BIOMECHANICAL AND METABOLIC EFFECTS OF USING PASSIVE AND POWERED ANKLE-FOOT PROSTHESES FOR SLOPED WALKING IN PERSONS WITH A TRANSTIBIAL AMPUTATION

by

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Biomechanical and Metabolic Effects of Using Passive and Powered Ankle-Foot Prostheses for Sloped Walking in Persons with a Transtibial Amputation

Thesis directed by Assistant Professor Alena M. Grabowski

People with a transtibial (below-knee) amputation walking on level ground with a passive-elastic prosthesis typically experience a 10-30% greater metabolic demand than non-amputees walking at the same speed. Battery-powered ankle-foot prostheses that provide push-off power to the user have been developed that normalize the metabolic demand and biomechanics for people with a leg amputation walking on level ground. Because ramps and slopes are encountered regularly, understanding how passive and powered prostheses affect walking energetics and biomechanics for individuals with a leg amputation on a variety of slopes is critical to designing robust ankle-foot prostheses. In this dissertation, I first sought to understand the role of the biological ankle during sloped walking at various speeds. To do this, I documented the correlation between metabolic power and individual leg mechanical power at slopes of 0°, ±3°, ±6°, and ±9° and walking speeds of 1.00, 1.25, and 1.50 m/s. Then, I determined the contribution of the ankle joint to individual leg positive and negative work. Next, I recruited 10 people with a unilateral transtibial amputation to walk 1.25 m/s on slopes of 0°, ±3°, ±6°, and ±9° with passive-elastic and powered ankle-foot prostheses. I collected metabolic energy expenditure, kinetic, and kinematic data. I tuned the powered prosthesis such that prosthetic ankle range of motion, peak moment, peak power, and net work matched biological ankle (or intact ankle of the subject) values at each slope. Use of the powered prosthesis significantly reduced net metabolic power required to walk up a +3° and +6° slope by 5% compared to a passive-elastic prosthesis. Using the powered prosthesis did not significantly alter affected or unaffected individual leg step-to-step transition, hip, or knee work compared to use of a passive-elastic prosthesis. However, prosthetic ankle positive work increased at all uphill slopes and prosthetic ankle net work was more positive at all slopes with use of the powered prosthesis compared to use of a passive-elastic prosthesis. My results indicate that use of powered ankle-foot prostheses can improve walking energetics and biomechanics at uphill slopes and thus quality of life for people with a leg amputation.

DEDICATION

This thesis is dedicated to Simon Montgomery, without whom I would have much less love, puppies, and food in my life. Thank you for your support and encouragement throughout this process.

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I would like to thank my family and friends for encouraging me to pursue my dream of getting this degree and studying locomotion biomechanics. Perhaps the greatest gratitude is to my advisor, Dr. Alena Grabowski, who has taught me more than I could imagine about research, writing, practicing patience, and being a good scientist. Dr. Grabowski has influenced my personal and professional life in many ways and I have the utmost gratitude for her guidance and patience with me over the last few years. Additionally, I would like to thank each of my committee members for giving me their valuable time, advice, and insights. Finally, I would like to thank the Department of Veterans Affairs, for providing the funding for this research.

TABLE OF CONTENTS

LIST OF TABLES

LIST OF FIGURES

1 INTRODUCTION

Level-ground walking can be mechanically characterized by a stance phase when only one foot is in contact with the ground, a step-to-step transition phase when both feet are in contact with the ground, and a leg swing phase when the foot is off the ground and advanced forward into the next step. Previous studies have characterized the biomechanics of the body during the single leg support (stance) phase using an inverted pendulum model, where the body's center of mass (COM) is represented by a point mass, and the leg by a rigid, massless strut [1-4]. The inverted pendulum model suggests that minimal external mechanical work is necessary for the body to sustain steady-speed level-ground walking during the stance phase, because there is a constant phasic exchange of gravitational potential and kinetic energy [1, 3]. However, during the step-to-step transition phase of level-ground walking (Figure 1.1) when both feet are on the ground, the leading leg absorbs mechanical work to slow the downward velocity of the COM while the trailing leg simultaneously produces mechanical work to redirect the COM velocity upward and forward [5-7]. Previous studies have quantified the external mechanical work done by each leg during the step-to-step transition and found that individual leg work to accelerate and redirect the COM constitutes the majority of the external mechanical work done during walking [1, 2, 4, 5, 8, 9].

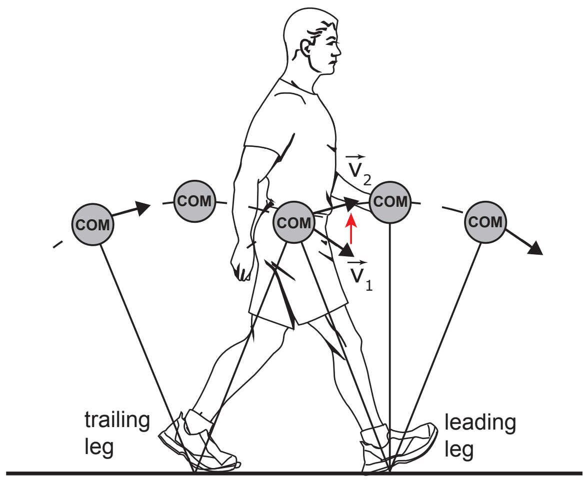

Figure 1.1. During the single leg stance phase of walking, the body's mechanics are modeled as an inverted pendulum, where the COM moves along the arc of the pendulum (dashed black lines). During the step-to-step transition, the COM velocity is moving downward and forward (V_1). The leading leg slows the progression of the COM by absorbing work as the trailing leg redirects and accelerates the COM upward and forward (V_2). The red arrow shows the change in direction of the COM during the step-to-step transition. Figure from Jeffers et al. 2015.

The work done by the legs on the COM can be broken down into the joint contributions from the muscles acting at the ankle, knee, and hip joints. On level ground, the muscles acting at the ankle and hip account for approximately 41% and 39%, respectively, of individual limb total positive power while walking at various speeds, while the knee accounts for approximately 20% [10]. The knee extensor muscles mostly absorb energy, or results in negative mechanical work or power, while the muscles at the hip and ankle provide both positive and negative work and power during the stance phase of walking. The hip muscles provide positive power at the beginning of stance to extend the leg and raise the COM to its highest point at mid-stance [11, 12]. The ankle provides push-off power at the end of the stance phase to transition the leg into the next step and to redirect and accelerate the COM [11-13]. Because the ankle provides its power in late stance during the step-to-step transition, populations who have little or no ankle function (e.g. people with an amputation, aging populations, stroke sufferers etc.) require the contralateral limb or more proximal joints to compensate for the lost power [14, 15]. Ankle-foot prostheses have been developed that provide normative (similar to biological) stance phase ankle power to the user to reduce the compensation from the contralateral limb or proximal muscles [16-21].

Each mechanical task involved in walking requires the muscles to contract and thus consumes metabolic energy. To determine the portion of the net metabolic power required to perform two of these tasks – supporting body weight and redirecting and accelerating the COM into the next step – Grabowski et al. manipulated subjects' weight and mass independently during level treadmill walking [22]. During the step-to-step transition, the portion of net metabolic power required to redirect the COM during level-ground walking for healthy adults comprises approximately 45%, whereas the portion of net metabolic power required to support body weight in single limb support is approximately 28% [22]. Further, the entirety of the 45% of net metabolic power needed for the step-to-step transition in level-ground walking is thought to be due to providing push-off power with the trailing leg [23].

Beyond level-ground walking, experiments have quantified metabolic energy expenditure during walking up or downhill at constant speeds. Walking at moderate slopes of 3° and 6°, net metabolic power per unit body mass increases 52% and 113%, respectively [24]. At extreme slopes of approximately 26° uphill and downhill, the metabolic cost of transport (COT) is significantly higher than on level ground [25, 26]. Interestingly, the minimum COT during walking at 1.00 m/s occurs at approximately -6° (-0.1 rad) and is higher at steeper uphill or downhill slopes (Figure 1.2) [25, 26].

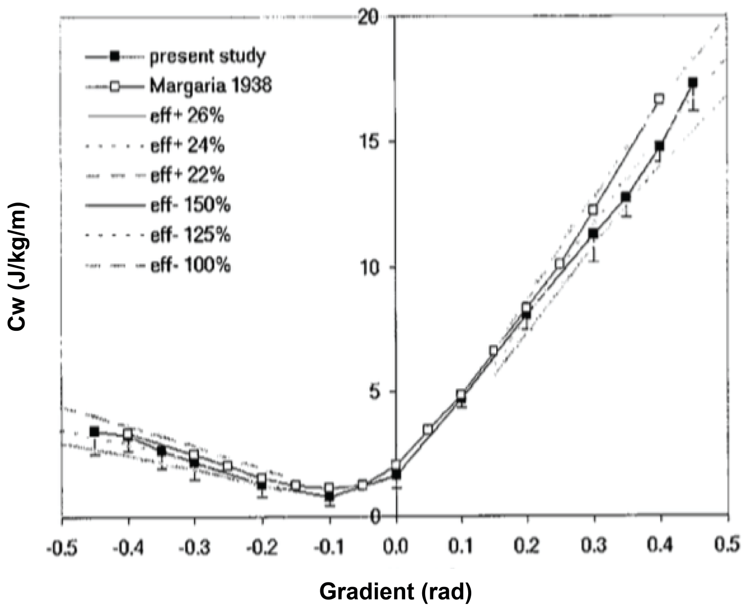

Figure 1.2. Metabolic COT (Cw) versus gradient of ten healthy adult males. Also plotted are data from [25]. Eff refers to the mechanical efficiency which was calculated as the ratio of mechanical work rate to raise or lower the COM divided by the metabolic rate. Figure adapted from [26].

When a person undergoes a lower-limb amputation, he/she is typically fitted with a passive elastic prosthesis, also known as an "energy storage and return" (ESAR) foot. These feet are made of a carbon fiber keel and return approximately 51% of the mechanical energy stored during the first part of the stance phase [15, 27]. ESAR prostheses are completely passive, meaning they are unable to generate mechanical power anew. They also do not contain a prosthetic "ankle" articulation. In contrast, powered prostheses are commercially available that have an on-board battery and motor and are able to generate mechanical power and supply that to the user during the stance phase of walking [16, 18, 19, 28-30]. These powered ankle-foot prostheses [31, 32] include a single degree of freedom articulation (dorsi- and plantar-flexion) at the prosthetic ankle, and are of similar size and weight to a biological foot/shank segment for the average adult male. These prostheses are controlled via on-board microprocessors and mimic biological ankle function.

A groundbreaking 2012 study [17] revealed that with the use of a powered ankle-foot prosthesis, people with a unilateral transtibial amputation were able to walk with similar biomechanics and metabolic cost compared to non-amputees over a range of speeds. Compared to walking with a passive-elastic prosthesis, walking with the powered prosthesis resulted in significantly greater positive work done at the ankle for the trailing leg during walking [17]. Specifically, subjects exhibited 38% less trailing leg work and 27% greater collision work in the leading leg when using an ESAR compared to powered prosthesis [17]. Knee adduction moments – those considered to lead to knee osteoarthritis are also significantly lower with use of a powered compared to ESAR prosthesis [33]. Perhaps a more clinically relevant measure is that of preferred walking speed. When walking with a bionic, powered prosthesis, subjects preferred to walk 23% faster than when using a passive-elastic prosthesis [17]. This and other studies have shown that with the use of a passive-elastic prosthesis, asymmetry in gait parameters such as ground reaction forces and leg work increase [34-38].

The effects of transtibial amputation on the biomechanics and metabolic costs of level-ground walking using different prostheses have been documented in the literature. However, studies are needed that compare biomechanics, metabolic COT, and muscle activity of the same subjects walking over various terrain while using different prostheses. Such studies will likely provide insight and valuable information about prosthetic design and prescription to researchers and clinicians.

The present dissertation will address: 1) the correlation between metabolic and mechanical power over a range of walking speeds and slopes in non-amputees, 2) the contribution of the ankle joint to individual leg work during sloped walking at a range of speeds in non-amputees, 3) the effects of using passive and powered ankle-foot prostheses during sloped walking on metabolic power and individual leg step-to-step transition work, and 4) the effects of using passive and powered ankle-foot prostheses during sloped walking on leg joint mechanics.

2 CHAPTER 1: THE CORRELATION BETWEEN METABOLIC AND INDIVIDUAL LEG MECHANICAL POWER DURING WALKING AT DIFFERENT SLOPES AND VELOCITIES

2.1 Abstract

During level-ground walking, mechanical work from each leg is required to redirect and accelerate the center of mass. Previous studies report a linear correlation between net metabolic power and the rate of step-to-step transition work during level-ground walking with different imposed step lengths. However, correlations between metabolic power and individual leg power during step-to-step transitions while walking on uphill/downhill slopes and at different velocities are not known. This understanding of these relationships between metabolic demands and biomechanical tasks can provide important information for design and control of biomimetic assistive devices such as leg prostheses and orthoses. Thus, we compared changes in metabolic power and mechanical power during step-to-step transitions while nineteen subjects walked at seven slopes (0^o, +/-3^o, +/-6^o, & +/-9^o) and three velocities (1.00, 1.25, & 1.50m/s). A quadratic model explained more of the variance (R^2 = 0.58-0.61) than a linear model (R^2 = 0.37-0.52) between metabolic power and individual leg mechanical power during step-to-step transitions across all velocities. A quadratic model explained more of the variance (R^2 = 0.57-0.76) than a linear model (R^2 = 0.52-0.59) between metabolic power and individual leg mechanical power during step-to-step transitions at each velocity for all slopes, and explained more of the variance (R^2 = 0.12-0.54) than a linear model (R^2 = 0.07-0.49) at each slope for all velocities. Our results suggest it is important to consider the mechanical function of each leg in the design of biomimetic assistive devices aimed at reducing metabolic costs when walking at different slopes and velocities.

2.2 Introduction

To walk, humans utilize metabolic energy to perform mechanical tasks such as generating force to support weight and performing work to redirect/accelerate the center of mass (COM) from step-to-step. COM dynamics have been well represented by an inverted pendulum model during single leg support for level-ground walking at constant velocities [39, 40]. This model suggests minimal mechanical work required to sustain steady-speed locomotion because of a constant phasic exchange of potential and kinetic energy [39, 41]. However, during step-to-step transitions, the leading leg absorbs mechanical work to slow downward movement of the COM while the trailing leg generates mechanical work to redirect the COM upward and forward [5-7, 42].

Individual leg mechanical power during step-to-step transitions changes with velocity over level-ground [5, 6, 43, 44], such that the leading leg absorbs and trailing leg generates work simultaneously during step-to-step transitions. However, both legs generate work while walking on uphill slopes and absorb work when walking on downhill slopes at 1.25 m/s [43]. Similarly, metabolic power increases with uphill slopes greater than and decreases with downhill slopes less than -3^{o} [26, 45-47]. However, step-to-step transition work and metabolic demands for the combined effects of slope and velocity have not been examined.

Previous research has shown strong correlations between individual leg power and mechanical metabolic power performed during step-to-step transitions for level-ground walking at 0.72-1.97 m/s with different imposed step lengths [44]. Donelan et al. found that when step length is varied, 79-89% of the variance in metabolic power is explained by individual leg mechanical power during the step-to-step transition [44]. This strong correlation between metabolic and individual leg step-to-step transition power during level-ground walking suggests a correlation may exist when walking uphill and downhill at different velocities. However, these correlations have not been established. Further, because previous research has shown that the ankle accounts for 46-89% of the external power required for level-ground walking [10, 48], understanding the correlations between metabolic and mechanical power is important for design, development, and control of robust biomimetic assistive devices such as leg prostheses and orthoses [49]. Previous studies suggest that prosthetic ankle power plays an important role in reducing metabolic demands during level-ground walking [17, 50].

