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The influence of increasing steady-state walking speed on muscle activity in below-knee amputees

N.P. Fey, A.K. Silverman, R.R. Neptune *

Department of Mechanical Engineering, The University of Texas at Austin, 1 University Station C2200, Austin, TX 78712, USA

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ABSTRACT

The goal of this study was to identify changes in muscle activity in below-knee amputees in response to increasing steady-state walking speeds. Bilateral electromyographic (EMG) data were collected from 14 amputee and 10 non-amputee subjects during four overground walking speeds from eight intact leg and five residual leg muscles. Using integrated EMG measures, we tested three hypotheses for each muscle: (1) there would be no difference in muscle activity between the residual and intact legs, (2) there would be no difference in muscle activity between the intact leg and non-amputee legs, and (3) muscle activity in the residual and intact legs would increase with speed. Most amputee EMG patterns were similar between legs and increased in magnitude with speed. Differences occurred in the residual leg biceps femoris long head, vastus lateralis and rectus femoris, which increased in magnitude during braking compared to the intact leg. These adaptations were consistent with the need for additional body support and forward propulsion in the absence of the plantar flexors. With the exception of the intact leg gluteus medius, all intact leg muscles exhibited similar EMG patterns compared to the control leg. Finally, the residual, intact and control leg EMG all had a significant speed effect that increased with speed with the exception of the gluteus medius.

1. Introduction

Below-knee amputation was the second most common type of amputation in the US from 1988 to 1996, with the leading causes linked to vascular disease and traumatic injuries (Dillingham et al., 2002). In 2005, approximately 1.6 million persons were living with limb loss in the US, with this number projected to more than double to 3.6 million by the year 2050. In addition, 40% of these individuals lost their limb due to a major lower extremity amputation (Ziegler-Graham et al., 2008). Previous walking studies have shown that below-knee amputees exhibit greater energy expenditure for a given walking speed, reduced self-selected walking speed and bilateral asymmetry relative to non-amputees (for review, see Hafner et al., 2002). Many of the differences between amputee and non-amputee gait are attributed to the functional loss of the ankle plantar flexors, which are critical to providing body support, forward propulsion and swing initiation in non-amputee walking (Neptune et al., 2001; Anderson and Pandy, 2003; Liu et al., 2006). As a result, significant compensatory mechanisms in amputee gait are needed to fulfill these walking subtasks. Successfully identifying and understanding these mechanisms may help improve prosthetic devices and provide a basis for tailoring rehabilitation methods specific to below-knee amputees.

A number of studies have used inverse dynamics-based analyses to compare ground reaction forces (GRFs) and joint kinetics between amputee and non-amputee walking. In general, these studies have shown amputees have greater intact leg GRFs, stance time and joint moments and powers compared to their residual leg and non-amputees (e.g., Sanderson and Martin, 1997; Nolan and Lees, 2000; Bateni and Olney, 2002; Nolan et al., 2003; Beyaert and Grumillier, 2008). Analyzing the residual leg, studies have identified decreased knee and increased hip moments and powers compared to non-amputees (e.g., Winter and Sienko, 1988; Gitter et al., 1991; Powers et al., 1998; Bateni and Olney, 2002). These studies have provided much insight into the compensatory mechanisms used in below-knee amputee gait at the joint level. The purpose of this study was to analyze muscle activity to gain additional insight into compensatory strategies at the individual muscle level.

Studies investigating changes in muscle activity during amputee walking have used electromyographic (EMG) data and identified significant differences in the muscle activity of several residual leg muscles compared to the intact leg and non-amputees. In particular, increased magnitude and duration of the residual leg uniarticular knee extensors (Culham et al., 1986; Winter and Sienko, 1988; Torburn et al., 1990; Pinzur et al., 1991; Perry and Shanfield, 1993; Powers et al., 1998; Rietman et al., 2002), biarticular

* Corresponding author. Tel.: +1 512 471 0848; fax: +1 512 471 8727.
E-mail address: rneptune@mail.utexas.edu (R.R. Neptune).

