion, CON subjects began at 1.3m/s and AMP at 1.0m/s. During the decreasing portion, CON subjects began at 2.6m/s and AMP at 2.3m/s. The lowest preferred running speeds and the highest preferred walking speeds were recorded as the transition speeds for the increasing and decreasing portions respectively. The starting speeds were different to accommodate AMP subjects' being uncomfortable attempting to walk above 2.3m/s while maintaining the range of speeds tested between the two groups. Subjects repeated the procedure a total of three times with the first used as practice to familiarize themselves with the task. The second and third times completing the procedure resulted in four recorded transition speeds. The presentation of increasing or decreasing speeds was randomized. Each individual's GTS was calculated by averaging the last four transition speeds together (two from incrementally increasing and two from incrementally decreasing treadmill speed).

Day 2: Instrumented Split-Belt Collection—Subjects habituated to walking and running on the custom dual-belt treadmill at self-selected speeds for three minutes with two-minutes rest. Subjects walked for 30-second trials at speeds 50, 60, 70, 80, 90, 100, 110, 120, and 130% of their previously determined GTS (with 100% corresponding to each individual's GTS). Previous work has investigated the walk-to-run transition at 15% (Prilutsky & Gregor, 2001) and 20% speed increments (Neptune & Sasaki, 2005). We chose 10% increments above and below the GTS to increase the resolution of our dataset. This allowed us to more precisely determine the relative speed where peak propulsive force production becomes limited without affecting our general conclusions. Presentation of speeds was randomized. Subjects rested seated for at least two minutes between each trial.

Kinetic & Kinematic Analysis

We placed retroreflective markers on each subject's right and left second metatarsophalangeal joint, lateral malleolus, heel, shank segment, lateral condyle, thigh segment, anterior superior iliac spine, and posterior superior iliac spine. For AMP subjects, amputated side (AMP-Amputated) markers on the prosthetic device were placed to match those on the intact side (AMP-Intact) (Powers et al., 1998; Silverman et al., 2008). Three-dimensional positions of the lower extremity marker data were collected by a six-camera Vicon motion analysis system (120Hz, Oxford, UK). A custom built, dual-belt treadmill instrumented with two commercial force plates (1080 Hz, Advanced Mechanical Technology Incorporated, Watertown, MA, USA) simultaneously measured ground reaction force data (Kram et al., 1998). We applied a low-pass, zero-phase lag, fourth order

Butterworth filter at 10 and 20 Hz to marker and force data respectively. We used a custom Matlab code to calculate mean sagittal plane kinematics. Joint angles for the hip, knee, and ankle measured during a static standing trial were defined as neutral. There were no significant differences in standing joint angles between the leg types (CON, AMP-Intact, and AMP-Amputated). Peak ankle plantarflexion was defined as the peak plantarflexion angle after 50% of the gait cycle. This was done to ensure the comparisons in plantarflexion angle between leg types were made during early swing. This distinction was made specifically because the AMP-Amputated side plantarflexion angle peaked for some subjects during the loading response. A vertical ground reaction force threshold of 32 N determined foot contact events, as it was identified in our lab as the lowest threshold to yield reliable and consistent event detection (Auyang et al., 2009; Selgrade and Chang, 2015; Yen and Chang, 2010). All data were time normalized to heel strike for each respective leg. CON data presented are from the right leg. Walking was defined as having a duty factor (stance time/stride time) 0.5 (Gatesy and Biewener, 1991). Post-hoc calculations of the duty factor confirmed walking was achieved for every subject at each speed.

Statistical Analysis

Due to the matched design and extensive subject matching criteria, we employed paired t-tests for between subject group (CON vs. AMP) assessments of gait transition speed and neutral joint angles. Significance was defined as $p\text{-value} < 0.05$. To assess the effect of changes in walking speed on peak forces, separate one-factor (9 speeds: 50–130%GTS), repeated measure analysis of variances (ANOVA) were run for each leg type (CON, Amp-Intact, or Amp-Amputated). When a significant main effect was found, subsequent post-hoc analyses were conducted with significance defined as a Bonferroni corrected $p\text{-value} < 0.05$. To assess the differences in peak flexion and extension joint angles between leg type, separate one-factor (leg type: CON, Amp-Intact, or Amp-Amputated) repeated measure ANOVAS were run for each of the six kinematic variables at each speed (50–130%GTS). When significant main effects of leg type were found subsequent post-hoc analyses was conducted with significance defined as a Bonferroni corrected $p\text{-value} < 0.05$.

RESULTS

Gait transition speeds

AMP subjects transitioned between gaits at $1.73 \pm 0.13\text{m/s}$, which was significantly slower than the CON subjects' GTS of $2.09 \pm 0.05\text{m/s}$ ($p < 0.01$) (Fig. 1). All AMP subjects transitioned between gaits at slower speeds than their matched CON. There was no significant hysteresis for CON subjects' transition speeds, in that the transition speeds determined by incrementally increasing and decreasing treadmill speeds were not significantly different (2.09 ± 0.10 and $2.08 \pm 0.08\text{m/s}$ respectively). AMP subjects' transition speeds from the incrementally increasing portion of the protocol ($1.75 \pm 0.12\text{m/s}$) showed a small but significant difference from the incrementally decreasing portion ($1.71 \pm 0.13\text{m/s}$) ($p < 0.01$). However, the average difference of 0.04m/s between AMP subjects' transition speeds was approximately a 2% difference and below the 0.10m/s resolution used for the incremental treadmill change.