Our purpose was to determine the correlations between individual leg step-to-step transition mechanical power and metabolic power during walking across a wide range of slopes and velocities. We sought to better understand the basic biomechanics and metabolic costs of unimpaired human walking. Metabolic power increases and leading (P_{lead}) and trailing (P_{trail}) leg mechanical powers are more positive on steeper uphill slopes compared to level-ground walking [51]. Metabolic power decreases, and P_{lead} and P_{trail} are more negative on steeper downhill slopes compared to level-ground walking [51]. Further, metabolic power increases, P_{lead} is more negative and P_{trail} is more positive at faster velocities [26, 43, 51]. We hypothesized that metabolic power would be correlated with P_{lead} and P_{trail} for all slopes (-9° to 9°) and velocities (1.00 m/s, 1.25 m/s and 1.50 m/s). We also hypothesized that the ratio of individual leg mechanical power during step-to-step transitions to the overall metabolic power, indicated as the individual limb power ratio (ILPR), would be similar across all slopes and velocities.

2.3 Methods

Nineteen subjects with no lower extremity or neurological injuries or pathologies volunteered [13 M, 6 F, mean [52]; 29.2 years (8.4 years); 69.6 kg (13.2 kg)] and gave informed written consent prior to participating in accordance with a protocol approved by the Department of Veteran Affairs' Human Subjects Institutional Review Board. Subjects walked on a dual-belt force-measuring treadmill (Bertec Corp., Columbus, OH) at seven slopes (0^o, +/-3^o, +/-6^o, & +/-9^o) and three velocities (1.00, 1.25, & 1.50 m/s). We used constant speeds for comparisons across conditions and with other studies [43]. First, we measured each subject's body mass and metabolic rate while standing quietly. Then, we measured metabolic rates and ground reaction forces during each six-minute walking trial. Trial order was randomized, and at least two minutes rest was enforced between trials. Data collection occurred over three days at the same time of day to reduce day-to-day variability in metabolic rates. Seven walking conditions were tested each day.

2.3.1 Metabolic Power

We measured rates of oxygen consumption ($\dot{V}O_2$; ml/min/kg) and carbon dioxide production ($\dot{V}CO_2$; ml/min/kg) using indirect calorimetry (Parvo Medics TrueOne 2400, Sandy, UT). We averaged $\dot{V}O_2$ and $\dot{V}CO_2$ from the last two minutes of each trial and calculated metabolic power using a standard equation [53]. We determined net metabolic power by subtracting standing from each trial's metabolic power.

2.3.2 Step-to-Step Transition Power

We measured ground reaction forces (*F*) at 1500 Hz from each leg and normalized all data to body mass (*m*). We filtered *F* with a fourth order low-pass Butterworth filter and 20 Hz cutoff frequency using a custom program (Matlab, Natick, MA). COM acceleration (*a*) with respect to time (*t*) was calculated as:

$$a_{ML}(t) = \frac{F_{ML}(t)}{m} \tag{2.1}$$

$$a_{parallel}(t) = \frac{F_{parallel}(t) - mg * \sin(\Theta)}{m} \tag{2.2}$$

$$a_{perp}(t) = \frac{F_{perp}(t) - mg * \cos(\Theta)}{m} \tag{2.3}$$

where medio-lateral (*ML*), parallel (*parallel*), and perpendicular (*perp*) components of force were calculated relative to the treadmill slope (Θ). COM velocity (*v*) was calculated as the integral of acceleration with respect to time:

$$v(t) = \int_0^t a(t)dt + v_o \tag{2.4}$$

We determined integration constants (v_o) for perpendicular (v_{perp}) and medio-lateral (v_{ML}) velocities by assuming that average *v* over a stride equaled zero. We determined v_o for parallel velocity ($v_{parallel}$) by assuming that average *v* over a stride equaled treadmill velocity. We calculated external mechanical power performed by each leg during step-to-step transitions using the method described by Donelan et al. [44]. We calculated step-to-step transition power absorbed and generated by each leg (P_{lead} and P_{trail}) on the COM as the sum of the dot products of ground reaction force (F) and COM velocity (v_{com}) during the step-to-step transition:

$$P_{lead} = F_{ML,lead} \cdot v_{ML,com} + F_{parallel,lead} \cdot v_{parallel,com} + F_{perp,lead} \cdot v_{perp,com} \tag{2.5}$$

$$P_{trail} = F_{ML,trail} \cdot v_{ML,com} + F_{parallel,trail} \cdot v_{parallel,com} + F_{perp,trail} \cdot v_{perp,com} \tag{2.6}$$

We defined step-to-step transition as the time when both legs were on the ground [44]. We detected heel-strike and toe-off with a force threshold of twice the average signal noise when nothing was in contact with the treadmill. Step-to-step transitions may extend beyond double support [5], but our primary interest was to understand correlations between metabolic power and mechanical power absorbed at initial contact and produced during late stance, thus we defined the step-to-step transition as the time of double support. We calculated each subject's average individual leg powers during step-to-step transitions from approximately 75 steps per subject. Then, we calculated the average and standard deviation of individual leg power from all subjects.

2.3.3 Relationship between Individual Leg Mechanical Power and Metabolic Power

Based on research by Donelan et al. [44], we quantified correlations between metabolic and individual leg mechanical power using linear models. Because leg muscles must produce force to move the COM forward during walking and thereby incur a metabolic cost, musculoskeletal models have optimized muscle activation squared to predict biomechanical data for a walking gait cycle [54-56] and approximate metabolic cost [57]. Based on this research, we also investigated quadratic models when correlating metabolic power and individual leg mechanical power. Further, we analyzed data for all slopes at each speed based on work by Farris and Sawicki [10] that demonstrated a quadratic change in individual leg work with faster walking speed.

2.3.4 Individual Limb Power Ratio (ILPR)

We quantified ILPR as the ratio between average individual leg mechanical power during step-to-step transitions and average metabolic power for the entire walking task. We acknowledge that other mechanical tasks account for the metabolic power required during the entire gait cycle, but sought to specifically understand how relative changes in step-to-step transition power affected metabolic power at different slopes and velocities. As such, ILPR shows the effect of each leg's step-to-step transition power on overall metabolic power, but underestimates ILPR for the entire walking task.

$$ILPR_{lead} = {}^{P_{mech,lead,S2S}}\!/_{P_{met}} \tag{2.7}$$

$$ILPR_{trail} = {}^{P_{mech,trail,S2S}}\!/_{P_{met}} \tag{2.8}$$

2.3.5 Statistics

We used two-way repeated measures ANOVAs (P<0.05) to compare metabolic power and step-to-step transition power of each leg at different slopes and velocities. Our within subject variables were slope with seven levels: 0^o, $\pm 3^o$, $\pm 6^o$, and $\pm 9^o$ and velocity with three levels: 1.00, 1.25, and 1.50 m/s. We correlated average metabolic power with average step-to-step transition power of each leg using linear and quadratic curve fitting regression models (RStudio, Boston, MA), where the strongest correlation delineated the best model.

2.4 Results

2.4.1 *Metabolic Power*

Average metabolic power was statistically different between all slopes except between -6° and -3° ($F(6,114) = 1161.69$, $P<0.001$). Greater metabolic power was required to walk with increasing slopes from -3° to 9° and decreasing slopes from -6° to -9° (Figure 2.1). Averaged across all three velocities, metabolic power was 20% greater at -9° compared to -6° (1.97 W/kg to 1.63 W/kg) and over five times greater at 9° compared to -3° (10.38 W/kg to 1.69 W/kg). On all slopes, metabolic power was statistically different between all velocities ($F(2,38) = 734.00$, $P<0.001$). We found a significant interaction effect between slope and velocity ($F(12,228) = 109.527$, $P<0.001$). At 1.00 and 1.25 m/s, metabolic power was statistically different for each slope, except for -6° and -3° (Figure 2.1). At 1.50 m/s metabolic power was statistically different for each slope, except -9° and -6° compared to -3°.

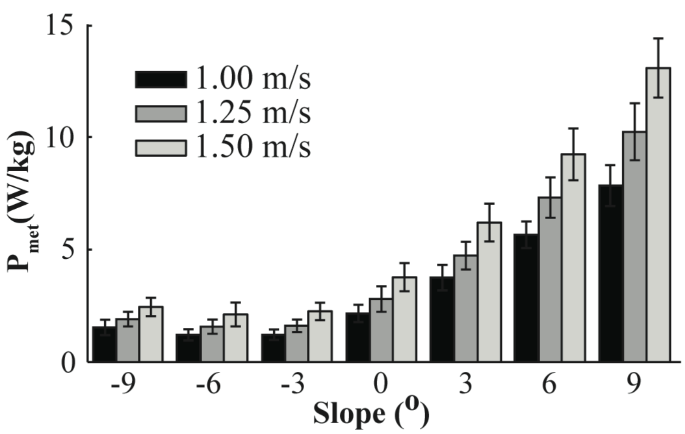

Figure 2.1. Average (S.D.) metabolic power (P_{met}) for seven slopes and three velocities (n = 19). In general, P_{met} increased at slopes greater or less than -3° and increased with velocity within each slope condition. Specific pairwise comparisons are described in the results section.

2.4.2 *Step-to-Step Transition Power*

Total leading leg step-to-step transition power (P_{lead}) decreased significantly with decreasing slopes and faster velocities ($F(6,114) = 186.34$, $P<0.001$; $F(2,38) = 261.01$, $P<0.001$; Figure

2.2b). P_{lead} from all velocities decreased by 118% from 9^o to -9^o (0.76 W/kg to -4.31 W/kg). P_{lead} from all slopes was over two-fold more negative at 1.50 m/s compared to 1.00 m/s (-2.11 W/kg to -0.66 W/kg). There was also a significant interaction effect between velocity and slope ($F(12,228) = 8.75$, $P<0.001$). At slopes of -9^o and -6^o, P_{lead} was more negative between 1.00 and 1.50 m/s and between 1.25 and 1.50 m/s. At -3^o, P_{lead} was more negative with faster velocities and all velocities were significantly different from each other. At 1.00 m/s, P_{lead} was more positive with increasing slopes between -9^o and 0^o, 3^o, 6^o, 9^o, between -6^o and 0^o, 3^o, 6^o, 9^o, between -3^o and 3^o, 6^o, 9^o, between 0^o and 6^o, 9^o, and between 3^o and 9^o. At 1.25 m/s, P_{lead} was more positive with steeper uphill slopes, except between 3^o and 6^o. At 1.50 m/s, P_{lead} was more positive with steeper uphill slopes, except between 0^o and 3^o, 3^o and 6^o, and 6^o and 9^o.

Trailing leg step-to-step transition power (P_{trail}) increased significantly with increasing slopes and faster velocities ($F(6,114) = 311.15$, $P<0.001$; $F(2,38) = 330.19$, $P<0.001$; Figure 2.2a). P_{trail} from all velocities increased nine-fold from -9^o to 9^o (-0.31 W/kg to 2.58 W/kg). P_{trail} from all slopes was 125% greater at 1.50 m/s compared to 1.00 m/s (1.80 W/kg to 0.86 W/kg). At each slope from -6^o to 9^o, we found a significant interaction effect for P_{trail} between velocity and slope ($F(12,228) = 4.898$, $P<0.001$). P_{trail} increased significantly with faster velocities, except for between 1.00 and 1.25 m/s at -6^o and between 1.25 and 1.50 m/s at 9^o. P_{trail} did not change with velocity at -9^o. More specifically, at 1.00 m/s, P_{trail} increased with slope, except between -9^o and -6^o, -3^o and 0^o, 0^o and 3^o, 3^o and 6^o, and 6^o and 9^o. At 1.25 m/s, P_{trail} increased with slope, except between 3^o and 6^o, and 6^o and 9^o. At 1.50 m/s, P_{trail} increased with slope, except between 0^o and 3^o, 0^o and 9^o, 3^o and 6^o, and 6^o and 9^o.

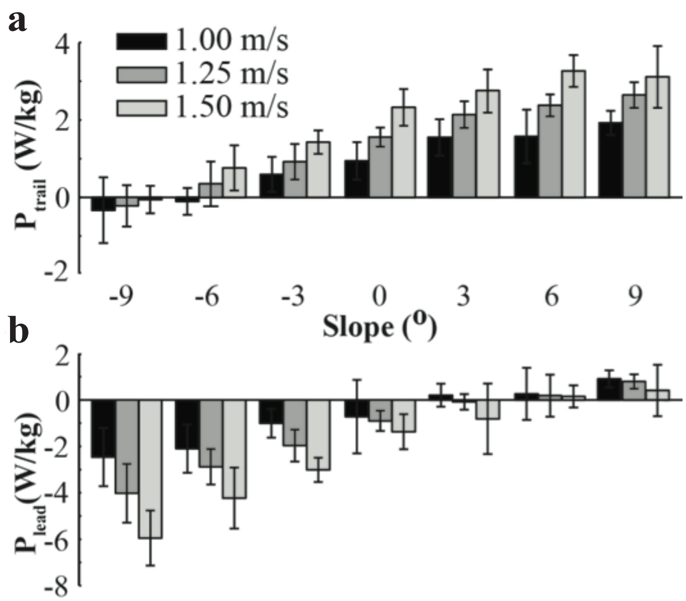

Figure 2.2. Average (S.D.) mechanical work of individual legs during step-to-step transitions. a) Mechanical work of the trailing leg (P_{trail}) during the step-to-step transition at each slope and velocity. P_{trail} was more positive with increasing slope from -9^o to 9^o and at faster velocities. b) Mechanical work of the leading leg (P_{lead}) during the step-to-step transition at each slope and velocity condition. P_{lead} was more positive with increasing slope from -9^o to 9^o and at slower velocities.

2.4.3 Relationship between Metabolic Power and Step-to-Step Transition Power

We fit our data with linear and quadratic models and found that 37% and 52%, respectively, of the variance in metabolic power was explained by P_{lead}. P_{trail} explained 58% and 61% of the variance in metabolic power with linear and quadratic models, respectively. Specifically, at each velocity across all slopes we found significant correlations between metabolic power and P_{lead} and P_{trail} using linear and quadratic models (Figure 2.3, $R^2 \geq 0.52$). At each velocity across all slopes, a quadratic model best described the relationship between overall metabolic power and P_{lead}; $R^2 = 0.76$, 0.72, and 0.71 for 1.00, 1.25, and 1.50 m/s, respectively and overall metabolic power and P_{trail}; $R^2 = 0.65$, 0.65, and 0.57 for 1.00, 1.25, and 1.50 m/s, respectively (Table 2.1). At each slope across all velocities, a quadratic model best described the relationship between overall metabolic power and P_{lead} and overall metabolic power and P_{trail} ($R^2 = 0.12$-0.54, Table

2.2). The strongest correlations were between P_{lead} and metabolic power, and P_{trail} and metabolic power at each velocity across all slopes. Thus, subsequent analyses were performed at each velocity across all slopes.