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hamstrings (Culham et al., 1986; Winter and Sienko, 1988; Torburn et al., 1990; Pinzur et al., 1991; Perry and Shanfield, 1993; Powers et al., 1998; Isakov et al., 2000, 2001; Schmalz et al., 2001; Rietman et al., 2002) and gluteus maximus (Winter and Sienko, 1988; Torburn et al., 1990; Perry and Shanfield, 1993) muscles have been reported. Intact leg muscle activity has consistently been found to be similar to non-amputees (Culham et al., 1986; Czerniecki, 1996; Rietman et al., 2002). Most of these studies, however, have either analyzed a small set of muscles or did not compare the amputee results with non-amputees during the same walking conditions. Furthermore, no study has examined amputee muscle activity over a wide range of walking speeds. In a recent modeling study of non-amputee walking, the functional roles of muscles were shown to be insensitive to changes in walking speed and that only the magnitudes of the muscle output changed with speed (Neptune et al., 2008). However, it is not clear whether the observed changes in muscle activity in below-knee amputees remain invariant as the mechanical energetic demands change with walking speed.

Two studies have investigated amputee EMG during two different conditions: self-selected and fast walking speeds (Torburn et al., 1990; Perry and Shanfield, 1993). However, both of these studies only report residual leg muscle activity from three muscle groups and did not assess differences across walking speeds. Silverman et al. (2008) examined both GRFs and joint kinetics across speeds from 0.6 to 1.5 m/s and observed consistent increases in GRFs and joint powers within the residual, intact and non-amputee legs as walking speed increased. They found a reduced residual leg braking impulse relative to non-amputees across speeds. In addition, they found increased residual leg positive hip work and reduced residual leg positive knee work over stance relative to non-amputees. These results suggest greater output from the residual leg hip extensors (e.g., the biarticular hamstrings and/or gluteus maximus) as a possible compensatory mechanism to increase the net residual leg propulsion in early stance. This mechanism would be consistent with modeling studies of non-amputee walking that have shown both of these muscles have the potential to generate a positive contribution to propulsion from early to mid-stance (Neptune et al., 2004; Liu et al., 2006). Investigating how changes in walking speed affect muscle activity in below-knee amputees would provide further insight into the compensatory mechanisms used.

The overall goal of this study was to examine bilateral EMG across a wide range of steady-state walking speeds to identify changes in muscle activity in below-knee amputees. To achieve this goal, we tested the following three hypotheses for each muscle analyzed: (1) there will be no difference in muscle activity between the residual and intact leg muscles, (2) there will be no difference in muscle activity between the intact leg and non-amputee muscles, and (3) muscle activity in the amputee residual and intact legs will exhibit increases in magnitude with increases in steady-state walking speed.

2. Methods

The data used in this analysis were part of a larger data set collected in Silverman et al. (2008) that had not been analyzed. The subjects, procedures and analyses performed relevant to the present study are described below.

2.1. Subjects

Participants included 14 unilateral, below-knee amputees (13 males, 1 female; 11 traumatic, 3 vascular; 45 ± 9 years; 199.7 ± 40.9 lbs; 69.2 ± 3.8 in.) and 10 non-amputee, control subjects (7 males, 3 females; 33 ± 12 years; 156.4 ± 30.0 lbs; 69.4 ± 4.2 in.). Each amputee used his or her own prosthesis, with its alignment and fit verified prior to testing by a licensed prosthetist with over 30 years of experience. Prosthesis type varied across subjects including nine energy storage and return and five solid-ankle, cushioned-heel feet. Post-amputation time prior to data collection was greater than or equal to 1.5 years (6 ± 3 years, average) for all amputees. All subjects were free of musculoskeletal disorders and leg pain and were proficient walkers who did not require the use of assistive devices. Each subject provided informed consent that was approved by The University of Texas at Austin and the South Texas Veterans Affairs Medical Center Institutional Review Boards prior to participation.

2.2. Procedures

Kinematic marker data were measured (120 Hz) using a motion capture system (Vicon, Oxford Metrics, Inc.) as each subject walked along a 10 m walkway. The specific marker set used is described in detail in Silverman et al. (2008). Each subject walked along the entire walkway at four randomly-ordered, steady-state speeds of 0.6, 0.9, 1.2 and 1.5 m/s. Average speed was verified using two infrared timing gates. Trials were repeated until a minimum of five gait cycles per foot were measured at each speed ±0.06 m/s.