Ground Reaction Forces

Significant main effects were found for speed (50–130%GTS) in all force variables except for AMP-Amputated peak vertical propulsive force production. Representative subject data show CON subjects' anterior-posterior peak propulsive force production increased with speed until 100%GTS, and decreased at speeds above the GTS (Fig. 2). CON subjects peak anterior-posterior propulsive forces significantly increased, when compared to the preceding speed at 60–80%GTS, and significantly decreased at 120%GTS (110%: $0.27 \pm 0.04 > 120\%: 0.23 \pm 0.05$, $p < 0.05$) (Fig. 3A, Table 2). Peak anterior-posterior braking forces for CON subjects significantly increased with speed from 60–100%GTS (Fig. 3B, Table 2). There were no significant differences in CON subject peak anterior-posterior braking forces at speeds above 100%GTS.

Peak anterior-posterior propulsive force production did not significantly decrease during walking above 100%GTS in unilateral, transtibial amputees (Fig. 2 & 3A). In AMP-Intact and AMP-Amputated sides, peak anterior-posterior propulsive forces significantly increased with speed from 60–110%GTS and 60–100%GTS respectively (Fig. 3A, Table 2). AMP-Intact peak anterior-posterior propulsive force production significantly increased at speed 110%GTS in comparison to 100%GTS (100%: $0.28 \pm 0.04 < 110\%: 0.30 \pm 0.04$, $p < 0.05$) (Fig. 3A, Table 2). Peak anterior-posterior braking forces significantly increased with walking speed from 60–110%GTS for AMP-Intact and AMP-Amputated sides ($p < 0.05$) (Fig. 3B, Table 2). Speeds above 110%GTS for AMP subjects' intact and amputated sides had no significant difference from the speed that immediately preceded them.

To assess how the peak propulsive forces changed over absolute speed rather than relative to the GTS, we applied second-order polynomial curves, as they provided the best fit to the parabolic data scatters of CON ($y = -0.16x^2 + 0.63x - 0.33$, $r^2 = 0.67$) and AMP-Intact ($y = -0.09x^2 + 0.42x - 0.17$, $r^2 = 0.71$) (Fig. 3C). We found that changes in AMP-Intact peak propulsive force production tracked with those of CON subjects. Similar results were seen with the AMP-Amputated peak propulsive force production, however, always lower in amplitude. AMP-Amputated peak propulsive force production ranged from $3.45 \pm 2.88\%$ body weight (50%GTS) to $9.2 \pm 4.20\%$ body weight (110%GTS) below AMP-Intact side peak anterior-posterior propulsive force production.

Kinematics

There were no significant main effects of leg type (CON, Amp-Intact, or Amp-Amputated) on joint angles during dorsiflexion or peak flexion or extension angles for the knee or hip at any speed relative to the GTS (50–130%) (Fig. 4, Table 3). Significant main effects for leg type were found at every relative speed for peak ankle plantarflexion. Post-hoc analysis revealed CON and AMP-Intact had significantly more peak plantarflexion than AMP-Amputated at every relative speed ($p < 0.05$). Post-hoc analysis also showed there was no significant difference between CON and AMP-Intact side peak plantarflexion angle at any relative speed. No significant differences between CON and AMP-Intact were found for peak flexion or extension angle at the ankle, knee, or hip at any relative speed. This verified that Amp-Intact kinematics were similar to those of CON subjects at each relative speed (Fig. 4). Mauchey's test of sphericity was satisfied for comparisons of ankle plantarflexion

and dorsiflexion and knee flexion and extension. Mauchey's test of sphericity was violated for hip flexion at speeds of 60–110%GTS and hip extension at 120%GTS, therefore degrees of freedom were corrected using Greenhouse-Geisser estimates of sphericity. Regardless, we found no main effects of leg type on peak hip flexion or extension.

DISCUSSION

This study aimed to determine the GTS of AMP subjects and assess whether intact side propulsive force production (indicative of intrinsic mechanical limits of the plantar flexor muscles) is a major determinant of their walk-to-run gait transition. Consistent with our first hypothesis, AMP subjects transition between gaits at a significantly slower absolute speed than CON subjects (1.73 ± 0.13 m/s and 2.09 ± 0.05 m/s respectively). The CON subjects' gait transition speeds were consistent with previous work that has defined able-bodied gait transition speeds ranging from 1.89 and 2.16 m/s (Diedrich and Warren, 1995, 1998; Hreljac, 1993a, 1995; Hreljac et al., 2001; Hreljac et al., 2008; Kram et al., 1997; Li and Ogden, 2012; Mercier et al., 1994; Prilutsky and Gregor, 2001; Raynor et al., 2002; Segers et al., 2006; Sentija et al., 2012; Thorstensson and Roberthson, 1987; Turvey et al., 1999; Van Caekenberghe et al., 2010). Although the gait transition speeds were determined using different ranges of speeds for CON and AMP subjects (CON: 1.3–2.6 m/s; AMP: 1.0–2.3 m/s), we believe this had no bearing on the difference found between the two groups. This is supported by the fact that Prilutsky and Gregor previously reported an able-bodied GTS of 2.1 ± 0.2 m/s, which is consistent with our findings of 2.09 ± 0.05 m/s, even though they utilized a different range of speeds to determine that GTS (1.4–3.0 m/s) (2001). The higher variability in AMP subject gait transition speeds were not surprising given the nature of pathologic gait. AMP subjects using passive-elastic prostheses also have increased variability in preferred walking speeds (Houdijk et al., 2009). Furthermore, despite the hysteresis between the AMP subject's increasing and decreasing gait transition speeds, we were confident that collecting data relative to the averaged GTS (1.73 ± 0.13 m/s), which is only ± 0.02 m/s from the average transition speeds achieved during the increasing and decreasing protocols, would allow for a more versatile dataset to make walk-to-run and run-to-walk comparisons in the future without affecting our main conclusions.