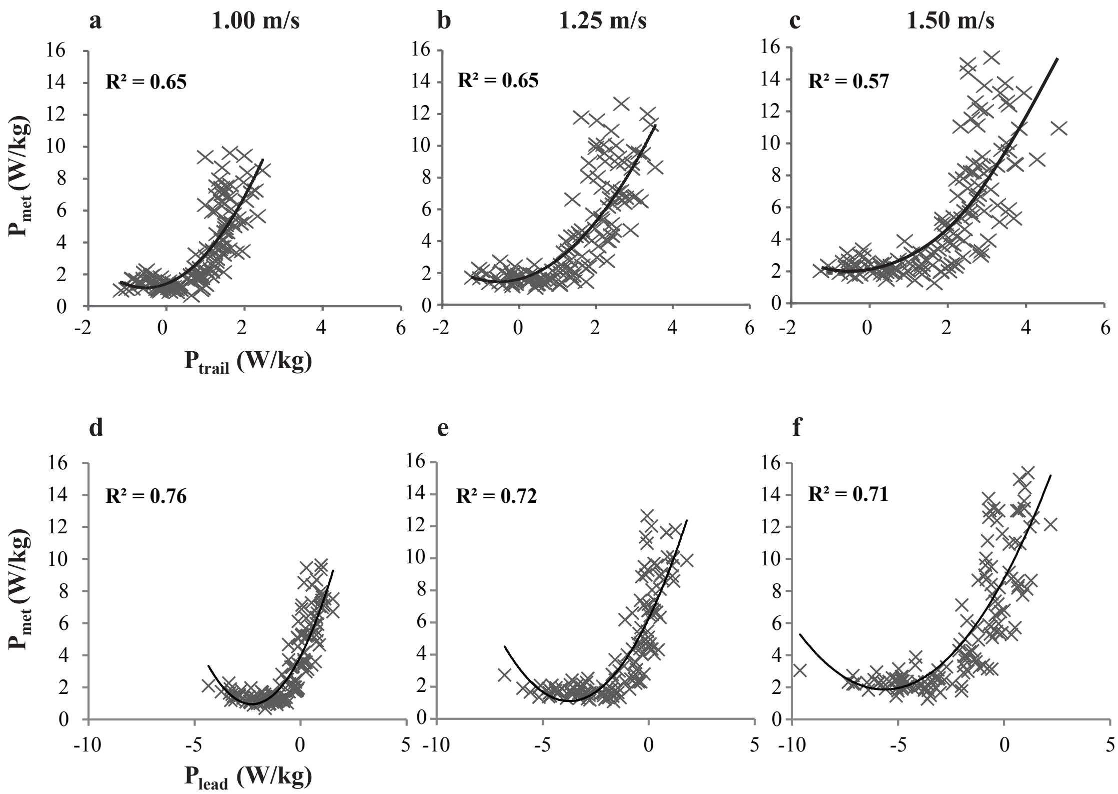

Figure 2.3. Relationship between metabolic power (P_{met}) and mechanical power of the individual legs during the step-to-step transition phase. For both the trailing leg (P_{trail}) (**A-C**) and leading leg (P_{lead}) (**D-F**), data are presented for all slopes for all subjects at each velocity tested. We found that a quadratic model best described the correlation between P_{met} and P_{trail} for all three velocities (R^2 = 0.57 to 0.65) and the correlation between P_{met} and P_{lead} for all three velocities (R^2 = 0.71 to 0.76).

Table 2.1. Linear and quadratic models describe the correlation between metabolic power (P_{met}) and trailing (P_{trail}) or leading (P_{lead}) leg mechanical power. All models yielded significant correlations (p<0.05). We determined the best model from the highest R^2 values (indicated in **bold**). * indicates p<0.01

	P_{met}* vs. *P_{trail}			***P_{met}* vs. *P_{lead}***		
	1.00 m/s	**1.25 m/s**	**1.50 m/s**	**1.00 m/s**	**1.25 m/s**	**1.50 m/s**
Linear	0.58*	0.59*	0.52*	0.56*	0.54*	0.58*
Quadratic	**0.65***	**0.65***	**0.57***	**0.76***	**0.72***	**0.71***

Table 2.2. Linear and quadratic models describe the correlation between metabolic power (P_{met}) and trailing (P_{trail}) or leading (P_{lead}) leg mechanical power. We determined the best model from the highest R^2 values (indicated in **bold**). $^{\#}$ indicates $p<0.05$, * indicates $p<0.01$

		-9°	-6°	-3°	0°	3°	6°	9°
P_{met} vs. P_{trail}	**Linear**	0.09$^{\#}$	0.27*	0.38*	0.41*	0.48*	0.39*	0.33*
	Quadratic	**0.09**	**0.28***	**0.38***	**0.46***	**0.54***	**0.40***	**0.39***
P_{met} vs. P_{lead}	**Linear**	0.35*	0.41*	0.47*	0.49*	0.25*	0.15*	0.07$^{\#}$
	Quadratic	**0.36***	**0.41***	**0.50***	**0.53***	**0.25***	**0.20***	**0.12**$^{\#}$

ILPR describes the extent to which total metabolic power was affected by each leg's mechanical power during step-to-step transitions (Figure 2.4). At all velocities, trailing leg ILPR ($ILPR_{trail}$) decreased by 33-41% during uphill compared to level-ground walking (Figure 2.4a) and by 32-40%, during downhill compared to level-ground walking (Figure 2.4a). At all slopes, there were no differences in $ILPR_{trail}$ between velocities (Figure 2.4a). $ILPR_{trail}$ was maximized at 0^o and decreased at slopes greater or less than 0^o at 1.00 m/s and 1.25 m/s. At 1.50 m/s, $ILPR_{trail}$ was maximized at -3^o. Quadratic regression models best described $ILPR_{trail}$ at each velocity across all slopes (R^2 = 0.21, 0.71, and 0.78 for 1.00 m/s, 1.25 m/s, and 1.50 m/s, respectively). At all velocities, leading leg ILPR ($ILPR_{lead}$) decreased by 51-82% during uphill compared to level-ground walking. On downhill slopes, $ILPR_{lead}$ increased four- to seven-fold compared to level-ground walking (Figure 2.4b). There were no differences in $ILPR_{lead}$ between velocities at all slopes (Figure 2.4b). Exponential regression models best described the $ILPR_{lead}$ data with R^2 = 0.71, 0.78, and 0.77 for 1.00 m/s, 1.25 m/s, and 1.50 m/s, respectively.

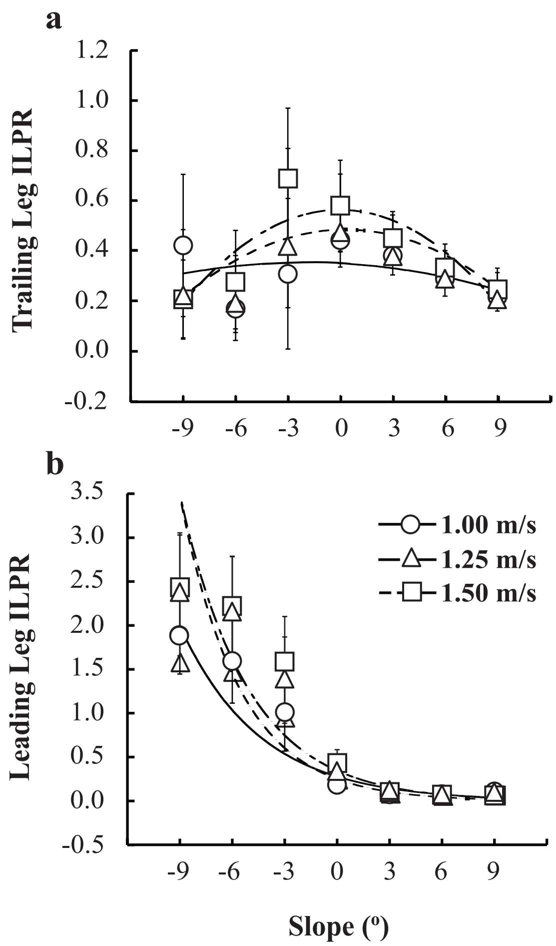

Figure 2.4. Average (S.D.) Individual Limb Power Ratio (ILPR) of the trailing and leading legs during step-to-step transitions at each slope and velocity. ILPR is calculated as the ratio of individual leg mechanical power during the step-to-step transition and overall metabolic power. **a**) Trailing leg ILPR was maximized at 0^o and decreased at slopes greater or less than 0^o at 1.00 m/s and 1.25 m/s. At 1.50 m/s, trailing leg ILPR was maximized at -3^o. Regression models were ILPR = $-0.0009s^2-0.0036s+0.3506$ at 1.00 m/s (R^2 = 0.21), ILPR = $-0.0031s^2+0.0019s+0.4851$ at 1.25 m/s (R^2 = 0.72), and ILPR = $-0.0046s^2-0.0002s+0.5642$ at 1.50 m/s (R^2 = 0.78), where s is slope (deg). **b**) Leading leg ILPR increased up to six-fold and decreased by up to 73% at -9^0 and 9^o, respectively, compared to 0^0. There were no differences in ILPR due to changes in velocity. Regression models were ILPR = $0.2783e^{-0.219s}$ at 1.00 m/s (R^2 = 0.71), ILPR = $0.2511e^{-0.292s}$ at 1.25 m/s (R^2 = 0.78), and ILPR = $0.3458e^{-0.257s}$ at 1.50 m/s (R^2 = 0.77).

2.5 Discussion

We determined correlations between individual leg mechanical power and metabolic power associated with step-to-step transitions when walking on level-ground, and a range of uphill and downhill slopes at three velocities. Metabolic power was greater for walking on slopes greater and less than -3°, results supported by previous findings (15, 19, 21, 22). At steeper uphill slopes, P_{lead} and P_{trail} were more positive compared to level-ground walking (Figure 2.3a and b) and in agreement with previous research [43]. Similar to previous results [43], P_{lead} and P_{trail} were more negative with steeper downhill slopes compared to level-ground walking (Figure 2.3a and b). At faster velocities, metabolic power increased, the leading leg absorbed more power, and the trailing leg generated more power compared to slower velocities. With faster velocities across all slopes, P_{lead} became more negative while P_{trail} became more positive. Our findings of step-to-step transition mechanical power are consistent with previous research [43], while changes in metabolic and mechanical power due to velocity at each slope are novel. Finally, we found that P_{lead} or P_{trail} accounted for 57-76% of the variance in overall metabolic power over a wide range of slopes and velocities. Donelan et al. (8) found strong linear correlations between metabolic and mechanical power during step-to-step transitions due to varied step lengths over level-ground. However, we found that a quadratic model best described metabolic power and P_{lead}, and metabolic power and P_{trail} over a wide range of slopes and velocities.

In support of our hypothesis, we found that metabolic power was correlated with P_{lead} and P_{trail} ($R^2 \geq 0.57$) for all slopes at each velocity. Specifically, when we applied a linear model to our data, we found that 52-59% of the variance in metabolic power could be explained by changes in P_{lead} and P_{trail} over all slopes at each velocity (Table 2.1). Using a curve estimation regression with metabolic power as the dependent variable and P_{lead} or P_{trail} as the independent variable, we compared results of quadratic and linear models. We found that a quadratic model best described the correlation between metabolic power and P_{lead}, (R^2 0.71-0.76) and between metabolic power and P_{trail} (R^2 = 0.57-0.65). The use of a quadratic model to correlate metabolic and mechanical power is supported by previous research [54, 56, 58] that estimates metabolic cost from muscle activation squared.

We calculated ILPR based on quadratic models to determine the influence of mechanical power on metabolic power (Figure 2.4). Metabolic power increased more than the mechanical power generated by both legs during the step-to-step transition while walking up steeper slopes, resulting in a lower ILPR compared to level-ground. At steeper downhill slopes, P_{lead} influenced overall metabolic power more than P_{trail}. When walking on steeper downhill slopes, the leading leg absorbed most of the power (Figure 2.2b), while the trailing leg absorbed relatively little power (Figure 2.2a). Previous research suggests that eccentric muscle action associated with power absorption is more efficient than concentric muscle action [59, 60]. For example, an $ILPR_{lead}$ of 2.5 suggests that a large amount of mechanical power is being absorbed during the step-to-step transition relative to the cheap metabolic cost of absorbing that power. Thus, the leading leg was more efficient than the trailing leg when walking down steeper slopes.

The underlying reason for quadratic models better explaining the variance in metabolic power at each velocity across slopes may result from each leg's role during step-to-step transitions. At steeper uphill slopes, both legs produced power to move the COM forward and up. Walking up steeper slopes required greater propulsive force and positive work by the leading and trailing legs [43]. Both legs generated more positive power and the muscles presumably performed concentric work at steeper uphill slopes, which likely explains the decrease in ILPR, as concentric muscle action is less efficient (i.e. more metabolically costly) than eccentric muscle action [59, 60]. At steeper downhill slopes, increased metabolic power was mostly due to changes in P_{lead} and not

P_{trail}. The leading leg absorbed most of the power and performed eccentric work while walking downhill compared to the trailing leg, which could explain why further increases in metabolic power with steeper downhill slopes are more strongly correlated with P_{lead} than with P_{trail}. Further, $ILPR_{lead}$ increased exponentially while $ILPR_{trail}$ decreased with steeper downhill slopes (Figure 2.4). We predict that at downhill slopes steeper than -9^{o}, the magnitude of work absorbed by the leading leg would presumably continue to increase and metabolic power would continue to increase [26].

Trailing leg propulsive forces become negligible at -9^{o} [43], suggesting that there are no propulsive forces at downhill slopes steeper than -9^{o}. As propulsive forces decrease, the trailing leg's contribution is small during step-to-step transitions. Thus, the exponential relationship showed little change in metabolic power due to changes in P_{trail} on downhill slopes. The trailing leg was less efficient at steeper downhill slopes; therefore the smaller trailing leg contribution likely explains the low correlation between metabolic power and P_{trail} on downhill slopes.

2.5.1 Implications for lower leg assistive device design

A non-linear model best describes the relationship between metabolic and mechanical power across all slopes at each velocity. These relationships suggest that biomimetic assistive devices such as prostheses and orthoses need power, dampening, and robust control for sloped walking. Such devices should specifically consider the role of the trailing and leading legs. Advances in leg prostheses allow for powered plantarflexion through use of a battery, actuators, and springs [29, 61, 62]. People with leg amputations using such powered ankle-foot prostheses have significantly lower metabolic demands compared to using passive-elastic prostheses and equivalent metabolic demands compared to non-amputees during level-ground walking at 0.75-1.75 m/s [17, 63]. Non-amputees using ankle-foot orthoses that provide powered plantarflexion during walking reduced their metabolic demand in all assisted compared to unassisted orthotic conditions on level-ground and uphill slopes [46]. These results suggest that changes in ankle power significantly affect individual leg power and the metabolic cost of walking. Previous studies show that biological ankle power generation during level-ground walking accounts for 46-89% of the total mechanical power [10, 48]. Thus, future studies are needed to understand how joint-level power affects the metabolic cost of walking at different slopes and velocities.

Based on our results, effective assistive device control should consider the type of walking condition (i.e. slope), total mechanical power output of the leg, and temporal control at different phases of the gait cycle. Our results suggest that the mechanical power demands of an assistive device would differ and depend on whether the affected leg is leading or trailing in a phase of the gait cycle during sloped walking at different velocities. Given the greater $ILPR_{lead}$ compared to $ILPR_{trail}$ during downhill walking, a robust assistive device must be capable of power production, and significant power absorption to minimize metabolic costs.

2.6 Conclusions

We quantified metabolic power and step-to-step transition power of the leading and trailing legs during walking over a wide range of slopes and velocities. We found that individual leg mechanical power during step-to-step transitions accounts for an average 65% of the metabolic power needed to walk at 21 combinations of slope and velocity, and that quadratic models best described the relationships between each leg's power and metabolic power. Future studies are needed to explore extreme slopes and slower and faster velocities. Our results provide a foundation for the design of powered leg assistive devices for robust constraints such as varied slopes and velocities during walking. Understanding the contributions of individual joints during walking at different slopes and velocities will advance prosthetic and orthotic designs. Future work should examine joint dynamics at similar slopes and velocities.

2.7 Acknowledgments

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3 CHAPTER 2: THE ANKLE JOINT'S CONTRIBUTION TO INDIVIDUAL LEG WORK DURING UPHILL AND DOWNHILL WALKING OVER A RANGE OF SPEEDS

3.1 Abstract

The ankle joint provides 42% of the total leg positive power required to walk on level ground at various speeds; however, it is unclear how the ankle joint contributes to leg work when walking up and down slopes. Determining the ankle joint's contribution would benefit the design of biomimetic assistive devices. Thus, 19 healthy adults walked 1.00, 1.25, and 1.50 m/s on level ground and ±3°, ±6°, and ±9° while we collected kinematic and kinetic data. We calculated individual leg work and ankle work over the entire stance phase. The ankle joint's relative contribution to total individual leg positive work did not change with speed across uphill slopes, but decreased from 116% at 0° to 57% at +9° across all speeds. The ankle joint's contribution to total negative individual leg work decreased from 83% at 1.00 m/s to 53% at 1.50 m/s across all downhill slopes, and decreased from 92% at 0° to 47% at -9° across all speeds. The ankle significantly contributes to walking on slopes and this contribution changes during sloped compared with level ground walking. Thus, for ankle-foot assistive devices to provide biomimetic function, they must adapt during walking for different speeds and for uphill and downhill slopes.