Electromyographic (EMG) data were collected at 1200 Hz using surface EMG electrodes (2, 12-mm disk sensor contacts, 18-mm interelectrode distance, medical grade stainless steel; Motion Lab Systems, Inc.) from eight intact leg muscles including the tibialis anterior (TA), medial gastrocnemius (GAS), soleus (SOL), vastus lateralis (VAS), rectus femoris (RF), biceps femoris long head (BF), gluteus medius (GMED), and gluteus maximus (GMAX), and from five residual leg muscles including VAS, RF, BF, GMED and GMAX. Detection mode and amplification characteristics included >100 dB at 65 Hz common-mode rejection, 20–500 Hz signal bandwidth, <1.2 μV root-mean-square (RMS) noise level, 20 ± 1% gain at 1 kHz and >100 MΩ input impedance. Electrodes were placed on the muscle belly along the line of action between the origin and insertion points based on guidelines provided by Perotto and Delagi (1994). Each electrode location was shaved and cleaned with alcohol prior to placing the electrodes. A ground electrode was placed on the sacrum. Electrodes were secured with tape and Coban (3M, Inc.) to minimize movement artifact. The non-amputee control subjects were instrumented as left-leg amputees using the same procedures above.

2.3. EMG analysis

To quantify muscle activity within specific regions of the gait cycle, braking (~0–50% stance), propulsion (~50–100% stance) and swing phases were defined using gait cycle events (heel-strike and toe-off of the ipsilateral and contralateral legs). To maximize the number of gait cycles included in the analysis, kinematic marker data of the foot segments were used to define the gait cycle events using Visual3D (C-Motion, Inc.). The braking phase was defined as the first double support phase plus the first half of the single-leg support, while the propulsion phase was defined as the second half of the single-leg support plus the second double support phase.

The raw EMG signals were filtered and processed in MATLAB (MathWorks, Inc.). The data were demeaned, and then smoothed using a moving, symmetric 80-ms (±40 ms) root-mean-square (RMS) window. A wrap-around technique was used at the beginning and end of each gait cycle to account for end-effects and to ensure a constant RMS window width. Smoothed data from each gait cycle were time-normalized, with the stance and swing phases normalized separately to eliminate the effect of stance occupying a larger percentage of the gait cycle at slower speeds (e.g., den Otter et al., 2004). Stance and swing were normalized to represent 60% and 40% of the gait cycle (150 and 100 normalization data points), respectively.

To assess where in the gait cycle differences in muscle activity occurred, integrated EMG (iEMG) for each muscle was calculated as the integral of the smoothed and time-normalized data in four regions, including braking, propulsion, swing and the entire gait cycle. These four iEMG quantities were calculated for each gait cycle and then averaged across trials for each subject at each condition. Each subject average iEMG magnitude during braking, propulsion, swing and over the gait cycle was normalized by their average iEMG magnitude over the entire gait cycle at the highest walking speed (1.5 m/s). The normalized iEMG magnitudes were then averaged across subjects within the amputee and control groups.

Subject average RMS profiles for each muscle were also generated at each speed to aid in the interpretation of the iEMG results. The subject averages were normalized to the peak value of their average RMS profile at 1.5 m/s. The normalized, average RMS profiles were then averaged across subjects within each group to calculate a group average RMS profile at each speed.

2.4. Statistical analysis

Statistical analyses of the iEMG magnitudes were performed using SPSS 16.0 GP (SPSS, Inc.). The analysis included five, three-factor (group, leg, speed), repeated measures ANOVAs for the upper-leg muscles and three, two-factor (group, speed) repeated measures ANOVAs for the intact and control leg ankle muscles. The group factor consisted of two levels (amputee, non-amputee). The leg factor consisted of two levels (residual leg, intact leg). The speed factor consisted of four levels (0.6, 0.9, 1.2 and 1.5 m/s). Between-subject comparisons were made between groups, and within-subject comparisons made between legs and across speeds. This statistical analysis was completed for each of the four regions described previously, including braking, propulsion, swing and the entire gait cycle. If an ANOVA had a significant main or interaction effect, pairwise comparisons were made using a Bonferroni adjustment for multiple comparisons to determine which values were significantly different ($p \leqslant 0.05$). iEMG quantities from the redundant upper-leg muscles of the two control legs were averaged together in the statistical analyses. Thus, the within-subject comparison of the control legs was not examined statistically.

3. Results

3.1. Residual and intact leg comparison

The first hypothesis that there would be no difference between residual and intact leg muscle activity was not supported in all muscles. There were significant differences in the muscle activation patterns between the residual and intact leg muscles during specific regions of the gait cycle (Figs. 1 and 2).