Although the ankle-foot componentry worn by the AMP subjects was not standardized, allowing subjects to wear their own passive-elastic ankle-foot components have yielded meaningful and consistent results of both muscle activations and walking mechanics within previous studies (Adamczyk and Kuo, 2014; Fey et al., 2010; Silverman et al., 2008). Given the consistent decrease in GTS across all AMP subjects with respect to their matched CON, and similar variability in AMP-subject outcome measures compared to CON, we are confident that our inclusion criterion of testing only AMP subjects with passive-elastic ankle-foot components was sufficient to maintain the group homogeneity.

Contrary to our second hypothesis, AMP subjects were able to continue generating higher peak propulsive forces on both their intact and amputated sides at speeds beyond their preferred GTS. Minimal differences in kinematics between CON subjects and the AMP-Intact side verified amputee subjects were not using a different walking strategy. Results from CON subjects exhibiting limited peak propulsive force production above GTS were

consistent with previous work (Neptune and Sasaki, 2005). The limited propulsive force production at higher than normal walking speeds found in CON subjects, supports previous findings implicating the plantar flexors and their unique force-length-velocity characteristics as a major determinant of the walk-to-run gait transition in able-bodied subjects (Arnold et al., 2013; Farris and Sawicki, 2012; Malcolm et al., 2009a; Neptune and Sasaki, 2005). Conversely, the AMP subjects' abilities to increase propulsive force production at walking speeds above their preferred GTS suggest the force-length-velocity characteristics of the plantar flexors are not a major determinant of the walk-to-run transition in unilateral, transtibial, amputees. In assessing the effects of absolute speed changes on propulsive force production, we observed the polynomial fits of CON and AMP-Intact anterior-posterior propulsive forces tracked together. Additionally, the 95% confidence intervals of the data almost completely overlay one another. Since AMP subjects had significantly slower GTS, these findings support the idea that that propulsive force production is a function of changes in absolute speed rather than relative to the preferred walk-to-run transition speed. Thus, there must be another variable or variables primarily responsible for the gait transition in the AMP population.

These findings do not rule out previous explanations for the determination of the GTS that feature a combination of determinants (Bartlett and Kram, 2008; Malcolm et al., 2009b). Bartlett and Kram found that selectively perturbing the demand on dorsiflexors, plantarflexors, and hip flexors resulted in altered walk-to-run transition speeds, however perturbing multiple muscle groups at once did not produce a summative effect (2008). They introduced the theoretical framework that a "critical threshold of influence" must be met in order for the walk-to-run transition to occur, in which an ultimate, underlying trigger can be influenced by various muscle groups, whose impact can change with walking conditions (e.g. incline walking). The "weakest link" concept, introduced by Malcolm et al., is another theoretical framework supporting the idea that there are multiple determinants or factors affecting the walk-to-run transition speed, and whichever one is stressed to a critical point first, ultimately determines the speed of the transition (2009). Most likely the impaired nature of AMP subject walking, due to loss of the foot-ankle complex, amplifies another determinant of the walk-to-run gait transition to a level in which it becomes the limiting factor. However, there is also the possibility that the same determinant of the walk-to-run gait transition is responsible for the CON and AMP subject transitions, and the plantar flexor force-length-velocity characteristics associated with the gait transition in CON subjects is correlative rather than causal. As we did not directly measure muscle fascicle force, fascicle shortening velocity, or fascicle length changes, future investigation of AMP subject gait transitions would benefit from using ultrasound to directly measure AMP-Intact plantar flexor force-length-velocity characteristics.

In conclusion, since unilateral, transtibial amputees transition between gaits at a slower speed than able-bodied controls and are still able to generate higher propulsive forces walking at speeds above their preferred gait transition speed, their walk-to-run transition is not likely dictated by the force-length-velocity characteristics of the intact plantar flexor muscles. This suggests that, as an experimental model, unilateral, transtibial amputees can provide unique insights for decoupling the previously identified performance limit of plantar flexor muscles from the preferred gait transition speed in order to probe other potential

determinants. To our knowledge, this is the first investigation of gait transitions in persons with a lower limb amputation.

Acknowledgments

This work was supported by NICHD-5T32HD055180 (TNG), NINDS-5R01NS069655 (YHC) and NSF PAC BCS-0847325 (YHC). The authors would like to thank our research participants, Georgia Tech's School of Applied Physiology, and Dr. Teresa Snow for her statistical expertise.