3.2 Introduction

Level-ground walking is characterized mechanically by a single support phase when one foot is in contact with the ground, a step-to-step transition phase when both feet are in contact with the ground, and a leg swing phase when one foot is advanced forward. Previous studies have described the single support phase of walking using an inverted pendulum model, where the body's center of mass (COM) is represented by a point mass, and the leg by a rigid, massless strut [1-4]. This model suggests that minimal external mechanical work is necessary to sustain steady-speed level-ground walking during the single support phase because there is a constant phasic exchange of potential and kinetic energy [1, 3]. Thus, during a single stride at a steady speed, the net external mechanical work done on the body's COM is presumably near zero [2, 3]. However, during the step-to-step transition phase of level-ground walking, the muscles of each leg must perform both negative and positive work on the COM [64]. The leading leg absorbs mechanical work as it slows the downward and forward velocity of the COM while the trailing leg simultaneously produces mechanical work to redirect and reaccelerate the COM upward and forward [5-7]. Previous studies have found that individual leg step-to-step transition work constitutes the majority of the external mechanical work done during walking [1, 2, 4, 5, 8, 9]. The work performed on the COM also requires ~45% of the net metabolic cost of level-ground walking [22], whereas supporting body weight (primarily during single support) comprises ~28% of the net metabolic cost [22].

Previous studies have used the combined limbs method to account for the total external mechanical work done by both legs on the COM during the single and double support phases of walking [1, 65]. Using this method, the simultaneous positive and negative work done by the leading and trailing legs during the step-to-step transition is mathematically canceled out. In contrast, the individual limbs method, which accounts for the work done by each individual leg during the step-to-step transition, provides a 31% higher estimate of the COM work [6], but a lower estimate of the ankle joint's contribution to COM work during walking. Using the combined limbs method, one study suggests that the muscles surrounding the ankle joint contribute ~80% of the overall mechanical leg work required to walk on level ground [48]. When using the individual limbs method to calculate leg work and power, the muscles surrounding the ankle provide ~40% of the total positive individual leg power during level-ground walking over a range of speeds [10].

When walking at different speeds on level ground, uphill, and downhill slopes, individual leg mechanical work during the step-to-step transition phase changes [43]; thus it is likely that ankle joint work also changes. Further, because leg work changes throughout the stance phase of walking, it is important to consider ankle joint mechanics and the ankle joint's contribution to leg mechanics during the entire stance phase rather than only during single support or the step-to-step transition. For example, Franz et al. [43] quantified the mechanical work done by the leading and trailing legs throughout the stance phase and demonstrated that each leg absorbs and generates mechanical work, respectively. The leading and trailing legs primarily generate positive work during uphill walking on slopes of +3° to +9°, and primarily absorb negative work when walking downhill on slopes of -3° to -9° throughout the entire stance phase [43]. In other words, walking uphill requires almost no negative leg work (0.14 to 0.02 J/kg) and walking downhill requires almost no positive leg work (0.05 to 0.18 J/kg) [43].

The power generated by the muscles surrounding the ankle, knee and hip joints differs in magnitude when walking uphill or downhill. Lay et al. [66] found that the ankle generates 2 times and the hip generates 7 times higher peak power when walking up a 39% (~ 21.5°) grade compared to level-ground walking. In contrast, the knee absorbs power, and with a 40% greater peak magnitude when walking downhill at 21.5° [66] compared to level ground. These data indicate that each joint individually adapts to different slopes and performs a different amount of work. Similar to walking uphill, increasing walking speed requires the legs to produce more positive power and perform more positive work on the COM. When walking on level ground, total leg positive power increases with speed, but each joint's contribution to total positive individual leg power remains constant; the ankle contributes ~42%, while the knee and hip contribute ~16% and ~42%, respectively [10]. However, the ankle joint's relative contribution to the individual leg mechanical work or power when walking over a range of speeds on different slopes has not yet been determined.

Peak ankle plantar-flexor, knee extensor, and hip joint extensor moments increase during uphill compared to level ground walking, whereas peak ankle plantar-flexor moment decreases, peak knee extensor moment increases, and peak hip extensor moment remains unchanged during downhill compared to level-ground walking [11]. Similarly, ankle and hip joint sagittal plane ranges of motion increase while the knee joint sagittal plane range of motion remains unchanged during uphill compared to level ground walking [11]. Conversely, peak knee joint flexion increases, while ankle and hip joint total range of motion remains unchanged during downhill compared to level ground walking [11]. These data suggest that each joint has a different role in successfully negotiating uphill and downhill slopes.

A more comprehensive understanding of the ankle joint's kinematics and contribution to mechanical work during walking at different speeds and slopes has important implications for the design and control of biomimetic assistive devices such as powered ankle-foot orthoses and prostheses for people that lack normative ankle function. Leg and ankle work are commonly calculated as the integral of power with respect to time. Additionally, ankle joint moment and range of motion and the relationship between these variables, or work loops, provide visual representations of ankle function and have been quantified for use in powered prosthetic control [67-70]. To better replicate biological ankle function, lower-limb prostheses now include an untethered power supply and actuators to produce biomimetic forces and torques about a prosthetic ankle joint during the stance phase of level-ground walking [17-19, 21]. Use of a powered ankle-foot prosthesis that permits a range of motion similar to and that is capable of generating net positive work equivalent to that of a biological ankle [29, 61, 62] has resulted in significantly lower metabolic costs for people with transtibial amputation during level-ground walking over a wide range of speeds (0.75 to 1.75 m/s) compared with use of a passive-elastic prosthesis, and in equivalent metabolic costs compared to non-amputees [17, 29]. While level-ground walking likely accounts for the majority of terrain encountered during daily living for persons with a leg amputation, the ability to vary speed and negotiate more complex terrain with varying slopes is equally important to return them to full function. Thus, a comprehensive analysis of biological ankle joint kinematics and kinetics during walking on varied terrain is needed to improve the design and control of biomimetic powered prostheses and to potentially facilitate the design and control of powered orthoses/assistive devices.

To inform the design of powered prostheses and assistive devices, we quantified the biological ankle joint's relative contribution to individual leg work during the entire stance phase of walking across a wide range of slopes and speeds; information that can be used for prosthetic and assistive device design. We also quantified biological ankle joint kinematics, work, and power across a wide range of uphill and downhill slopes and speeds. This can inform the control strategies and actuation required for prostheses and assistive devices. Based on previous studies, we anticipated that total positive individual leg and ankle joint work would increase with uphill slopes and hypothesized that 1) the ratio of total positive ankle work to total positive individual leg work would be greater at steeper uphill slopes compared with level-ground walking. Based on previous studies, we expected that total negative individual leg and ankle work would increase in magnitude with downhill slopes and hypothesized that 2) the ratio of total negative ankle to total negative individual leg work would be smaller at steeper downhill slopes compared with level-ground walking. We predicted that ankle joint range of motion (ROM) would not significantly change with speed or downhill slope but would increase with steeper uphill slopes [11]. Finally, we hypothesized that 3) peak ankle joint moment and peak ankle joint power would increase on uphill slopes compared with level ground, would decrease on downhill slopes compared to level ground and would increase with speed on all slopes in accordance with previous studies [11, 71].

3.3 Methods

3.3.1 Subjects

All subjects gave their written informed consent prior to participating in this study according to a protocol approved by the United States Department of Veteran Affairs' Human Subjects Institutional Review Board and in accordance with the principles expressed in the Declaration of Helsinki. Nineteen healthy human subjects with no prior history of lower limb or neurological injury or pathology volunteered [13 M, 6 F, mean age 29.8 years (SD 8.83); mean mass 70.0 kg (SD 13.8)].

3.3.2 Experimental Protocol

We placed reflective markers on each subject using a modified Helen Hayes full body marker set. Markers were placed bilaterally on joint centers and clusters of at least 4 markers were placed on each body segment. Subjects walked on a dual-belt force-measuring treadmill (Bertec Corp., Columbus, OH) on seven slopes (0^o, $\pm3^o$, $\pm6^o$, & $\pm9^o$) and at three speeds (1.00, 1.25, & 1.50 m/s) per slope. We simultaneously collected motion (100 Hz) and force data (1000 Hz) for thirty seconds during each trial (21 trials total) and randomized the order of the trials.

3.3.3 Kinetics and Kinematics

We measured three dimensional (*3D*) ground reaction forces (*GRFs*) from each leg and normalized all data to each subject's mass (*m*). We filtered *GRFs* with a fourth order low-pass Butterworth filter and 20 Hz cut-off frequency using a custom program (Matlab, Mathworks, Natick, MA). We used an eight-camera motion capture system to simultaneously measure *3D* kinematics (100 Hz; Vicon Motion Systems; Oxford, UK). Then, we filtered the kinematic data with a zero phase shift fourth-order Butterworth low-pass filter and 10 Hz cut-off frequency using Visual3D software (C-Motion, Gaithersburg, MD). We then calculated ankle joint angles, moments, and powers (Fig. 3.1) based on calculated joint centers and filtered forces using Visual3D. We determined ground contact by detecting a 20 N threshold in the vertical *GRF* using a custom program (Matlab, Mathworks, Natick, MA).

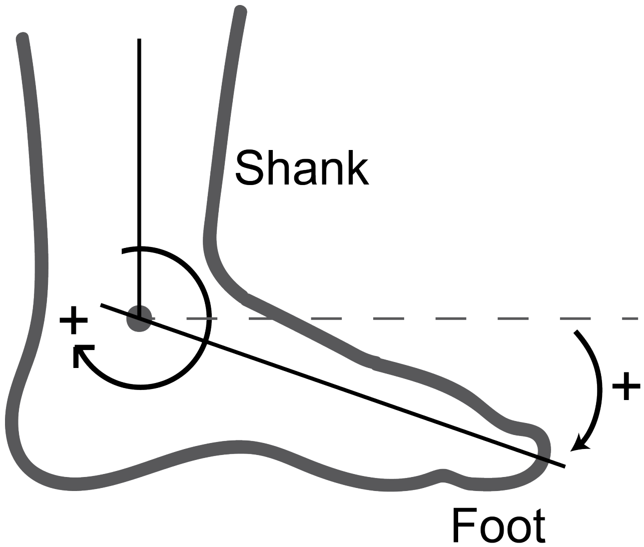

Figure 3.1. Schematic of the shank and foot segments. The plus symbols show the direction of positive ankle angle and plantarflexion moment. An ankle angle of zero degrees is indicated by the dashed line relative to the shank.

Net ankle joint work (W_{netA}) was calculated for each leg as the integral of ankle joint power (P_A) from touch-down (*td*) to toe-off (*to*) (stance phase) with respect to time (Eq. 3.1):

$$W_{netA} = \int_{td}^{to} P_A dt \qquad (3.1)$$

Positive and negative ankle joint works were then calculated as the integral of ankle joint power during the periods where ankle joint power was positive or negative, respectively, during the stance phase.

To determine the external mechanical power from each individual leg, we calculated the acceleration of the center of mass (a) with respect to time (t) as:

$$a_{ML}(t) = \frac{F_{ML}(t)}{m} \qquad (3.2)$$

$$a_{parallel}(t) = \frac{F_{parallel}(t) - mg*\sin(\Theta)}{m} \qquad (3.3)$$

$$a_{perp}(t) = \frac{F_{perp}(t) - mg*\cos(\Theta)}{m} \qquad (3.4)$$

where medio-lateral (*ML*), parallel (*parallel*), and perpendicular (*perp*) components of GRFs were calculated relative to the treadmill, respectively, and Θ is the slope of the treadmill. Then we calculated the velocity of the center of mass (v) as the integral of acceleration with respect to time:

$$v(t) = \int_0^t a(t)dt + v_o \qquad (3.5)$$

We determined integration constants (v_o) for perpendicular (v_{perp}) and medio-lateral (v_{ML}) velocities by assuming that the average v over a stride equaled zero and determined v_o for parallel velocity ($v_{parallel}$) by assuming that the average v over a stride equaled the treadmill velocity. We calculated the external mechanical power performed by each individual leg (P_{leg}) on the COM as the sum of the products of the *GRF* and center of mass velocity (v_{com}) in the medio-lateral, parallel, and perpendicular planes:

$$P_{leg} = GRF_{ML}v_{ML,com} + GRF_{parallel}v_{parallel,com} + GRF_{perp}v_{perp,com} \qquad (3.6)$$

Positive and negative leg works were calculated as the integral of leg power during the periods where leg power was positive or negative, respectively, for the stance phase. The ankle joint's contribution to individual leg work was calculated as the ratio of ankle work to leg work multiplied by 100:

$$Ankle's\ Contribution\ (\%) = \frac{Ankle\ Work}{Individual\ Leg\ Work} * 100 \qquad (3.7)$$

3.3.4 Statistics

We used two-way repeated measures ANOVAs (P<0.05) to compare net ankle joint work and the ankle joint's contribution to individual leg work at different slopes and speeds as well as the interaction between slope and speed. Our within subject variables were slope with seven levels:

0°, ±3°, ±6°, and ±9° and speed with three levels: 1.00, 1.25, and 1.50 m/s. Significant differences were further analyzed with pair-wise independent t-tests using a critical p-value corrected via the Bonferroni method. Statistical analyses were done using RStudio statistical software (RStudio, Boston, MA).

3.4 Results

3.4.1 Ankle and Leg Total Positive and Negative Work

We found main effects in total positive ankle work due to changes in speed and slope as well as an interaction effect between speed and slope ($P<0.001$). As speed changed from 1.00 m/s to 1.50 m/s, across all slopes, the average total positive ankle work over a stride increased 84% ($P<0.001$, Figure 3.2, Table 3.1). As slope changed from -9° to 0°, across all speeds, the average total positive ankle work increased 69% ($P<0.001$, Figure 3.2, Table 3.2). As slope changed from 0° to 9°, across all speeds, the average total positive ankle work increased 66% ($P<0.001$, Figure 3.2, Table 3.2). Across all slopes, the magnitudes of total positive ankle work at all speeds were different from each other ($P<0.005$, Table 3.1). Across all speeds, the magnitudes of total positive ankle work at all uphill and downhill slopes were different from level ground ($P<0.001$, Table 3.2).

We found main effects in total positive leg work due to changes in speed and slope as well as an interaction effect between speed and slope ($P<0.001$). As speed changed from 1.00 m/s to 1.50 m/s, across all slopes, the average total positive leg work over a stride increased 72% ($P<0.001$, Figure 3.2, Table 3.1). As slope changed from -9° to 0°, across all speeds, the average total positive leg work doubled ($P<0.001$, Figure 3.2, Table 3.2). As slope changed from 0° to

9°, across all speeds, average total positive leg work tripled (*P*<0.001, Figure 3.2, Table 3.2). Across all slopes, the magnitude of total positive leg work at 1.00 m/s was different from 1.50 m/s (*P*<0.001, Table 3.1). Across all speeds, the magnitudes of total positive leg work at all uphill and downhill slopes were different from 0° (*P*<0.001, Table 3.2).

We found main effects of speed (P<0.05) and slope (P<0.001) on total negative ankle work but no interaction effect between speed and slope. As speed changed from 1.00 m/s to 1.50 m/s, across all slopes, the magnitude of average total negative ankle work increased 8% (*P*<0.001, Figure 3.2, Table 3.1). As slope changed from -9° to 0°, across all speeds, the magnitude of average total negative ankle work decreased 43% (*P*<0.001, Figure 3.2, Table 3.2). As slope changed from 0° to 9°, across all speeds, the magnitude of average total negative ankle work decreased 66% (*P*<0.001, Figure 3.2, Table 3.2). Across all speeds the magnitudes of total negative ankle work at all uphill and downhill slopes were different from 0° except at 3° (*P*<0.001, Table 3.2).