Differences occurred in the residual leg BF during braking. In this region, significantly higher residual leg activity was found relative to the intact leg at all speeds: 0.223 (mean) ± 0.034 (standard deviation) at 0.6 m/s, 0.239 ± 0.027 at 0.9 m/s, 0.270 ± 0.027 at 1.2 m/s and 0.379 ± 0.019 at 1.5 m/s compared to 0.168 ± 0.026, 0.191 ± 0.017, 0.231 ± 0.025 and 0.297 ± 0.019, respectively ($p \leqslant 0.040$, Fig. 1). During swing, the residual leg BF exhibited reduced activity at all speeds relative to the intact leg: 0.146 ± 0.021 at 0.6 m/s, 0.244 ± 0.029 at 0.9 m/s, 0.335 ± 0.030 at 1.2 m/s and 0.430 ± 0.030 at 1.5 m/s compared to 0.229 ± 0.017, 0.333 ± 0.023, 0.426 ± 0.023 and 0.531 ± 0.028, respectively ($p \leqslant 0.011$, Fig. 1). Other differences in muscle activity occurred in VAS during braking. In this region, significantly higher residual leg VAS activity was found relative to the intact leg at all speeds: 0.456 ± 0.044 at 0.6 m/s, 0.480 ± 0.039 at 0.9 m/s, 0.540 ± 0.036 at 1.2 m/s and 0.628 ± 0.027 at 1.5 m/s compared to 0.355 ± 0.030, 0.412 ± 0.035, 0.503 ± 0.034 and 0.562 ± 0.024, respectively ($p \leqslant 0.016$, Fig. 1).

Differences between the residual and intact legs also occurred in RF. During braking, the residual leg RF exhibited significantly higher activity than the intact leg at the three highest speeds: 0.403 ± 0.040 at 0.9 m/s, 0.438 ± 0.038 at 1.2 m/s and 0.462 ± 0.031 at 1.5 m/s compared to 0.285 ± 0.027, 0.311 ± 0.025 and 0.400 ± 0.019, respectively ($p \leqslant 0.019$, Fig. 1).

3.2. Intact and control leg comparison

Intact and control leg muscle activation patterns were very similar across walking speeds (Figs. 1 and 2). However, one difference did occur in GMED during the propulsion phase, which had significantly higher activity in the intact leg, 0.288 ± 0.017, compared to the control leg, 0.187 ± 0.020, at 1.5 m/s ($p < 0.001$, Fig. 1). Therefore, the hypothesis that there would be no difference in muscle activity between the intact and control leg muscles was not supported in GMED during propulsion at 1.5 m/s. However, this hypothesis was supported in the remaining muscles analyzed.

3.3. Speed dependence of the residual and intact legs

In general, the amputee muscle activation patterns increased in magnitude with walking speed similar to the control subjects (Fig. 1). There was a significant speed effect ($p < 0.001$) for all muscles over the gait cycle, with the exception of GMED, which was generally insensitive to changes in walking speed (Fig. 1). Therefore, with the exception of GMED, the hypothesis that the muscle activity of the residual and intact legs would increase with speed was supported. However, there were differences in muscle activity when specific regions of the gait cycle were analyzed separately.

During braking, there was significantly higher activity in the residual leg VAS relative to the intact leg at all speeds ($p \leqslant 0.016$, Fig. 1). However, the largest difference in VAS activity between the residual and intact legs occurred at the slowest speeds: 0.101 ± 0.029 (residual-intact) at 0.6 m/s and 0.069 ± 0.020 at 0.9 m/s, compared to 0.037 ± 0.014 at 1.2 m/s and 0.065 ± 0.018 at 1.5 m/s (Fig. 1). While VAS activity increased with speed during braking, differences in the residual leg were not as consistent across speeds compared to the intact leg. During braking, the intact leg VAS had significant increases in activity between 0.6 and 0.9 m/s (0.057 ± 0.019, $p = 0.048$) and 0.9 and 1.2 m/s (0.092 ± 0.016, $p < 0.001$). In contrast, the residual leg VAS exhibited no significant increase with speed between 0.6 and 0.9 m/s (0.024 ± 0.018) and a significant increase in activity between 0.9 and 1.2 m/s (0.060 ± 0.019, $p = 0.031$). Thus, the VAS response to increasing walking speed differed between legs, and was not as consistent in the residual leg.