References

- Adamczyk PG, Kuo AD. Mechanisms of gait asymmetry due to push-off deficiency in unilateral amputees. *IEEE Trans Neural Syst Rehabil Eng*. 2014; 23:776–785. [PubMed: 25222950]
- Anderson FC, Pandy MG. Individual muscle contributions to support in normal walking. *Gait & Posture*. 2003; 17:159–169. [PubMed: 12633777]
- Arnold EM, Hamner SR, Seth A, Millard M, Delp SL. How muscle fiber lengths and velocities affect muscle force generation as humans walk and run at different speeds. *Journal of Experimental Biology*. 2013; 216:2150–2160. [PubMed: 23470656]
- Auyang AG, Yen JT, Chang YH. Neuromechanical stabilization of leg length and orientation through interjoint compensation during human hopping. *Experimental Brain Research*. 2009; 192:253–264. [PubMed: 18839158]
- Bartlett JL, Kram R. Changing the demand on specific muscle groups affects the walk-run transition speed. *Journal of Experimental Biology*. 2008; 211:1281–1288. [PubMed: 18375853]
- D'Andrea S, Wilhelm N, Silverman AK, Grabowski AM. Does use of a powered ankle-foot prosthesis restore whole-body angular momentum during walking at different speeds? *Clinical Orthopaedics and Related Research*. 2014; 472:3044–3054. [PubMed: 24781926]
- Diedrich FJ, Warren WH. Why change gaits? Dynamics of the walk-run transition. *Journal of Experimental Psychology-Human Perception and Performance*. 1995; 21:183–202. [PubMed: 7707029]
- Diedrich FJ, Warren WH. The dynamics of gait transitions: Effects of grade and load. *Journal of Motor Behavior*. 1998; 30:60–78. [PubMed: 20037021]
- Farris DJ, Sawicki GS. Human medial gastrocnemius force-velocity behavior shifts with locomotion speed and gait. *Proc Natl Acad Sci U S A*. 2012; 109:977–982. [PubMed: 22219360]
- Fey NP, Silverman AK, Neptune RR. The influence of increasing steady-state walking speed on muscle activity in below-knee amputees. *Journal of Electromyography and Kinesiology*. 2010; 20:155–161. [PubMed: 19303796]
- Gailey RS, Wenger MA, Raya M, Kirk N, Erbs K, Spyropoulos P, Nash MS. Energy-expenditure of trans-tibial amputees during ambulation at self-selected pace. *Prosthetics and Orthotics International*. 1994; 18:84–91. [PubMed: 7991365]
- Gatesy SM, Biewener AA. Bipedal locomotion: Effects of speed, size and limb posture in birds and humans. *Journal of Zoology*. 1991; 224:127–147.
- Genin JJ, Bastien GJ, Franck B, Detrembleur C, Willems PA. Effect of speed on the energy cost of walking in unilateral traumatic lower limb amputees. *European Journal of Applied Physiology*. 2008; 103:655–663. [PubMed: 18478251]
- Herr HM, Grabowski AM. Bionic ankle-foot prosthesis normalizes walking gait for persons with leg amputation. *Proceedings of the Royal Society B-Biological Sciences*. 2012; 279:457–464.
- Houdijk H, Pollmann E, Groenewold M, Wiggerts H, Polomski W. The energy cost for the step-to-step transition in amputee walking. *Gait & Posture*. 2009; 30:35–40. [PubMed: 19321343]
- Hreljac A. Determinants of the gait transition speed during human locomotion: kinetic factors. *Gait & Posture*. 1993a; 1:217–223.
- Hreljac A. Preferred and energetically optimal gait transition speeds in human locomotion. *Medicine and Science in Sports and Exercise*. 1993b; 25:1158–1162. [PubMed: 8231761]