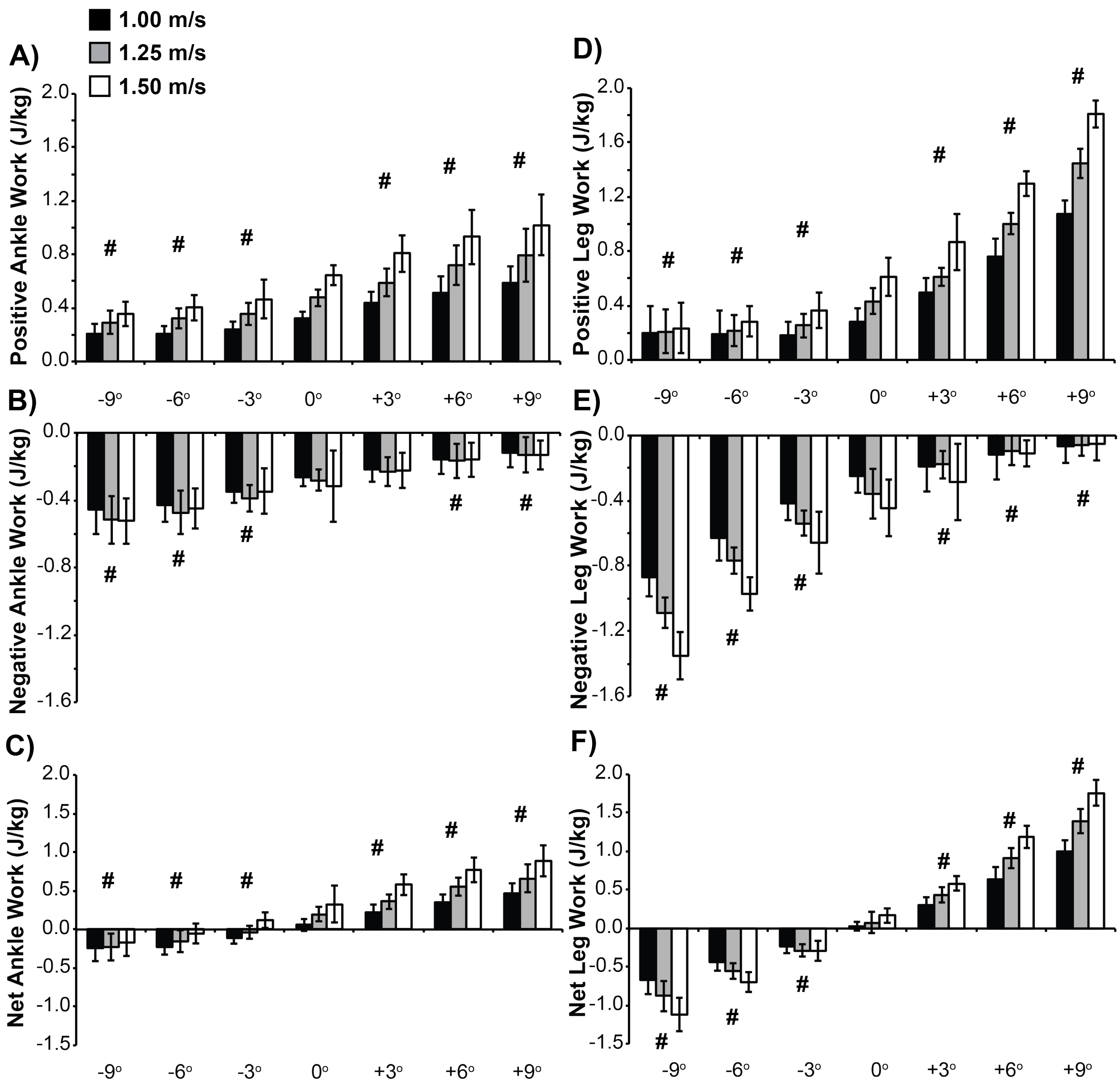

Figure 3.2. Average (S.D.) A) ankle joint total positive work, B) ankle joint total negative work, C) ankle joint net work, D) leg total positive work, E) leg total negative work, and F) leg net work, for all subjects walking 1.00, 1.25, and 1.50 m/s at -9° through +9°. # indicates significantly different from level ground. Significant differences between speeds are discussed in the Results section.

Table 3.1. Average (S.D.) ankle joint and leg biomechanics for all subjects walking 1.00 m/s, 1.25 m/s, and 1.50 m/s across all slopes. Percentage of Ankle/Leg Positive Work data are for level ground and all uphill slopes. Percentage of Ankle/Leg Negative Work data are for level ground and all downhill slopes. *indicates a significant difference from 1.00 m/s. #indicates a significant difference from 1.25 m/s.

Speed (m/s)	Total Ankle Positive Work (J/kg)	Total Leg Positive Work (J/kg)	Total Ankle Negative Work (J/kg)	Total Leg Negative Work (J/kg)	Ankle/Leg Positive Work (%)	Ankle/Leg Negative Work (%)	Ankle Range of Motion (rad)	Ankle Peak Moment (Nm/kg)	Ankle Peak Power (W/kg)
1.00	0.36# (0.08)	0.45 (0.13)	-0.28 (0.09)	-0.36 (0.12)	87 (37)	83 (37)	0.55 (0.11)	1.36# (0.42)	2.74# (1.05)
1.25	0.51* (0.11)	0.60 (0.10)	-0.31 (0.10)	-0.44 (0.09)	87 (30)	68 (25)	0.58 (0.16)	1.53* (0.44)	3.83* (1.39)
1.50	0.66*# (0.14)	0.78* (0.14)	-0.31 (0.13)	-0.55* (0.15)	88 (26)	53*# (20)	0.60 (0.24)	1.68* (0.50)	4.95*# (1.87)

Table 3.2. Average (S.D.) ankle joint and leg biomechanics for all subjects walking on slopes 0°, ±3°, ±6°, and ±9° across all speeds. **Bold** indicates a significant difference from 0°.

Slope (deg)	Total Ankle Positive Work (J/kg)	Total Leg Positive Work (J/kg)	Total Ankle Negative Work (J/kg)	Total Leg Negative Work (J/kg)	Ankle/Leg Positive Work (%)	Ankle/Leg Negative Work (%)	Ankle Range of Motion (rad)	Ankle Peak Moment (Nm/kg)	Ankle Peak Power (W/kg)
-9	**0.29** (0.08)	**0.21** (0.18)	**-0.50** (0.14)	**-1.10** (0.12)	-	**47** (14)	0.53 (0.15)	**1.01** (0.29)	**2.40** (0.72)
-6	**0.31** (0.08)	**0.23** (0.13)	**-0.45** (0.12)	**-0.79** (0.11)	-	**60** (17)	0.50 (0.14)	**1.16** (0.32)	**2.74** (1.05)
-3	**0.35** (0.10)	**0.27** (0.11)	**-0.36** (0.10)	**-0.54** (0.12)	-	**72** (18)	0.54 (0.33)	**1.37** (0.23)	3.21 (1.08)
0	0.48 (0.06)	0.44 (0.11)	-0.29 (0.11)	-0.35 (0.14)	116 (29)	92 (33)	0.55 (0.09)	1.56 (0.22)	3.78 (1.16)
+3	**0.61** (0.11)	**0.66** (0.13)	-0.22 (0.09)	**-0.22** (0.16)	**96** (22)	-	0.58 (0.08)	**1.75** (0.34)	**4.56** (1.60)
+6	**0.72** (0.16)	**1.02** (0.10)	**-0.16** (0.10)	**-0.11** (0.11)	**71** (17)	**-**	**0.64** (0.14)	**1.87** (0.41)	**4.99** (1.80)
+9	**0.80** (0.18)	**1.44** (0.10)	**-0.13** (0.09)	**-0.06** (0.09)	**57** (11)	**-**	**0.68** (0.10)	**1.94** (0.48)	**5.24** (1.91)

We found main effects of speed and slope on total negative leg work (P<0.001) as well as an interaction effect between speed and slope. As speed changed from 1.00 m/s to 1.50 m/s, across all slopes, the magnitude of average total negative leg work increased 53% (P<0.001, Figure 3.2, Table 3.1). As slope changed from -9° to 0°, across all speeds, the magnitude of average total negative leg work decreased 68% (P<0.001, Figure 3.2, Table 3.2). As slope changed from 0° to

9°, across all speeds, the magnitude of average total negative leg work decreased 83% (P<0.001, Figure 3.2, Table 3.2). Across all slopes, the magnitude of total negative leg work at 1.00 m/s was different from 1.50 m/s (Table 3.1). Across all speeds, the magnitudes of total negative leg work at all uphill and downhill slopes were different from level ground except at 3° (P<0.001, Table 3.2).

3.4.2 Ankle's Contribution to Individual Leg Work

We found no effect of speed (P=0.90) but a main effect of uphill slope on the ankle's relative contribution to total positive individual leg work (P<0.001). We did not find an interaction between speed and slope (P=0.09). As speed changed from 1.00 m/s to 1.50 m/s, across all uphill slopes, the ankle joint's relative contribution to total positive individual leg work remained unchanged at 87.0% and 87.6%, respectively (Figure 3.3, Table 3.1). At uphill slopes of 0° to 9°, across all speeds, the ankle contributed 116% and 57% of the total positive individual leg work, respectively (P<0.001, Figure 3.3, Table 3.2). Across all speeds, the ankle's relative contribution to total positive individual leg work at each of the uphill slopes were different from 0° (P<0.001, Table 3.2).

We found main effects of speed and downhill slope on the ankle's contribution to total negative individual leg work and an interaction effect between speed and downhill slope (P<0.001). As speed changed from 1.00 m/s to 1.50 m/s, across all downhill slopes, the ankle's contribution to total negative individual leg work decreased from 83% to 53% (P<0.001 Figure 3.3, Table 3.1). At downhill slopes of -9° and 0°, across all speeds, the ankle contributed 47% and 92% of the total negative individual leg work, respectively (P<0.001, Figure 3.3, Table 3.2). Across all downhill slopes, the ankle's contributions to total negative individual leg work at all speeds were different from each other except between 1.00 m/s and 1.25 m/s (Table 3.1). Across all speeds, the ankle's contribution to total negative individual leg work at downhill slopes were different from 0° (P<0.001, Table 3.2).

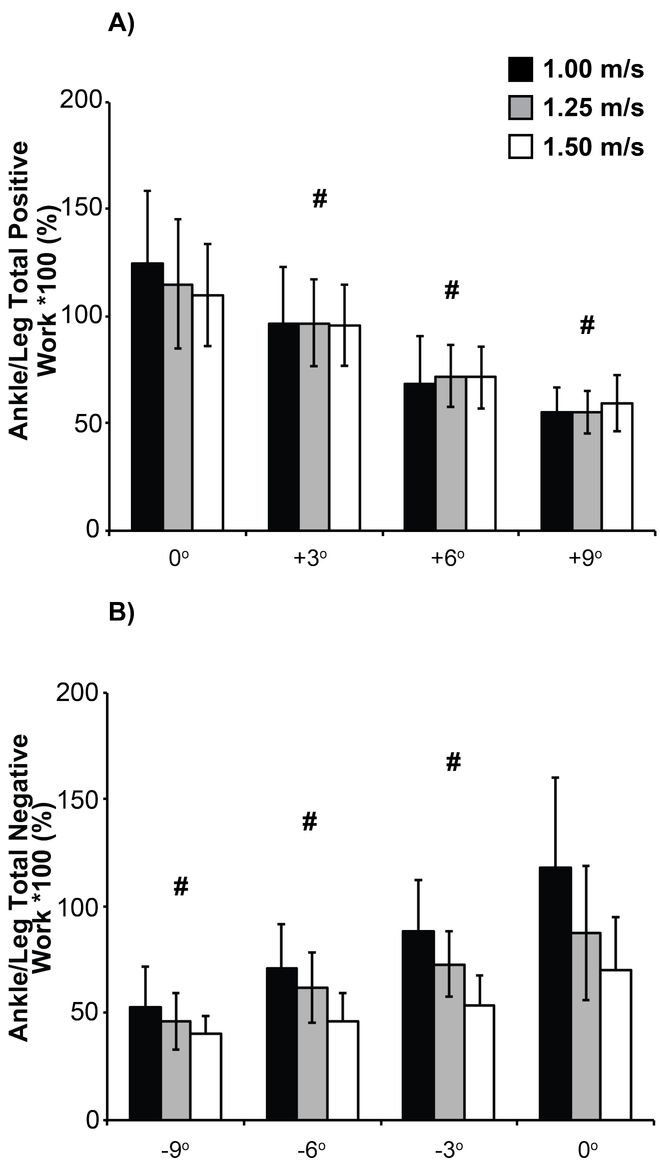

Figure 3.3. Average (S.D.) percentage of A) ankle/leg total positive work for level and all uphill slopes and B) ankle/leg total negative work level and all downhill slopes. Subjects walked 1.00, 1.25, and 1.50 m/s. # indicates significantly different from level ground. Significant differences between speeds are discussed in the Results section.

3.4.3 Ankle ROM

We found main effects of speed ($P<0.05$) and slope ($P<0.001$) on ankle ROM but did not find an interaction effect between speed and slope ($P=0.24$). As speed changed from 1.00 m/s to 1.50 m/s, across all slopes, average ankle ROM increased 9% ($P<0.001$) (Figure 3.4, Table 3.1). As slope changed from -9° to 0°, average ROM, across all speeds, increased 4% ($P<0.001$, Figure 3.4, Table 3.2). As slope changed from 0° to 9°, average ROM, across all speeds, increased 24% ($P<0.001$, Figure 3.4, Table 3.2). There was no main effect of speed on ankle ROM averaged across all slopes ($P>0.006$, with Bonferroni correction, Table 3.1). However, across all speeds, ankle ROM values at uphill 6° and 9° slopes were different from 0° ($P<0.001$, Table 3.2).

3.4.4 Ankle Moment

We found main effects of speed and slope on peak ankle moment during the stance phase ($P<0.001$, Figure 3.4). However, we did not find an interaction effect between speed and slope on peak ankle moment ($P>0.05$). As walking speed changed from 1.00 to 1.50 m/s, across all slopes, peak ankle moment increased 24% ($P<0.001$, Figure 3.4, Table 1). Similarly, as slope changed from -9° to 0°, across all speeds, peak ankle moment increased 54% ($P<0.001$, Figure 3.4, Table 3.2). Further, as slope changed from 0° to 9°, across all speeds, peak ankle moment increased 24% ($P<0.001$, Figure 3.4, Table 3.2). Across all slopes, the magnitudes of peak ankle moment at all speeds were different from each other except at 1.25 m/s compared with 1.50 m/s ($P<0.006$, Table 3.1). Across all speeds, the magnitudes of peak ankle moment at all uphill and downhill slopes were different from level ground ($P<0.001$, Table 3.2).