Similarly, during braking RF exhibited significantly higher activity in the residual leg than in the intact leg at 0.9, 1.2 and 1.5 m/s ($p \leqslant 0.019$, Fig. 1). The largest differences between the residual and intact legs occurred at 0.9 m/s (0.118 ± 0.036) and 1.2 m/s (0.127 ± 0.027), compared to the difference at 1.5 m/s (0.062 ± 0.025). Also, the intact leg RF experienced a significant increase in muscle activity between 1.2 and 1.5 m/s (0.089 ± 0.011, $p < 0.001$) while the residual leg RF exhibited no significant increase in activity between 1.2 and 1.5 m/s (0.024 ± 0.029). Intact leg RF activity increased at a greater rate at higher walking speeds and was the most reduced relative to the residual leg at 0.9 and 1.2 m/s, while residual leg RF activity was heightened at 0.9, 1.2 and 1.5 m/s and showed little change from 1.2 to 1.5 m/s (Figs. 1

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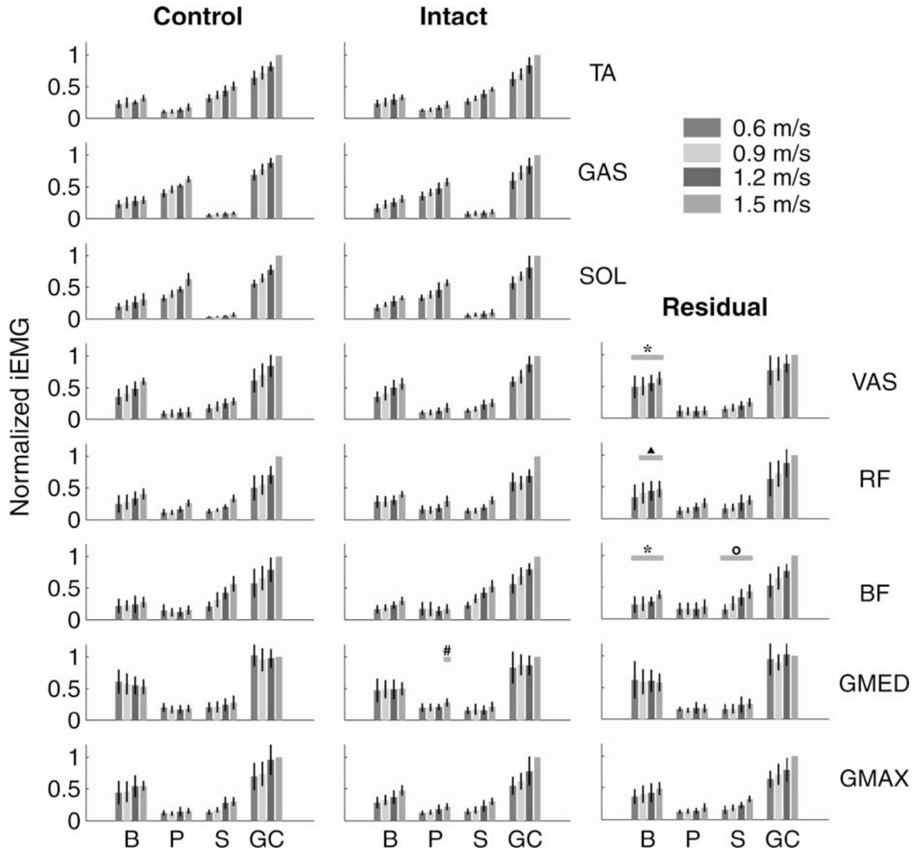

Fig. 1. Muscle integrated EMG (iEMG) magnitude across walking speeds within braking (B), propulsion (P), swing (S) and over the gait cycle (GC). Vertical lines indicate ±1 group standard deviation. Significant differences associated with hypotheses 1 and 2 are indicated as follows: residual leg magnitude is greater than the intact leg at all walking speeds (*), residual leg magnitude is less than the intact leg at all walking speeds (o), residual leg magnitude is greater than the intact leg at the three highest walking speeds (▲) and intact leg quantity is greater than the control leg at 1.5 m/s (#).

and 2). Thus, RF response to increased walking speed differed between the residual and intact legs during braking.