- Hreljac A. Determinants of the gait transition speed during human locomotion: kinematic factors. *Journal of Biomechanics*. 1995; 28:669–677. [PubMed: 7601866]
- Hreljac A, Arata A, Ferber R, Mercer JA, Row BS. An electromyographical analysis of the role of dorsiflexors on the gait transition during human locomotion. *Journal of Applied Biomechanics*. 2001; 17:287–296.
- Hreljac A, Limamura RT, Escamilla RF, Edwards WB, MacLeod T. The relationship between joint kinetic factors and the walk-run gait transition speed during human locomotion. *Journal of Applied Biomechanics*. 2008; 24:149–157. [PubMed: 18579907]
- Kram R, Domingo A, Ferris DP. Effect of reduced gravity on the preferred walk-run transition speed. *Journal of Experimental Biology*. 1997; 200:821–826. [PubMed: 9076966]
- Kram R, Griffin TM, Donelan JM, Chang YH. Force treadmill for measuring vertical and horizontal ground reaction forces. *Journal of Applied Physiology*. 1998; 85:764–769. [PubMed: 9688758]
- Li L, Ogden LL. Muscular activity characteristics associated with preparation for gait transition. *Journal of Sport and Health Science*. 2012; 1:27–35.
- Malcolm P, Fiers P, Segers V, Van Caekenberghe I, Lenoir M, De Clercq D. Experimental study on the role of the ankle push off in the walk-to-run transition by means of a powered ankle-foot-exoskeleton. *Gait & Posture*. 2009a; 30:322–327. [PubMed: 19576776]
- Malcolm P, Segers V, Van Caekenberghe I, De Clercq D. Experimental study of the influence of the m. tibialis anterior on the walk-to-run transition by means of a powered ankle-foot exoskeleton. *Gait & Posture*. 2009b; 29:6–10. [PubMed: 18620862]
- Mercier J, Legallais D, Durand M, Goudal C, Micallef JP, Prefaut C. Energy-expenditure and cardiorespiratory responses at the transition between walking and running. *European Journal of Applied Physiology and Occupational Physiology*. 1994; 69:525–529. [PubMed: 7713073]
- Minetti AE, Ardigo LP, Saibene F. The transition between walking and running in humans: metabolic and mechanical aspects at different gradients. *Acta Physiologica Scandinavica*. 1994; 150:315–323. [PubMed: 8010138]
- Neptune RR, Sasaki K. Ankle plantar flexor force production is an important determinant of the preferred walk-to-run transition speed. *Journal of Experimental Biology*. 2005; 208:799–808. [PubMed: 15755878]
- Neptune RR, Zajac FE, Kautz SA. Muscle mechanical work requirements during normal walking: the energetic cost of raising the body's center-of-mass is significant. *Journal of Biomechanics*. 2004; 37:817–825. [PubMed: 15111069]
- Nolan L, Wit A, Dudzinski K, Lees A, Lake M, Wychowski M. Adjustments in gait symmetry with walking speed in trans-femoral and trans-tibial amputees. *Gait & Posture*. 2003; 17:142–151. [PubMed: 12633775]
- Powers CM, Rao S, Perry J. Knee kinetics in trans-tibial amputee gait. *Gait & Posture*. 1998; 8:1–7. [PubMed: 10200393]
- Prilutsky BI, Gregor RJ. Swing- and support- related muscle activations differentially trigger human walk-run and run-walk transitions. *The Journal of Experimental Biology*. 2001:2277–2287. [PubMed: 11507111]
- Raynor AJ, Yi CJ, Abernethy B, Jong QJ. Are transitions in human gait determined by mechanical, kinetic or energetic factors? *Human Movement Science*. 2002; 21:785–805. [PubMed: 12620720]
- Sanderson DJ, Martin PE. Lower extremity kinematic and kinetic adaptations in unilateral below-knee amputees during walking. *Gait & Posture*. 1997; 6:126–136.
- Segers V, Aerts P, Lenoir A, De Clercq D. Spatiotemporal characteristics of the walk-to-run and run-to-walk transition when gradually changing speed. *Gait & Posture*. 2006; 24:247–254. [PubMed: 16314100]
- Segers V, Lenoir M, Aerts P, De Clercq D. Influence of M. tibialis anterior fatigue on the walk-to-run and run-to-walk transition in non-steady state locomotion. *Gait & Posture*. 2007; 25:639–647. [PubMed: 17049861]
- Selgrade BP, Chang YH. Locomotor control of limb force switches from minimal intervention principle in early adaptation to noise reduction in late adaptation. *Journal of Neurophysiology*. 2015; 113:1451–1461. [PubMed: 25475343]

- Sentija D, Rakovac M, Babic V. Anthropometric characteristics and gait transition speed in human locomotion. *Human Movement Science*. 2012; 31:672–682. [PubMed: 21835478]
- Silverman AK, Fey NP, Portillo A, Walden JG, Bosker G, Neptune RR. Compensatory mechanisms in below-knee amputee gait in response to increasing steady-state walking speeds. *Gait & Posture*. 2008; 28:602–609. [PubMed: 18514526]
- Silverman AK, Neptune RR. Muscle and prosthesis contributions to amputee walking mechanics: A modeling study. *Journal of Biomechanics*. 2012; 45:2271–2278. [PubMed: 22840757]
- Snyder RD, Powers CM, Fontaine C, Perry J. The effect of five prosthetic feet on the gait and loading of the sound limb in dysvascular below-knee amputees. *Journal of Rehabilitation Research and Development*. 1995; 32:309–315. [PubMed: 8770795]
- Thorstensson A, Rotherthson H. Adaptations to changing speed in human locomotion: speed of transition between walking and running. *Acta Physiologica Scandinavica*. 1987; 131:211–214. [PubMed: 3673618]
- Torburn L, Perry J, Ayyappa E, Shanfield SL. Below-knee amputee gait with dynamic elastic response prosthetic feet: a pilot study. *Journal of Rehabilitation Research and Development*. 1990; 27:369–384. [PubMed: 2089148]
- Turvey MT, Holt KG, LaFiandra ME, Fonseca ST. Can the transitions to and from running and the metabolic cost of running be determined from the kinetic energy of running? *Journal of Motor Behavior*. 1999; 31:265–278.
- Van Caekenberghe I, Segers V, De Smet K, Aerts P, De Clercq D. Influence of treadmill acceleration on actual walk-to-run transition. *Gait & Posture*. 2010; 31:52–56. [PubMed: 19796948]
- Waters RL, Perry J, Antonelli D, Hislop H. Energy cost of walking of amputees: the influence of level of amputation. *Journal of Bone and Joint Surgery*. 1976; 58:42–46. [PubMed: 1249111]
- Winter DA, Sienko SE. Biomechanics of below-knee amputee gait. *Journal of Biomechanics*. 1988; 21:361–367. [PubMed: 3417688]
- Yen JT, Chang YH. Rate-dependent control strategies stabilize limb forces during human locomotion. *Journal of the Royal Society Interface*. 2010; 7:801–810.
- Zajac FE, Neptune RR, Kautz SA. Biomechanics and muscle coordination of human walking - Part I: Introduction to concepts, power transfer, dynamics and simulations. *Gait & Posture*. 2002; 16:215–232. [PubMed: 12443946]
- Zajac FE, Neptune RR, Kautz SA. Biomechanics and muscle coordination of human walking Part II: Lessons from dynamical simulations and clinical implications. *Gait & Posture*. 2003; 17:1–17. [PubMed: 12535721]