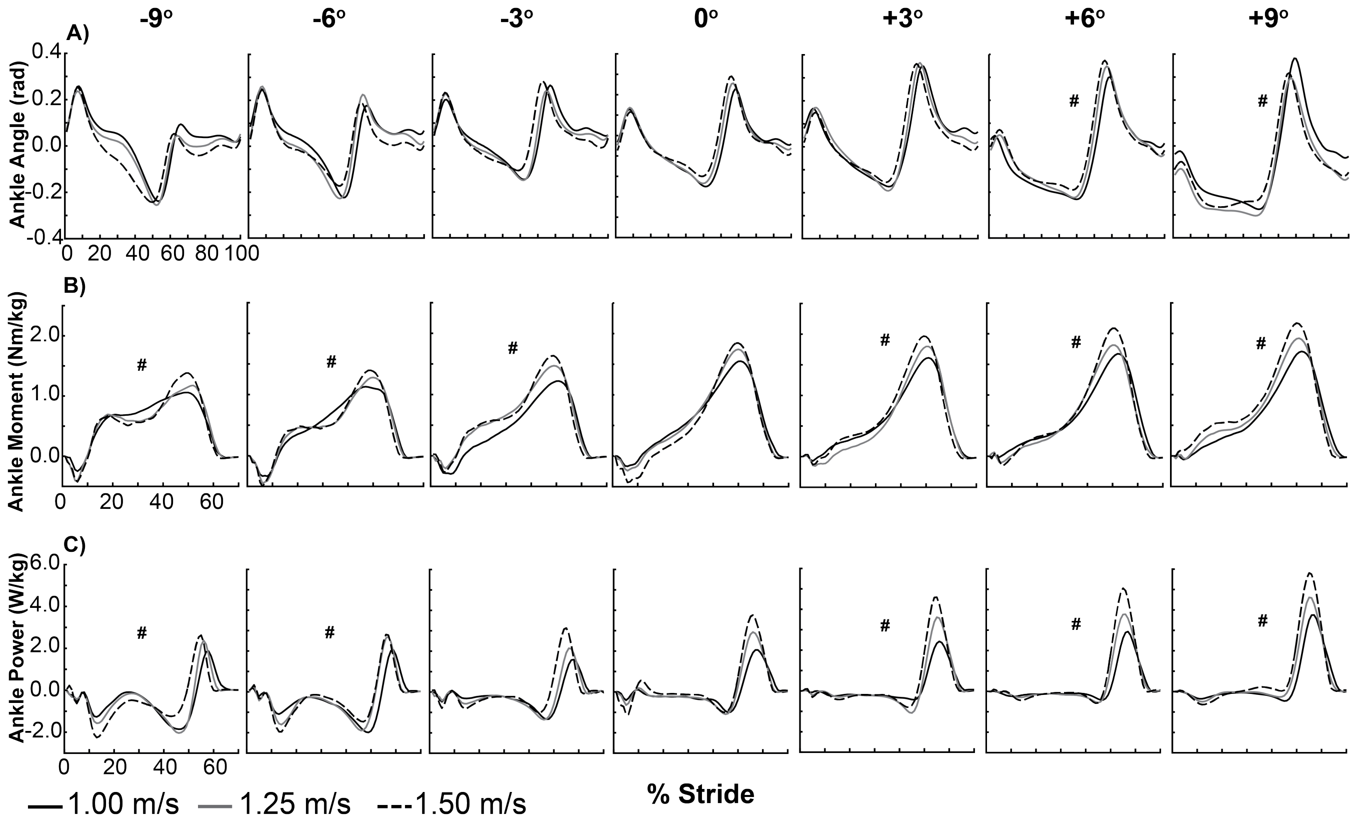

Figure 3.4. Ensemble averages of 10 strides for 19 subjects' ankle joint A) angles, B) moments, and C) powers versus % of a stride starting at heel-strike for each walking speed at each slope. Ankle angles are presented for the entire stride and ankle joint moments and powers are presented for the stance phase only (0-70% of a stride). Each column represents a different slope from -9° (left-most column) to +9° (right-most column). # indicates significantly different from level ground ankle ROM, peak ankle moment, and peak ankle power. Significant differences between speeds are discussed in the Results section.

3.4.5 Ankle Power

We found main effects of speed and slope on peak ankle power during the stance phase as well as an interaction effect between speed and slope (P<0.001). Specifically, as walking speed changed from 1.00 to 1.50 m/s, across all slopes, peak ankle power increased 81% (P<0.001, Figure 3.4, Table 3.1). Similarly, as slope changed from -9° to 0°, across all speeds, peak ankle power increased 58% (P<0.001, Figure 3.4, Table 3.2). Further, as slope changed from 0° to 9°, across all speeds, peak ankle power increased 39% (P<0.001, Figure 3.4, Table 3.2). Across all slopes, the magnitudes of peak ankle power at all speeds were different from each other (P<0.001, Table 3.1). Across all speeds, the magnitudes of peak ankle power at all uphill and downhill slopes were different from 0° except at -3° (P<0.001, Table 3.2).

3.5 Discussion

3.5.1 Ankle and Leg Total Positive and Negative Work

Our predictions regarding total ankle and leg positive and negative work were supported. Total positive ankle and individual leg work increased with faster speed and steeper uphill slopes and decreased with steeper downhill slopes. Similarly, as predicted, the magnitude of negative work done at both the ankle and leg increased with downhill slopes, i.e. the steepest downhill slope required the most negative work and the steepest uphill slope required the least negative work.

The greater total positive ankle and leg work with increasing speed and slope - as well as the greater magnitude of total negative work absorbed by the ankle and leg with decreasing slope - are in line with previously published studies [10]. We found similar values for net leg work compared to Franz et al. [43], but small differences in positive (0.18 J/kg) and negative (0.06 J/kg) work. These differences may be due to different data collection and analysis methods. We used a dual-belt instrumented treadmill to record ground reaction forces under each leg and then used ground reaction force data to calculate center of mass work from each leg throughout the stance phase. Previous studies have used an over-ground force platform within a ramp [11, 71] or used ground reaction force data collected from a single leg and assumed symmetry between legs [43]. It is possible that by using data from both legs and averaging over many steps, our data include fluctuations and asymmetries between the legs; resulting in a similar magnitude of net leg work but slightly higher positive leg work and slightly lower negative leg work. Our results corroborate the trends in leg and ankle total positive and total negative work presented by previous studies [10, 43, 71]. However, we also determined the contribution from the ankle relative to the overall leg, which provides insight into leg and joint function during the entire stance phase of walking.

3.5.2 Ankle's Contribution to Individual Leg Work

Similar to Farris and Sawicki [10], we found that the ankle joint's relative contribution to total positive individual leg work did not change with speed. We found that the ankle's contribution to total positive leg work was 116% on level ground over multiple speeds, whereas Farris and Sawicki [10] reported the ankle's contribution to total positive leg power as 42%. There are two important differences between our study and that of Farris and Sawicki. First, we defined and calculated individual leg work using the individual limbs method based on force data rather than summing joint work/power as was done in Farris and Sawicki [10]. Second, we report the ankle joint's contribution to total positive and negative leg mechanical *work*, not total positive leg mechanical *power*. Unlike previous studies, we also present the ankle joint's contribution to total negative individual leg work, which decreases with increasing speed and with steeper downhill slope.

Using the combined limbs method, Winter estimated that the ankle contributed 80% of the leg work during level ground walking [14, 48]. However, Winter reported ankle and knee positive and negative work, but did not report limb work calculations. Thus, it is unclear how the estimated contribution of the ankle was calculated for level-ground walking at a steady speed [48].

Our first hypothesis was not supported because the ankle's relative contribution to total positive work did not increase with steeper uphill slopes. With steeper uphill slopes, the ankle's contribution to total positive individual leg work decreased (from 116% at 0° to 57% at +9°). Our second hypothesis was supported; with steeper downhill slopes the ankle's contribution to total negative individual leg work decreased (from 92% at 0° to 47% at -9°).

3.5.3 Ankle ROM

We predicted that ankle ROM would not change with speed or downhill slope but found that ankle ROM increased with faster speed and decreased at steeper downhill slopes. In support of our prediction, and in agreement with Lay et al. [11], ankle ROM increased at steeper uphill slopes, as did peak plantarflexion angle. Both ankle ROM and peak ankle plantarflexion were lower at the steepest downhill slope compared to level ground (Figure 3.4). As documented in Lay et. al. [11], the ankle is mostly plantar-flexed during downhill walking and dorsi-flexed during the early and mid-stance phases of uphill walking. There were relatively small (1-3°) differences in ankle ROM between our results and previously published results, which is likely due to differences in marker placement and may be due to differences in over ground versus treadmill data collection methods. Ankle ROM changes with both speed and slope but more so to accommodate uphill slopes, indicating that mechanical limitations to ROM in assistive devices may limit their function at faster walking speeds and/or when walking uphill.

3.5.4 Ankle Moment

Our third hypothesis regarding peak ankle moment was supported. Peak ankle moment increased with increases in speed and slope. The external mechanical work done on the center of mass must increase when walking uphill and decrease when walking downhill to counteract gravity. Thus, the legs must perform more positive work on the center of mass when walking uphill and less positive work on the center of mass when walking downhill. We found that peak ankle moment (and thus total positive ankle work) increases when walking uphill and decreases when walking downhill, again establishing that ankle function varies when navigating slopes.

3.5.5 Ankle Power

Our third hypothesis regarding peak ankle power was also supported. Peak ankle power increased with increases in speed and slope to provide more positive limb power to overcome gravity when walking uphill. The ankle provides a large portion [11, 71] of the propulsive power to move the center of mass forward and up during the step-to-step transition, even on downhill slopes. Further, ankle power fluctuates more than ankle moment or ROM to accommodate sloped walking. For example, when walking at +9°, ankle power increases 39% while ankle moment and ROM only increase 24% compared to walking on level ground.

3.5.6 Application to Assistive Device Design

Understanding the function of the biological ankle as well as its relationship to overall limb function during the stance phase of uphill and downhill walking at various speeds is critical for design and control of robust prostheses, orthoses, and other assistive devices. We found that ankle range of motion, peak moment, peak power, total positive, total negative and net work change with speed and slope. We also found that the ankle's contribution to the leg changes with speed and slope and thus should be accounted for in the control of powered assistive devices.

The significant dependence of ankle ROM on both speed and slope suggests that both should be design criteria for biomimetic ankle-foot devices for walking. Currently, powered prostheses are capable of a range of motion similar to a biological ankle, 15° of dorsi-flexion and 25° of plantar-flexion [29]. However, we found peak dorsiflexion angles between 9° and 18° and peak plantar-flexion angles of between 15° and 22° in the biological ankle for negotiating various speeds and slopes indicating that current designs are not capable of fully replicating biological ankle function for a variety of speeds and slopes.

Powered ankle-foot prostheses are capable of generating normative work and power for a variety of speeds on level ground [16, 18-21]. However, it is unknown how use of these devices affects walking mechanics on uphill and downhill slopes. Powered ankle-foot prostheses use data input from sensors embedded in the device as well as from biological ankle work loops for level-ground walking to modulate the control of the device [16, 19, 29, 62]. Our results show that the relationships between ankle moment and angle, the relationship between ankle work and leg work, and the required contributions from each change significantly with walking speed and slope. The design of hardware or control systems for assistive devices that incorporate biological ankle mechanics during uphill and downhill walking may restore normative biomechanics and reduce the metabolic cost of walking for individuals with impaired or no ankle function.

3.5.7 Future Directions for Research

Our results suggest that the biological ankle joint facilitates the absorption and generation of work during level and sloped walking at various speeds. Future studies are needed to determine the contribution of the hip and knee joints while negotiating various terrains during walking. Providing a more complete understanding of the mechanical work done at each biological joint relative to the leg during walking under various conditions will lead to enhanced design and biomimetic control of robust prostheses, orthoses, and assistive devices.

3.6 Conclusion

The ankle joint's relative contribution to total individual leg positive and negative work varies substantially with slope. With steeper uphill slopes, the ankle's contribution to total leg positive work significantly decreases from an average of 116% on level ground to 57% at +9° across all speeds; with steeper downhill slopes, the ankle's contribution to total negative leg work decreases from 92% on level ground to 47% at -9° across all speeds. Additionally, ankle range of motion, peak moment, and peak power significantly increase with increasing slope and decrease with decreasing slope. Thus, replicating biological ankle function over a range of speeds and slopes with prosthetic or assistive devices will require dynamic changes in positive and negative work, as well as peak moments, powers, and range of motion. Our results will inform control and design of biomimetic ankle-foot prostheses and assistive devices for walking at various speeds over a range of slopes. The development and implementation of such devices has the potential to restore normative mechanics and metabolic costs for users and has the potential to increase quality of life [72-75], level of mobility and function [34, 76, 77], and may reduce comorbidities (such as knee osteoarthritis and low back pain) commonly incurred by persons with impaired or no ankle function [33, 73, 74].

4 CHAPTER 3: BIOMECHANICAL AND METABOLIC EFFECTS OF USING A POWERED COMPARED TO PASSIVE-ELASTIC ANKLE-FOOT PROSTHESIS DURING UPHILL AND DOWNHILL WALKING

4.1 Abstract

People with a leg amputation typically have metabolic demands that are 10-30% higher than non-amputees walking at the same speed over level ground. The increased metabolic demand is likely due to reduced prosthetic ankle push-off work with use of passive-elastic prostheses. Powered ankle-foot prostheses have been developed that normalize the metabolic cost and step-to-step transition work of people with a leg amputation during level-ground walking over a range of speeds, but the effects on uphill and downhill slopes remain unclear. We sought to understand the effects of using a passive-elastic compared to powered ankle-foot prosthesis on metabolic cost and step-to-step transition work of people with a leg amputation during sloped walking. 10 people with a transtibial amputation walked 1.25 m/s on 0°, ±3°, ±6°, and ±9° using passive-elastic and powered ankle-foot prostheses while we collected metabolic energy expenditure, kinematic, and kinetic data. We calculated net metabolic power, individual leg step-to-step transition work, and net individual leg work symmetry over a stride. Use of the powered prosthesis reduced net metabolic power by 5% when walking on uphill slopes of +3° and +6° ($p<0.05$), and individual leg step-to-step transition work was not significantly different compared to use of a passive-elastic prosthesis. However, net individual leg work over an entire stride was more symmetric on uphill slopes of +6° and +9° ($p<0.01$) with use of the powered compared to passive-elastic prosthesis. Use of a powered ankle-foot prosthesis has the potential to reduce metabolic demand and increase leg work symmetry during walking on uphill slopes for people with a leg amputation.

4.2 Introduction

The stance phase mechanics of level-ground walking can be described as an inverted pendulum where the body's center of mass (CoM) is represented as a point mass atop the leg and the leg is represented by a massless, rigid strut [1, 2, 65]. This model suggests that minimal mechanical work is required to maintain steady-speed level-ground walking due to the phasic exchange of kinetic and gravitational potential energy [1, 2, 65]. However, during the step-to-step transition the leading and trailing legs simultaneously perform positive and negative external mechanical work on the CoM, respectively [6]. To perform mechanical work throughout stance and the step-to-step transition, the muscles of the legs of non-amputees consume metabolic energy such that supporting body weight (primarily during single support) comprises ~28%, and redirecting and accelerating the CoM (primarily during the step-to-step transition) comprises ~45% of the net metabolic power of walking on level ground [22]. More specifically, the horizontal propulsive force (primarily provided by the gastrocnemius and soleus) to redirect and accelerate the CoM of non-amputees prior to and during the step-to-step transition comprises ~50% of the net metabolic power required to walk on level ground [23]. Additionally, over a wide range of speeds, the muscles surrounding the ankle joint provide ~41% of the total individual leg positive mechanical power when walking on level ground [10] and the ratio of ankle joint positive work to individual leg positive work ranges from 0.47 to 1.16 when walking on slopes of -9° to +9°, respectively [78]; indicating that the function provided by the muscles surrounding the ankle joint is critical to steady-state walking at various speeds and slopes.

People with transtibial amputation that use a passive-elastic prosthesis cannot generate net positive prosthetic ankle power and therefore compensate by generating positive power with more proximal muscles in their affected leg compared to the legs of non-amputees [79]. This reallocation of leg power to more proximal joints may be responsible for the 10-30% higher metabolic demand typically reported for people with a unilateral transtibial amputation compared with non-amputees during walking at the same speed over level ground [17, 80]. However, one study found that young, active Service members and Veterans with a leg amputation using a passive-elastic prosthesis have similar metabolic demands compared to non-amputees walking at the same speed on level ground [81]. Use of a prosthesis that provides stance-phase power has normalized metabolic cost and biomechanics (unaffected limb knee adduction moment, preferred walking speed, step-to-step transition individual leg work, net ankle work) for persons with unilateral transtibial amputation compared to age, height, and weight-matched non-amputees and has improved metabolic cost and biomechanics compared to use of a passive-elastic prosthesis over a wide range of speeds on level ground [17, 33, 80]. While level ground has been the focus of many studies investigating the effects of amputation, ramps are often encountered in activities of daily living. The effects of walking uphill and downhill on joint kinetics and kinematics have been documented for healthy adult populations but it remains unclear how use of a powered ankle-foot prosthesis affects the metabolic cost and mechanics of walking uphill and downhill.

From recent studies, it is clear that the individual leg work over the entire stance phase for non-amputees changes when walking on uphill and downhill slopes [11, 43, 71]. For example, positive individual leg work increases and negative individual leg work decreases with steeper uphill slopes until the work done by each leg is almost entirely positive when walking up a +9° slope [43]. Similarly, the magnitude of negative leg work increases and positive leg work decreases with steeper downhill slopes until the work done by each leg is almost entirely negative when walking down a -9° slope [43]. Step-to-step transition work also increases with speed and uphill slope in people with a transtibial amputation [17, 80]. When people with a unilateral transtibial amputation use a passive-elastic prosthesis to walk on level ground, their step-to-step transition work is compromised such that their unaffected leg, when leading, absorbs more negative work and when trailing, generates greater positive work compared to non-amputees [17, 80, 82]. Further, their affected leg, when leading, absorbs less negative work, and when trailing, provides less positive work compared to non-amputees [5, 17, 80]. Use of a powered ankle-foot prosthesis has normalized the step-to-step transition work for people with an amputation walking on level ground at a range of speeds [17] and up a 5° slope [80], but the effects on individual leg work and step-to-step transition work when walking with a powered prosthesis on a range of uphill and downhill slopes remain unknown.