4. Discussion

The goal of this study was to examine EMG of both the intact and residual legs across a wide range of steady-state walking speeds to identify changes in muscle activity in below-knee amputees. The hypothesis that there would be no difference in muscle activity between the residual and intact legs was not supported in all muscles. Across all speeds, the residual leg BF and VAS muscles were significantly different than the intact leg. The residual leg BF had heightened activity during braking at all speeds compared to the intact leg, which acted to increase the positive hip work of these subjects in early stance (Silverman et al., 2008). This result is also consistent with increased residual leg hip power relative to non-amputees that has been reported during the first half of stance in other amputee studies (Winter and Sienko, 1988; Gitter et al., 1991; Bateni and Olney, 2002). Previous modeling and simulation studies of non-amputee walking have shown that BF has the potential to contribute positively to propulsion throughout stance (Neptune et al., 2004; Liu et al., 2006). Thus, increased residual leg BF activity in early stance would provide an increased positive contribution to the A/P GRF, and therefore reduce the net braking GRF compared to the intact and control legs in early stance. This result is consistent with the decreased braking impulse measured in these subjects as reported in Silverman et al. (2008). Heightened residual leg BF activity during braking has also been previously shown in amputee walking at moderate speeds (e.g., Perry and Shanfield, 1993; Powers et al., 1998; Isakov et al., 2000; Schmalz et al., 2001). Our results show this compensation by BF also occurs across a wide range of speeds.

We also found increased activity of the residual leg VAS compared to the intact leg at all speeds. Increased residual leg VAS activity from early to mid-stance has been previously shown at moderate speeds (e.g., Powers et al., 1998). Our results show the adaptation of VAS occurs across a wide range of speeds and appears necessary to provide additional body support in the absence of the ankle plantar flexors, consistent with its functional role found in modeling studies of non-amputee walking (Anderson and Pandy, 2003; Neptune et al., 2004). In addition to increased BF and VAS activity, heightened activity in the residual leg RF during braking also occurred at the higher speeds of 0.9, 1.2 and 1.5 m/s. RF has been shown to provide body support when active during braking in non-amputee walking (Neptune et al., 2004). Thus, RF appears to be working in synergy with VAS to provide additional body support.

The hypothesis that there would be no difference in muscle activity between the intact and control leg muscles was supported with the exception of GMED. During propulsion at 1.5 m/s, the intact leg GMED activity was significantly higher than the control leg (Fig. 1). However, during propulsion GMED is primarily inactive or has a very small magnitude (Figs. 1 and 2). In addition, the difference was only found at the highest walking speed. Thus, this difference does not appear to represent a major compensatory mechanism. The remaining muscles support the hypothesis that there will be no difference in activity between intact and control leg muscles. These results are consistent with previous studies that