Figure 1.

Gait transition speeds: Unilateral, transtibial amputees (AMP) transition between gaits at significantly slower speeds than able-bodied controls (CON) (* $p < 0.01$). Mean $\pm$ standard deviation.

Figure 2.

Ground reaction forces: Representative vertical and anterior-posterior ground reaction forces (%Body Weight) during walking plotted for each relative speed 50–130% GTS from a matched control (CON) and unilateral, transtibial amputee (AMP) subject (lightest blue to darkest blue: 50–90% GTS; black: 100% GTS; lightest red to darkest red: 110–130% GTS).

Figure 3.

Propulsive and braking forces: All forces are represented as a % of body weight. (A) Mean peak anterior-posterior propulsive force $\pm$ standard deviation with respect to relative speed (% GTS) for CON subjects (grey circles), AMP-Intact side (black squares), and AMP-Amputated (white squares). (B) Mean peak anterior-posterior braking force $\pm$ standard deviation with respect to % GTS for CON subjects (grey circles), AMP-Intact side (black squares), and AMP-Amputated (white squares). (C) Peak anterior-posterior propulsive force with respect to absolute speed (m/s) plotted as quadratic polynomial curves with the 95% confidence interval of the data for each group; CON (grey line, white 95% confidence interval) ($y = -0.16x^2 + 0.63x - 0.33$, $r^2 = 0.67$), AMP-Intact (black line, grey 95% confidence interval) ($y = -0.09x^2 + 0.42x - 0.17$, $r^2 = 0.71$). (D) Peak anterior-posterior braking force with respect to absolute speed (m/s) plotted as quadratic polynomial curves with the 95% confidence interval of the data for each group; CON (grey line, white 95% confidence interval) ($y = -0.08x^2 + 0.40x - 0.21$, $r^2 = 0.61$), AMP-Intact (black line, grey 95% confidence interval) ($y = -0.01x^2 + 0.20x - 0.08$, $r^2 = 0.76$).

Figure 4.

Kinematics: Mean ankle, knee, and hip kinematics of all subjects walking at each speed relative to the gait transition speed (50–130% GTS). CON (grey line), AMP-Intact (black solid line), AMP-Amputated (black dashed line).

Table 1

Subject Information:

Matched Pair #	Sex	Leg Length (cm)		Weight (kg)		Age (years)		Transfemoral Amputation			Prosthesis Characteristics		
		AMP (CON)		AMP (CON)		AMP (CON)		Type	Side	Years Since	Years with Foot	Foot Component Type	Suspension
1	F	93.0 (93.5)		68.4 (70.9)		37.3 (23.5)		Traumatic	Right	5.4	1.2	Runway (Freedom)	Elevated Vacuum
2	F	87.2 (86.6)		56.0 (61.3)		28.5 (25.7)		Congenital (fibularhemilemia)	Right	28.0	1.0	Variflex (Ossur)	Supracondylar
3	F	87.3 (86.3)		53.1 (56.5)		38.2 (28.8)		Traumatic	Left	2.1	0.6	Seattle Triumph (TruLife)	Elevated Vacuum
4	F	97.5 (86.5)		69.2 (59.7)		27.9 (32.4)		Congenital (pseudoarthritis of the tibia)	Left	0.7	0.7	Accent (College Park)	Suction & Sleeve
5	F	81.5 (83.3)		65.5 (62.6)		18.3 (21.4)		Traumatic	Left	2.6	2.5	Highlander (Freedom)	Pin
6	M	101.3 (98.3)		77.4 (81.5)		60.0 (51.9)		Traumatic	Right	3.9	2.0	Renegade AT (Freedom)	Elevated Vacuum
7	M	91.0 (94.0)		83.9 (89.3)		35.3 (36.5)		Traumatic	Right	11.9	2.0	Renegade (Freedom)	Elevated Vacuum
8	M	88.0 (89.0)		63.5 (65.7)		21.4 (21.7)		Traumatic	Right	1.4	11.0	Renegade (Freedom)	Elevated Vacuum
9	M	102.5 (100.8)		64.5 (70.0)		21.4 (23.0)		Congenital (club foot)	Right	3.0	1.5	Variflex (Ossur)	Suction & Sleeve
10	M	94.0 (93.8)		97.8 (87.3)		42.0 (32.6)		Traumatic	Right	10.1	2.0	Trias (OttoBock)	Elevated Vacuum