We sought to determine if use of a powered ankle-foot prosthesis results in lower metabolic demand and more symmetric individual leg work for persons with a unilateral transtibial amputation compared with use of a passive-elastic prosthesis during walking on uphill and downhill slopes. We hypothesized that use of a powered prosthesis compared to use of a passive-elastic prosthesis would, at each slope: 1) require less net metabolic power, 2) result in lower unaffected leg leading step-to-step transition work, 3) result in greater affected leg trailing step-to-step transition work, and 4) improve individual leg work symmetry over an entire stride.

4.3 Methods

4.3.1 Subject Recruitment

Ten healthy adults with a traumatic unilateral transtibial amputation (6M, 4F, mean ± SD: age 42 ±11 yrs, height 1.70 ±0.08 m, and mass 81.3 ±14.7 kg) participated. All subjects provided written informed consent according to the Declaration of Helsinki and the US Department of Veterans Affairs institutional review board. Subjects self-reported that they were at or above a Medicare K3 classification, free of neurological, cardiovascular, and musculoskeletal disease other than that associated with a unilateral leg amputation.

4.3.2 Experimental Protocol

4.3.2.1 Tuning of the Powered Prosthesis

First, a certified prosthetist from BionX Medical Technologies aligned the powered prosthesis (BiOM T2, BionX Medical Technologies, Inc. Bedford, MA, USA) to each subject. We then placed reflective markers on subjects' lower limbs in a manner consistent with a modified Helen Hayes marker set with markers placed over joint centers and clusters of at least 4 markers placed over each segment. We placed reflective markers at the approximate locations of the prosthetic foot 1st and 5th metatarsal heads, posterior calcaneus, and medial and lateral malleoli. We placed "malleoli" markers for the powered prosthesis on the encoder, which coincided with the center of rotation in the sagittal plane (Figure 4.1). We placed "malleoli" markers for the passive-elastic prosthesis on the medial and lateral edges of the carbon fiber prosthesis at the most dorsal point of the keel (Figure 4.1). Subjects then walked 1.25 m/s on a dual-belt force-measuring treadmill (Bertec Corp., Columbus, OH, USA) for at least 45 seconds at each slope of 0°, ±3°, ±6°, and

±9° while we simultaneously measured kinematics (Vicon, Oxford, UK) and ground reaction forces.

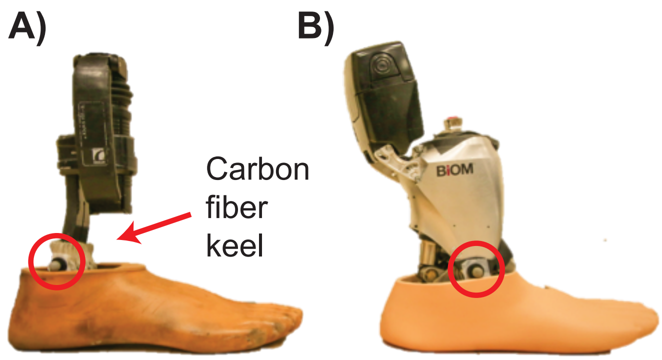

Figure 4.1. Lateral "malleolus" marker placement (circled) on the A) most dorsal point of keel on the passive-elastic prosthesis and B) encoder/center of rotation for the BiOM powered prosthesis.

To objectively tune the powered prosthesis at each slope, we calculated prosthetic ankle angles, moments, powers, and work normalized to subject mass with the prosthesis after each 45-second trial and compared these data with averages from 20 non-amputees walking at the same speed and slope [83]. We then iteratively tuned the powered prosthesis for each slope using a tablet with software provided by the manufacturer (BionX Medical Technologies, Bedford, MA, USA) until prosthetic ankle range of motion, peak moment, peak power, and net work were equivalent to the unaffected ankle and matched non-amputee averages within one standard deviation (Figure 4.2).

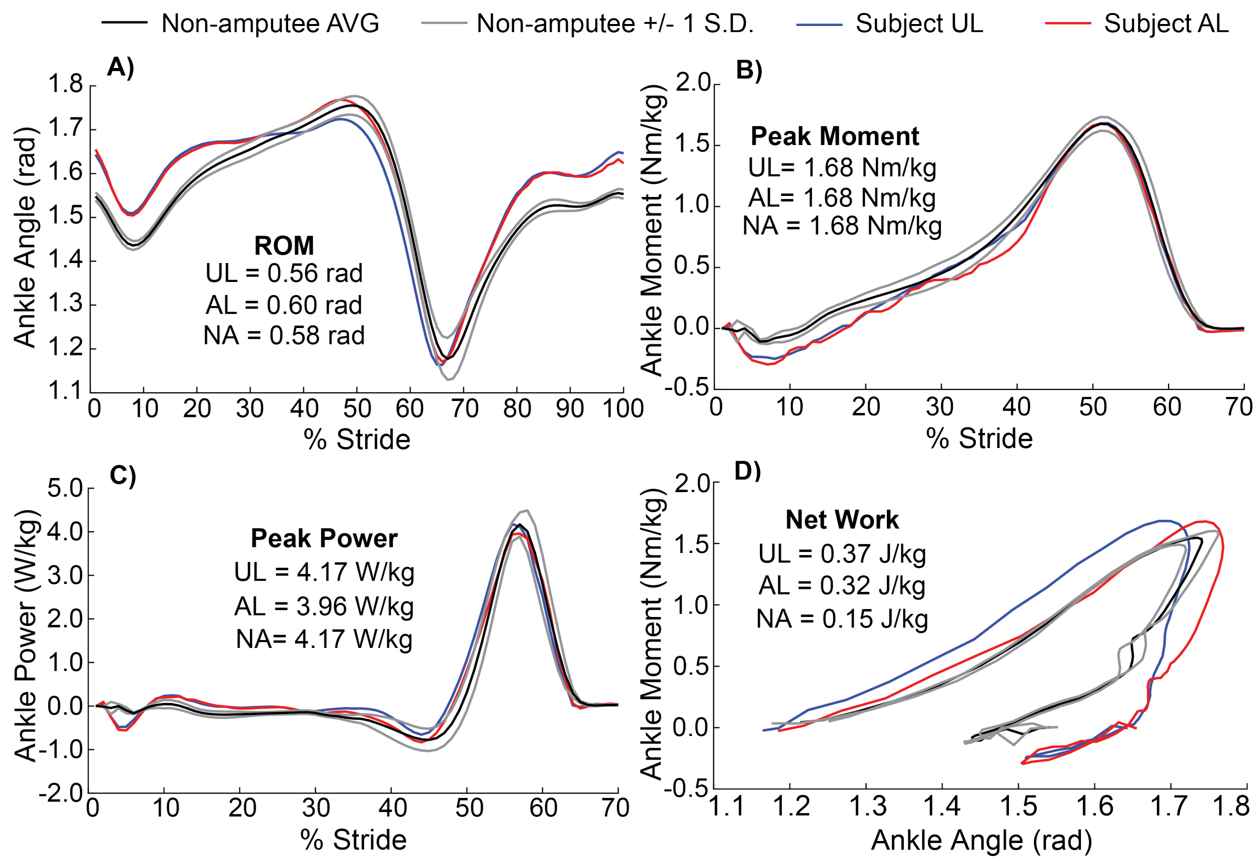

Figure 4.2. Example of BiOM tuning at +3° for a representative subject walking at 1.25 m/s to match non-amputee (black) A) ankle range of motion, B) peak ankle moment, C) peak ankle power, and D) net ankle work. The subject's unaffected leg (UL) biological ankle (blue) and affected leg (AL) prosthetic ankle (red) match non-amputee (black) average ankle kinematics and kinetics within 1 S.D. (grey).

4.3.2.2 Metabolic Data Collection

Each subject had approximately 6 hours of acclimation to the powered prosthesis during tuning. Then, subjects walked 1.25 m/s for 5 minutes at each of seven slopes (0°, ±3°, ±6°, ±9°) on 2 separate days that were at least 22 hours apart while we measured their rates of oxygen consumption and carbon dioxide production via indirect calorimetry (ParvoMedics, Salt Lake City, UT). On one day, subjects used their own passive-elastic prosthesis (Table 4.1) and on another day subjects used the powered prosthesis. The order of days was randomized. We averaged the last 2 minutes of each 5-minute trial and calculated metabolic power using a standard equation [53]. We calculated net metabolic power by subtracting the metabolic power required for standing quietly from the power required for each walking condition and normalized to subject body mass including the appropriate prosthesis. Metabolic testing sessions were at the same time of day; subjects fasted for at least 2 hours prior to metabolic data collection. Trial order for each day was randomized to mitigate order effects.

Table 4.1. Subject anthropometrics and their own passive-elastic prosthetic feet.

Sex	Height (m)	Mass w/ BiOM (kg)	Mass w/ ESAR (kg)	Passive-elastic foot model
F	1.66	59.5	58.0	Freedom Innovations Renegade
F	1.66	65.3	61.7	Ottobock Triton IC60
F	1.68	69.4	68.5	Össur Pro-flex XC
M	1.75	72.1	70.3	Freedom Innovations Renegade
M	1.71	78.0	77.0	Össur Vari-flex
F	1.71	84.1	81.8	Össur Vari-flex XC
M	1.82	89.4	88.9	College Park Soleus
M	1.85	96.2	95.3	Össur Proflex
M	1.83	97.1	95.5	Ability Dynamics Rush 81
M	1.82	102.3	100.2	Ability Dynamics Rush 87
AVG (SD)	1.70 (0.08)	81.3 (14.72)	79.7 (14.98)	-

4.3.2.3 Kinetic and Kinematic Data Collection

Subjects walked while using the powered prosthesis for approximately 10 hours over 5 experimental sessions on the treadmill at the same speed and slopes prior to the session where we measured kinetic and kinematic data for analyses. All experimental sessions were at least 24 hours and no more than 2 weeks apart. The first two sessions were each 2-3 hours long and dedicated to tuning the powered prosthesis at each slope. During the third through fifth sessions (each 1.5 hours in length) we measured metabolic energy expenditure from subjects using the powered prosthesis tuned at each slope, the powered prosthesis tuned for level ground, and a passive-elastic prosthesis during walking at each slope. Then, in the sixth session (approximately 2.5 hours in length), subjects walked 1.25 m/s on a dual-belt force-measuring treadmill (Bertec

Corp., Columbus, OH) at slopes of 0°, ±3°, ±6°, and ±9° using their own passive-elastic prosthesis (Table 4.1) and the powered prosthesis while we simultaneously measured kinematics at 100 Hz and ground reaction forces (GRFs) at 1000 Hz. We filtered GRF data using a fourth order recursive Butterworth filter with a 30 Hz cutoff and kinematic data were filtered using a sixth order recursive Butterworth filter with a 7 Hz cutoff. We used a perpendicular GRF threshold of 20 N to determine ground contact. We averaged at least 5 strides (heel-strike to heel-strike of the same foot) for each subject at each condition and calculated an ensemble average of all ten subjects.

We determined the external mechanical power from each individual leg by calculating the acceleration of the center of mass (a) with respect to time (t):

$$a_{ML}(t) = \frac{F_{ML}(t)}{m} \tag{1}$$

$$a_{parallel}(t) = \frac{F_{parallel}(t) - mg*\sin(\Theta)}{m} \tag{2}$$

$$a_{perp}(t) = \frac{F_{perp}(t) - mg*\cos(\Theta)}{m} \tag{3}$$

where medio-lateral (ML), parallel ($parallel$), and perpendicular ($perp$) GRF components were calculated relative to the treadmill, m is body mass including the appropriate prosthesis, and Θ is the treadmill slope. Then we calculated the center of mass (COM) velocity (v) as the integral of acceleration with respect to time:

$$v(t) = \int_0^t a(t)dt + v_o \tag{4}$$

We determined integration constants (v_o) for perpendicular (v_{perp}) and medio-lateral (v_{ML}) velocities by assuming the average v over a stride equaled zero and determined v_o for parallel velocity ($v_{parallel}$) by assuming the average v over a stride equaled the treadmill velocity. We calculated the external mechanical power performed by each individual leg (P_{leg}) on the COM as the sum of the products of the GRFs and center of mass velocities (v_{com}) in the medio-lateral, parallel, and perpendicular planes:

$$P_{leg} = GRF_{ML} v_{ML,com} + GRF_{parallel} v_{parallel,com} + GRF_{perp} v_{perp,com} \quad (5)$$

We calculated net work as the integral of leg power with respect to time over an entire stride. We defined symmetry as the ratio of the affected leg net work divided by the unaffected leg net work. To determine step-to-step transition work, we integrated leg power with respect to time during double support when the unaffected leg was leading and the affected leg was trailing [17].

4.3.2.4 Statistical Analyses

We used paired independent t-tests to determine differences in net metabolic power between prostheses at each slope with a significance level of 0.05. Because we tested the effect of using a prosthesis at each slope on net metabolic power and thus performed only one t-test per slope, we did not correct our critical alpha value according to the Bonferroni method. Similarly, to determine the effect of using a prosthesis on individual leg step-to-step transition work and symmetry we performed paired independent t-tests at each slope with a significance level of 0.05.

4.4 Results

When tuning the powered prosthesis for each subject at each slope, subjects' ankle biomechanics matched non-amputee values within one standard deviation at slopes of 0°, ±3° and +6°. For the remaining slopes (-6°, ±9°) subjects' ankle biomechanics using the powered prosthesis matched the unaffected ankle joint range of motion, peak moment, peak power, and net work values and tuning parameters were utilized that resulted in the most symmetric values.

We found an effect of prosthetic foot type on net metabolic power at +3° and +6° for subjects with an amputation where the use of the powered prosthesis resulted in 5% lower net metabolic power at both slopes compared with use of a passive-elastic prosthesis ($p=0.021$ and $p=0.013$, respectively, Figure 4.3, Table 4.2). At +9°, net metabolic power with use of the powered prosthesis trended lower (13%, $p=0.155$) but because only 2 subjects were able to complete the five-minute metabolic trial with primarily oxidative metabolism (RER < 1.0) using their passive-elastic prosthesis and only 3 using the powered prosthesis, our effect size was reduced.

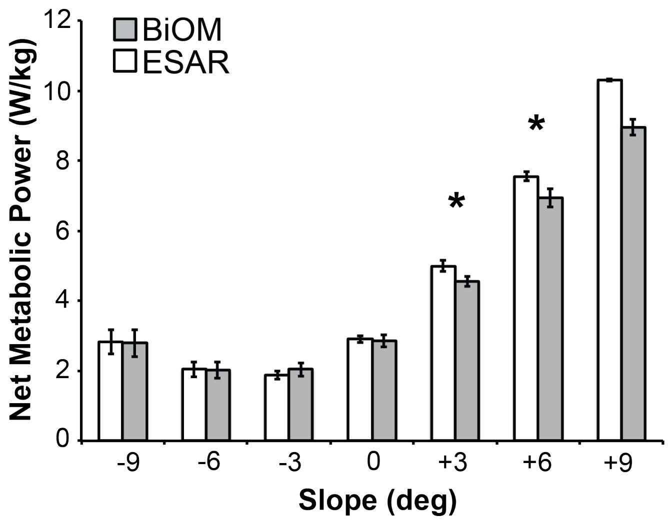

Figure 4.3. Net metabolic power (S.E.) of subjects with an amputation walking 1.25 m/s on slopes of -9° to +9° with powered (grey) and passive-elastic ESAR (white) ankle-foot prostheses. *indicates a significant difference between BiOM and ESAR prostheses.