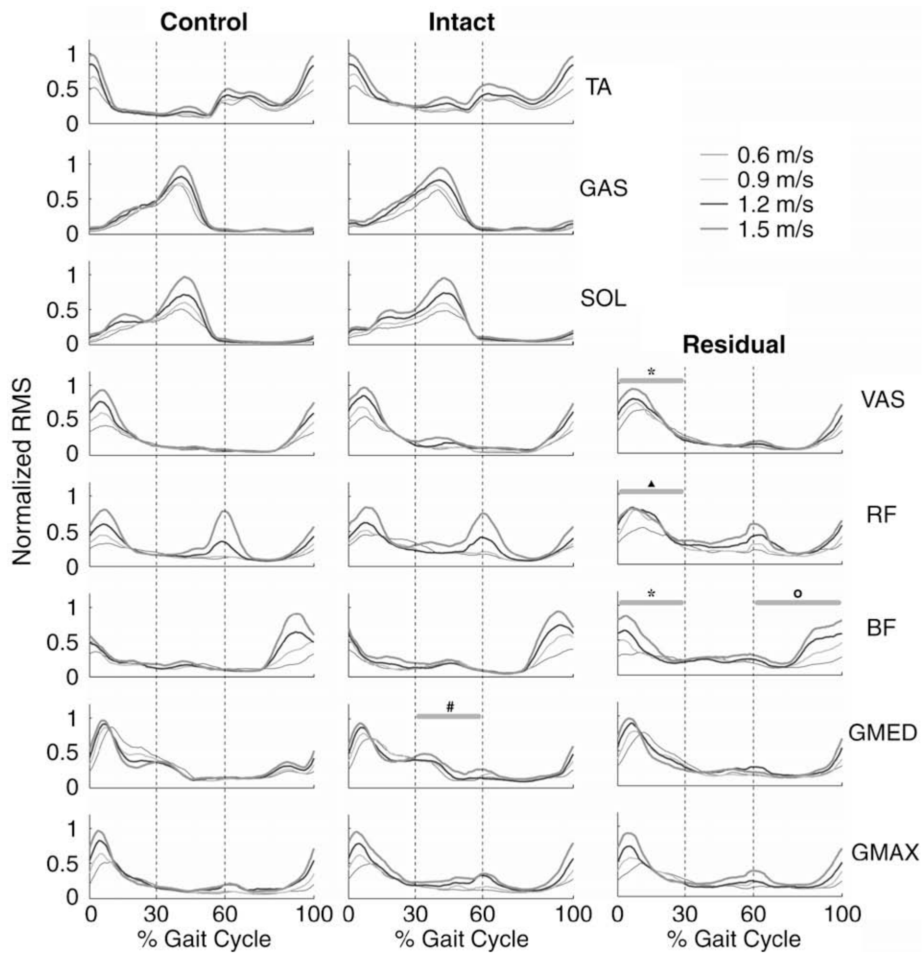

Fig. 2. Normalized root-mean-square (RMS) EMG patterns over the gait cycle across walking speeds. Braking, propulsion and swing correspond to ~0–30%, 30–60%, and 60–100% of the gait cycle, respectively. Significant differences associated with hypotheses 1 and 2 are indicated as follows: residual leg magnitude is greater than the intact leg at all walking speeds (*), residual leg magnitude is less than the intact leg at all walking speeds (o), residual leg magnitude is greater than the intact leg at the three highest walking speeds (▲) and intact leg quantity is greater than the control leg at 1.5 m/s (#).

found intact leg muscle activity to be the same as non-amputees while walking at moderate speeds (e.g., Culham et al., 1986; Czerniecki, 1996; Rietman et al., 2002), and our results show this remains over a wide range of walking speeds. However, Silverman et al. (2008) identified increased intact leg positive hip work over stance at walking speeds of 1.2 and 1.5 m/s in these subjects. Also, increased intact leg hip power has been reported during stance in other amputee studies (Bateni and Olney, 2002). Therefore, an intact leg compensatory mechanism may exist that was not detected by an adaptation in the muscle activity of these muscles, or it occurred in muscles not measured (e.g., adductor magnus).

The hypothesis that muscle activity in the residual and intact legs will increase in magnitude with increased steady-state walking speed was supported with the exception of GMED (Fig. 1). The insensitivity of GMED to walking speed is consistent with its primary role to provide body support from early to mid-stance (Anderson and Pandy, 2003). The increases in all other muscle activity with walking speed (Figs. 1 and 2) have also been observed in studies of non-amputee walking (Nilsson et al., 1985; Hof et al., 2002; den Otter et al., 2004; van Hedel et al., 2006). These studies have shown muscle activity, regardless of its method for quantification (peak magnitude, integrated area, mean value, or gain functions), increases in magnitude with increasing walking speed (Nilsson et al., 1985; den Otter et al., 2004; van Hedel et al., 2006). Although our analysis of muscle activity showed a walking speed effect, not all muscles showed consistent increases in activity in the residual and intact legs or specific regions in the gait cycle.

Our data showed increased activity in the residual leg VAS compared to the intact leg, with the difference being largest at the slower walking speeds (0.6 and 0.9 m/s, Fig. 1). The differences in VAS activity across speeds for each leg followed the same pattern, but at slower walking speeds the residual leg VAS was heightened to a greater extent compared to the intact and control legs during braking (Fig. 2), which resulted in fewer significant differences across speeds. Increased residual leg VAS activity from early to mid-stance has been previously shown in amputee walking at moderate speeds (e.g., Powers et al., 1998). Our results also show that VAS activity is heightened at slower walking speeds with a slower decay in magnitude during braking (Figs. 1 and 2). Thus, the residual leg VAS in early stance may provide additional body support in the absence of the ankle plantar flexors (Anderson and Pandy, 2003; Neptune et al., 2004), especially at slower speeds. Others have suggested that increased co-contraction of the residual leg BF and VAS muscles (both muscles showing heightened activity in early stance) is indicative of decreased stability due to the loss of the plantar flexors (e.g., Rietman et al., 2002). The co-contraction of these muscles appears heightened at slower walking speeds (Figs. 1 and 2), and may produce different effects as speed is modulated. A reduction in residual leg positive knee work relative to control subjects was found in these same subjects walking at 1.2 and 1.5 m/s (Silverman et al., 2008). Thus, at higher walking speeds