Table 2

Walking Propulsive & Braking Forces:

		Peak Anterior-Posterior Ground Reaction Forces				Peak Vertical Ground Reaction Forces			
		% GTS	CON	AMP- Intact	AMP-Amputated	CON	AMP-Intact	AMP-Amputated	
Propulsion (Body Weight)	50	0.16 ± 0.02	0.13 ± 0.02	0.10 ± 0.02	0.10 ± 0.02	1.05 ± 0.04	0.97 ± 0.04	0.97 ± 0.03	
	60	0.19 ± 0.01*	0.16 ± 0.02*	0.11 ± 0.02*	0.11 ± 0.02*	1.09 ± 0.05*	1.00 ± 0.06*	0.99 ± 0.05	
	70	0.23 ± 0.02*	0.18 ± 0.03*	0.13 ± 0.02*	0.13 ± 0.02*	1.16 ± 0.06*	1.04 ± 0.07*	1.01 ± 0.06	
	80	0.27 ± 0.02*	0.22 ± 0.03*	0.15 ± 0.02*	0.15 ± 0.02*	1.19 ± 0.07	1.10 ± 0.10*	1.02 ± 0.07	
	90	0.29 ± 0.02	0.25 ± 0.04*	0.17 ± 0.02*	0.17 ± 0.02*	1.21 ± 0.10	1.14 ± 0.12*	1.03 ± 0.09	
	100	0.31 ± 0.03	0.28 ± 0.04*	0.19 ± 0.03*	0.19 ± 0.03*	1.20 ± 0.15	1.17 ± 0.13	1.05 ± 0.12	
	110	0.27 ± 0.04	0.30 ± 0.04*	0.20 ± 0.05	0.20 ± 0.05	1.12 ± 0.14	1.18 ± 0.14	1.05 ± 0.17	
Braking (Body Weight)	120	0.23 ± 0.05*	0.29 ± 0.05	0.21 ± 0.06	0.21 ± 0.06	1.00 ± 0.18	1.14 ± 0.18	1.03 ± 0.24	
	130	0.21 ± 0.05	0.26 ± 0.07	0.18 ± 0.06	0.18 ± 0.06	0.98 ± 0.18	1.06 ± 0.23	0.96 ± 0.25	
	50	-0.13 ± 0.02	-0.09 ± 0.03	-0.08 ± 0.02	-0.08 ± 0.02	1.04 ± 0.04	1.04 ± 0.04	1.04 ± 0.04	
	60	-0.17 ± 0.02*	-0.12 ± 0.03*	-0.09 ± 0.01*	-0.09 ± 0.01*	1.12 ± 0.07*	1.06 ± 0.06	1.05 ± 0.05	
	70	-0.20 ± 0.02*	-0.14 ± 0.03*	-0.11 ± 0.02*	-0.11 ± 0.02*	1.20 ± 0.08*	1.11 ± 0.08*	1.06 ± 0.07	
	80	-0.24 ± 0.01*	-0.17 ± 0.03*	-0.13 ± 0.02*	-0.13 ± 0.02*	1.27 ± 0.08*	1.20 ± 0.08*	1.10 ± 0.09	
	90	-0.27 ± 0.02*	-0.21 ± 0.05*	-0.16 ± 0.03*	-0.16 ± 0.03*	1.36 ± 0.11*	1.30 ± 0.09*	1.16 ± 0.11*	
	100	-0.31 ± 0.05*	-0.24 ± 0.05*	-0.18 ± 0.03*	-0.18 ± 0.03*	1.49 ± 0.09*	1.41 ± 0.12*	1.24 ± 0.11*	
	110	-0.31 ± 0.06	-0.28 ± 0.06*	-0.22 ± 0.04*	-0.22 ± 0.04*	1.55 ± 0.11	1.50 ± 0.11*	1.37 ± 0.10*	
	120	-0.29 ± 0.09	-0.28 ± 0.06	-0.23 ± 0.06	-0.23 ± 0.06	1.63 ± 0.19	1.58 ± 0.17	1.49 ± 0.10*	
	130	-0.29 ± 0.09	-0.27 ± 0.09	-0.22 ± 0.08	-0.22 ± 0.08	1.63 ± 0.18	1.60 ± 0.19	1.58 ± 0.16	

Control (CON) and unilateral, transtibial amputee intact (AMP-Intact) and amputated (AMP-Amputated) side peak anterior-posterior and vertical ground reaction forces (during braking and propulsion (mean ± standard deviation) for walking at speeds 50–130% of the gait transition speed. Paired t-tests were performed to assess whether or not each speed was significantly different from the preceding slower speed (*p<0.05). Bold means significantly lower than preceding slower speed.

Peak Kinematics

Table 3

		% Gait Transition Speed										
		50	60	70	80	90	100	110	120	130		
Ankle Plantarflexion	CON	-20.0 ± 6.3	-22.3 ± 5.7	-23.1 ± 5.1	-23.5 ± 5.5	-24.3 ± 5.2	-24.1 ± 4.8	-23.0 ± 5.4	-23.2 ± 5.8	-22.4 ± 5.9		
	AMP-Intact	-15.4 ± 6.1	-16.9 ± 5.8	-17.7 ± 5.5	-18.9 ± 5.8	-19.6 ± 6.0	-20.6 ± 6.0	-20.6 ± 6.0	-20.5 ± 6.5	-19.6 ± 7.0		
	AMP-Amputated	-3.5 ± 2.4 ^{*,†}	-4.0 ± 2.9 ^{*,†}	-4.0 ± 2.9 ^{*,†}	-4.3 ± 2.9 ^{*,†}	-4.3 ± 2.9 ^{*,†}	-4.5 ± 2.7 ^{*,†}	-4.4 ± 2.7 ^{*,†}	-5.1 ± 3.2 ^{*,†}	-4.5 ± 2.5 ^{*,†}		
Ankle Dorsiflexion	CON	9.3 ± 3.5	9.4 ± 3.1	9.6 ± 2.8	8.8 ± 3.9	8.8 ± 4.0	9.0 ± 4.2	10.2 ± 4.4	11.9 ± 5.9	13.3 ± 6.4		
	AMP-Intact	8.5 ± 4.4	8.5 ± 5.1	8.1 ± 5.1	8.1 ± 4.9	7.6 ± 5.3	7.3 ± 5.1	7.3 ± 5.4	7.8 ± 5.8	10.4 ± 5.6		
	AMP-Amputated	6.8 ± 2.4	7.1 ± 2.9	7.4 ± 3.0	7.9 ± 3.2	8.4 ± 3.5	8.7 ± 3.5	9.2 ± 3.8	8.9 ± 3.9	8.4 ± 4.1		
Knee Flexion	CON	63.4 ± 5.9	64.4 ± 5.7	64.6 ± 5.0	63.8 ± 4.8	64.1 ± 4.7	63.8 ± 5.3	65.8 ± 4.7	68.5 ± 5.0	71.4 ± 5.9		
	AMP-Intact	59.3 ± 7.2	61.6 ± 7.0	61.9 ± 6.5	62.3 ± 6.4	62.6 ± 6.4	62.8 ± 6.2	63.9 ± 6.3	65.4 ± 6.5	67.1 ± 7.2		
	AMP-Amputated	63.0 ± 11.0	64.6 ± 11.9	67.7 ± 11.2	69.2 ± 11.9	71.0 ± 11.9	72.9 ± 11.9	74.5 ± 12.2	77.1 ± 13.4	80.7 ± 14.1		
Knee Extension	CON	0.6 ± 5.7	0.5 ± 5.4	0.5 ± 5.4	0.9 ± 5.2	1.1 ± 5.3	1.1 ± 5.8	1.6 ± 5.7	3.0 ± 6.2	4.6 ± 7.0		
	AMP-Intact	2.5 ± 4.0	2.4 ± 3.4	1.4 ± 3.7	0.9 ± 3.5	-0.1 ± 3.1	-0.1 ± 3.5	-0.8 ± 3.7	-0.9 ± 3.6	2.8 ± 5.7		
	AMP-Amputated	10.9 ± 13.0	9.2 ± 13.5	8.7 ± 12.8	7.9 ± 12.6	6.8 ± 12.4	7.1 ± 12.2	6.6 ± 12.5	6.6 ± 12.8	10.4 ± 15.6		
Hip Flexion	CON	27.7 ± 3.9	29.9 ± 3.5	31.0 ± 4.1	33.7 ± 4.4	35.9 ± 4.0	38.7 ± 3.2	40.9 ± 2.5	42.4 ± 3.6	42.8 ± 3.8		
	AMP-Intact	25.5 ± 6.9	27.1 ± 7.3	29.1 ± 7.7	30.9 ± 8.1	34.1 ± 9.0	37.5 ± 9.0	39.8 ± 9.6	42.4 ± 9.0	43.9 ± 9.4		
	AMP-Amputated	27.1 ± 3.6	27.5 ± 3.9	29.1 ± 4.0	30.7 ± 4.0	32.6 ± 4.0	35.7 ± 5.0	37.8 ± 5.5	41.0 ± 5.8	44.3 ± 7.2		
Hip Extension	CON	-12.1 ± 4.2	-14.0 ± 3.9	-16.6 ± 3.6	-17.5 ± 3.2	-18.8 ± 3.9	-19.7 ± 4.1	-18.1 ± 2.4	-15.5 ± 3.2	-15.0 ± 3.0		
	AMP-Intact	-10.9 ± 6.7	-11.6 ± 6.0	-13.9 ± 6.0	-16.2 ± 6.4	-17.6 ± 6.7	-17.8 ± 7.3	-18.8 ± 8.0	-18.1 ± 8.1	-14.9 ± 10.5		
	AMP-Amputated	-9.6 ± 4.1	-11.6 ± 4.9	-12.7 ± 4.9	-14.4 ± 5.0	-15.1 ± 5.5	-14.7 ± 5.5	-15.2 ± 6.3	-14.2 ± 6.8	-11.8 ± 8.9		

Peak kinematic joint angles in degrees (mean ± standard deviation) for CON, AMP-Intact, and AMP-Amputated. Significance presented from Bonferroni post-hoc testing when significant main effects were present after one-way repeated measures ANOVA.

* Significantly different from CON;

[†] Significantly different from AMP-Intact (p<0.05, Bonferroni corrected).