BF may act to reduce the net knee extensor moment in the residual leg during braking, while at slower walking speeds VAS may play a more prominent role to provide body support. A reduced residual leg knee extensor moment at the highest walking speeds would also be consistent with another study of amputee walking at 1.2 m/s and 1.6 m/s that reported the residual leg knee moment to remain flexor during braking (Sanderson and Martin, 1997).

We also found RF activity during braking to be different between the residual and intact legs. The intact leg RF activity was reduced compared to the residual leg at 0.9, 1.2 and 1.5 m/s, with the largest differences occurring at 0.9 and 1.2 m/s (Fig. 1). Also, intact leg RF activity experienced a significant increase from 1.2 to 1.5 m/s, while the residual leg RF did not. The results of the intact leg RF are consistent with previous non-amputee walking studies showing RF activity to increase at a greater rate at higher walking speeds, and show little to no activity at slower speeds (Yang and Winter, 1985; Hof et al., 2002; den Otter et al., 2004). As previously noted, the increased residual leg RF activity during braking is consistent with the need for additional body support in the absence of the plantar flexors (Neptune et al., 2004).

In summary, the EMG patterns of the residual and intact legs were similar and increased with speed. However, differences in several residual leg muscles were observed. Differences occurred in the residual leg BF, VAS and RF, which showed increased activity from early to mid-stance compared to the intact leg. These compensations were consistent with the need for additional body support and forward propulsion in the absence of the plantar flexors (Anderson and Pandy, 2003; Neptune et al., 2004; Liu et al., 2006). In addition, the compensation of the residual leg VAS appears to be heightened at slower walking speeds to provide needed lower limb stability and body support. With the exception of GMED during the propulsion phase at 1.5 m/s, all intact leg muscles showed similar muscle activation patterns compared to the control leg. Finally, GMED activity was shown to be insensitive to changes in walking speed over the gait cycle in the residual, intact and control legs, which is consistent with its primary role to provide body support (Anderson and Pandy, 2003). These results suggest that rehabilitation methods that improve the output from BF, VAS and RF in the residual leg may be an effective way to compensate for the lost ankle plantar flexors. Future work will be directed at identifying the influence of specific prosthetic design characteristics (e.g., foot-ankle stiffness) on these compensatory mechanisms and how prosthetic designs can be optimized to minimize the necessary changes in muscle activity.

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Nicholas P. Fey received his B.S. and M.S.E. degrees in mechanical engineering from The University of Texas at Austin in 2006 and 2008, respectively. He is currently working on his Ph.D. in mechanical engineering as a graduate research assistant in the Neuromuscular Biomechanics Laboratory at The University of Texas at Austin. His research interests include experimental and musculoskeletal modeling analyses of amputee gait, neural control of human movement, and prosthetic design optimization.

Anne K. Silverman received her B.S.E. degree in mechanical engineering from Arizona State University in 2005 and her M.S.E. degree in mechanical engineering from The University of Texas at Austin in 2007. She is currently working on her Ph.D. in mechanical engineering as a graduate research assistant in the Neuromuscular Biomechanics Laboratory at The University of Texas at Austin. Her research interests include experimental analysis of human locomotion, musculoskeletal modeling and simulation of amputee gait, and rehabilitation methods.

Richard R. Neptune received B.S., M.S. and Ph.D. degrees in mechanical engineering from the University of California, Davis, in 1991, 1993 and 1996, respectively. He completed a postdoctoral fellowship at the University of Calgary, Calgary, AB, Canada, from 1996 to 1998. He then served as a Biomedical Engineer at the VA Palo Alto Rehabilitation Research and Development Center in Palo Alto, CA from 1998 to 2000. He is currently an Associate Professor in the Department of Mechanical Engineering at The University of Texas at Austin. His research interests include musculoskeletal modeling and simulation of human movement, prosthetic and orthotic design optimization, neuromotor control of movement and sports biomechanics.