Table 4.2. Net metabolic power (W/kg) for all subjects walking 1.25 m/s with the BiOM powered prosthesis and their own passive-elastic energy storage and return (ESAR) prosthesis at each tested slope.

Slope (deg)	**Foot**	**Sub 01**	**Sub 02**	**Sub 03**	**Sub 04**	**Sub 05**	**Sub 06**	**Sub 07**	**Sub 08**	**Sub 09**	**Sub 10**	**AVG (S.E.)**
+9	BiOM	9.06	-	-	-	-	-	9.28	8.53	-	-	8.96 (0.22)
	ESAR	-	-	-	-	-	-	10.33	10.27	-	-	10.30 (0.03)
+6	BiOM	6.65	-	7.84	6.78	-	7.58	6.63	5.94	-	8.08	7.07 (0.29)
	ESAR	7.38	-	-	7.78	-	7.46	7.29	7.28	-	7.38	7.43 (0.07)
+3	BiOM	4.33	4.95	4.92	4.44	4.63	4.83	4.83	3.61	4.97	5.28	4.68 (0.15)
	ESAR	4.40	6.00	5.42	4.79	4.72	4.53	4.98	4.52	5.25	4.40	4.90 (0.16)
0	BiOM	3.60	3.63	2.65	2.66	2.01	2.51	3.02	2.59	3.00	3.00	2.87 (0.16)
	ESAR	2.56	3.23	3.32	2.96	2.85	3.05	2.81	2.40	2.82	2.56	2.86 (0.09)
-3	BiOM	2.11	2.87	2.92	2.58	1.47	1.89	1.73	1.15	1.99	2.04	2.07 (0.18)
	ESAR	1.58	2.00	2.67	1.42	1.90	1.65	2.10	1.56	1.84	1.58	1.83 (0.12)
-6	BiOM	2.26	2.57	3.60	1.61	1.69	1.74	1.88	1.06	2.15	2.24	2.08 (0.22)
	ESAR	1.88	2.04	3.82	1.67	1.86	1.87	1.89	1.34	1.80	1.88	2.01 (0.21)
-9	BiOM	2.98	3.42	5.85	2.18	1.94	2.63	2.61	1.51	2.40	2.60	2.81 (0.38)
	ESAR	2.81	2.64	5.68	2.96	2.61	2.67	2.81	1.37	2.18	2.81	2.85 (0.35)

We did not find a significant effect of prosthetic foot type on individual leg step-to-step transition work for either the leading unaffected leg or the trailing affected leg (p>0.05, Figure

4.4). We did, however, find a significant effect of prosthetic foot type on individual leg net work symmetry at uphill slopes of +6° and +9° ($p<0.01$, Figure 4.5) where use of the powered prosthesis improved symmetry by 212% (from 0.26 with passive-elastic to 0.81 with powered) and 180% (from 0.33 with passive-elastic to 0.93 with powered) compared to use of the passive-elastic prosthesis, respectively.

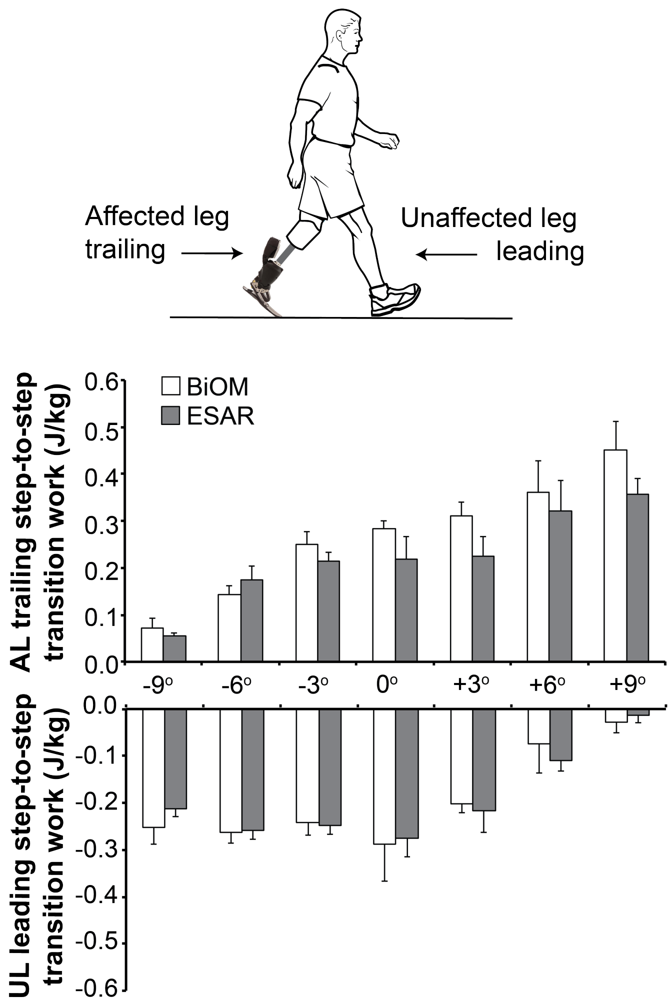

Figure 4.4. Positive affected leg (AL) trailing step-to-step transition work (±S.E., J/kg, top) and negative unaffected leg (UL) step-to-step transition work (±S.E., J/kg, top) for subjects with an amputation walking 1.25 m/s on slopes of -9° to +9° with powered (white) and passive-elastic ESAR (grey) ankle-foot prostheses.

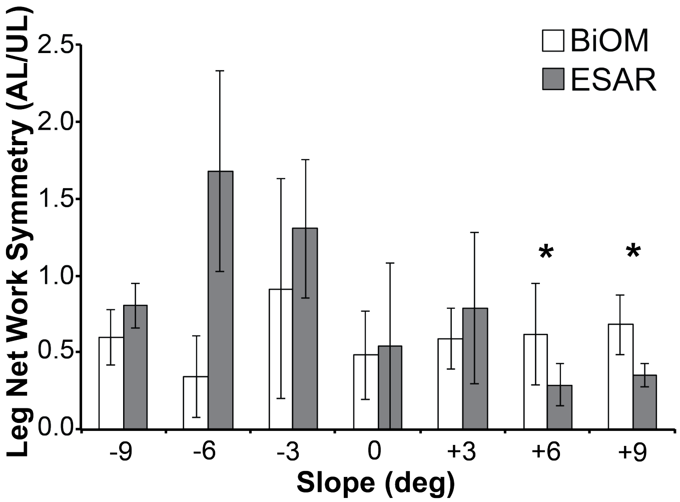

Figure 4.5. Net leg work over an entire stride symmetry (AL/UL) for subjects walking 1.25 m/s on slopes of -9° to +9° with powered (white) and passive-elastic ESAR (gray) ankle-foot prostheses. *indicates significant difference between BiOM and ESAR.

4.5 Discussion

In partial support of our first hypothesis, we found that use of the powered ankle-foot prosthesis significantly reduced net metabolic power for subjects with a unilateral transtibial amputation at +3° and +6° and net metabolic power trended lower at +9° with use of the powered compared to passive-elastic prosthesis. The differences between our data and those of Herr and Grabowski [17] on level ground could be due to different powered prosthesis tuning strategies, and/or the mass of the user. We tuned the powered prosthesis for each slope by measuring and comparing prosthetic ankle biomechanics to the ankle biomechanics of non-amputees using the same motion capture system rather than using data provided by the prosthesis. There are ten tuning parameters that we adjusted in the powered prosthesis to potentially change the prosthetic ankle kinematics and kinetics realized by each subject. Changes to the tuning parameters results in adjustments to the response of the proprietary control strategy, but does not guarantee specific kinematics and kinetics because each subject may respond differently to the tuning changes. Though no studies have systematically varied tuning parameters and investigated the effects on the metabolic demand and biomechanics of walking, the powered prosthesis uses prosthetic ankle angle and rate of change of ankle angle to determine the state space control and appropriate prosthetic ankle response throughout the gait cycle [29, 62, 84] and thus can vary between subjects. Additionally, our subjects had an average mass of 81.3 kg whereas in Herr and Grabowski [17], subjects had an average mass of 98.5 kg. The minimum subject mass suggested for prescription of the powered prosthesis is 80 kg. We included subjects outside this recommendation because our goal was to represent a typical active population of people with transtibial amputation and thus all subjects' data were used in our analyses. If we had excluded subjects lighter than 80 kg in the present study, our results would have been the same; there were no additional significant differences when using subjects >80 kg compared to using all subjects' data. While Mattes et al. [85] found that adding mass to a passive-elastic prosthesis – until the total mass and inertia matched the intact limb, similar to the powered prosthesis [17] – resulted in greater gross metabolic power of approximately 21 W per 1 kg added, it is unclear how the combination of prosthetic ankle power and added distal mass relative to the unaffected leg change metabolic demand. Furthermore, the addition of prosthetic ankle push-off work alone does not necessarily reduce the metabolic cost of walking on level ground for people with a transtibial amputation [86].

Our second hypothesis that use of the powered prosthesis would result in reduced unaffected leg leading step-to-step transition work was unsupported. Similarly, we refute our third hypothesis that use of the powered prosthesis would increase affected leg trailing step-to-step transition work. While we found significant increases in prosthetic ankle net work at all uphill slopes and -3° during tuning, this increase in work did not change the individual leg step-to-step transition work calculated via the individual limbs method [6]. The difference in our results and those of Herr & Grabowski [17] or Russell-Esposito et al. [80] could be due to the aforementioned powered prosthetic tuning, the prosthetic model, or the method of data collection. The prosthetic model used in this study was the commercially available BiOM T2 and the prosthetic model used in Herr & Grabowski [17] was a prototype of the BiOM developed prior to commercialization. Thus, some changes in the control algorithm could exist between the two prosthetic models. Further, our ground reaction force data were collected over at least 5 consecutive strides on an instrumented treadmill whereas data from Herr & Grabowski [17] and Russell-Esposito [80] were collected on force plates mounted in a walkway; thus, force plate targeting may have contributed to differences in step-to-step transition work.

Our fourth and final hypothesis that individual leg net work symmetry over a stride would improve (be closer to 1.0) with use of the powered prosthesis was supported at +6° and +9°. The differences at these uphill slopes was most likely due to the inability of the passive-elastic prosthesis to generate net positive power and provide it to the user during push-off. It is possible that because walking at a steady speed up steep slopes requires a large increase in external mechanical work done on the center of mass, the typical compensation strategy observed in individuals using a passive-elastic prosthesis (i.e. the unaffected leg and the muscles acting at the hip joint of the affected leg increase positive power output [79]) is no longer sufficient to maintain constant external mechanical work and thus a constant speed [82].

In the present study, we calculated external mechanical work but it is entirely possible that internal work changes with use of the powered prosthesis when walking uphill and downhill

[87]. It is also likely that external work done at each joint in both limbs changes significantly with use of the powered prosthesis compared to use of a passive-elastic prosthesis. Thus, future studies should investigate external mechanical work contributions from the muscles acting at the ankle, knee, and hip joints for both the affected and unaffected legs to better understand the effects of walking while using powered and passive-elastic ankle-foot prostheses on different uphill and downhill slopes.

4.6 Conclusion

We found that the use of a powered ankle-foot prosthesis reduced the net metabolic power required to walk up a +3° and +6° slope and trended lower at +9° but did not change the net metabolic power required to walk on level ground or on downhill slopes of -3°, -6°, -9° compared with use of a passive-elastic prosthesis for people with a transtibial amputation. However, use of the powered ankle-foot prosthesis improved leg net work symmetry at uphill slopes of +6° and +9°, potentially highlighting the limitations of passive-elastic prostheses to accommodate walking up steeper slopes. Though net work at the prosthetic ankle was significantly greater when using the powered prosthesis during the tuning process, individual leg step-to-step transition work did not change compared to using a passive-elastic prosthesis. To gain further insight into the biomechanical modifications due to use of a powered ankle-foot prosthesis, future studies should investigate joint level contributions for persons with a unilateral transtibial amputation walking on various slopes with powered and passive-elastic prostheses.

5 CHAPTER 4: EFFECTS OF PASSIVE AND POWERED ANKLE-FOOT PROSTHESES ON LEG JOINT WORK DURING SLOPED WALKING

5.1 Abstract

The muscles acting at the ankle joint provide ~42% of the total positive individual leg power required for level-ground walking over a range of speeds and provide push-off work utilized during the step-to-step transition phase. People with a leg amputation using passive-elastic prostheses exhibit reduced prosthetic ankle power and push-off work compared to non-amputees and compensate by increasing affected leg (AL) hip joint work and unaffected leg (UL) joint and leg work during walking. Use of powered ankle-foot prostheses normalizes step-to-step transition work during level-ground walking over a range of speeds, but the effects on joint work during level-ground, uphill, and downhill walking have not been assessed. We investigated how use of passive-elastic and powered ankle-foot prostheses affect leg joint biomechanics during sloped walking. 10 people with a transtibial amputation walked 1.25 m/s on a dual-belt instrumented treadmill at 0°, ±3°, ±6°, and ±9° using passive-elastic and powered prostheses while we measured kinematic and kinetic data. We calculated ankle, knee, hip, and individual leg positive, negative, and net work. Over a stride, AL prosthetic ankle positive work was 23-30% higher ($p<0.05$) during walking on uphill slopes of +6°, and +9°, AL prosthetic ankle net work increased (was more positive) by 1 to 10 times ($p\leq0.005$) on all uphill and downhill slopes, and AL net work was more positive ($p<0.05$) at uphill slopes of +3° and +9° with use of the powered compared to passive-elastic prosthesis. Use of a powered compared to passive-elastic ankle-foot prosthesis significantly increased AL prosthetic ankle and leg net work on uphill and downhill slopes. Greater prosthetic ankle positive and net work through use of a powered prosthesis could

improve symmetry between the legs of people with a transtibial amputation and thus would presumably improve preferred walking speed, functional mobility and quality of life.

5.2 Introduction

The stance phase of level-ground walking has been mechanically described as an inverted pendulum [1-4]. This model of walking suggests that no net external mechanical work is required to maintain steady speed walking on level-ground due to the phasic energy exchange of gravitational potential and kinetic energy [2, 3]. However, during the step-to-step transition phase when both feet are in contact with the ground, the individual legs simultaneously perform positive and negative work on the body's center of mass (COM) [6]. The leading leg performs negative work to slow the downward acceleration of the COM while the trailing leg performs positive work to redirect and accelerate the COM into the next step [6]. The muscles surrounding the hip, knee, and ankle joints each contribute to the overall work done by the individual leg on the COM.

During the stance phase of level-ground walking in non-amputees, the muscles acting at the hip and ankle provide both positive and negative work and power while the muscles acting at the knee joint primarily provide energy absorption or negative mechanical work and power. The muscles acting at the hip predominantly provide positive power at the beginning of the stance phase to extend the leg and raise the COM to its highest point at mid-stance [11, 12]. The muscles acting at the ankle mainly provide push-off power at the end of the stance phase to transition the leg into the swing phase, and redirect and accelerate the COM [11-13]. The muscles acting at the knee perform negative work or absorb energy at the end of the swing phase and beginning of the stance phase, indicating that these muscles primarily absorb work and decelerate the shank prior to heel strike [10]. For example, when walking at 1.50 m/s on level ground, the muscles acting at the hip and ankle perform 62% and 180% more positive than negative work, respectively, and the muscles acting at the knee perform 74% more negative than positive work over a stride, further demonstrating the functional roles of the muscles acting at each joint to maintain steady speed walking [71]. Across a wide range of speeds, the contributions of the muscles acting at the hip, ankle, and knee to total leg positive power do not change during level-ground walking [10]. More specifically, for non-amputees the muscles acting at the hip, ankle, and knee provide 39%, 41%, and 20% of the total positive leg power over speeds ranging from 0.75 to 2.00 m/s during walking on level ground [10].

Each joint's contribution and work change on uphill and downhill slopes. When non-amputees walk up a 10° incline, the hip and ankle work combine to provide 86% and 95% of leg positive and net work, respectively and to walk down a 10° decline the muscles acting at the knee perform 58% and 81% of the negative and net leg work [71]. Furthermore, as non-amputees walk up a 21.3° slope, peak positive hip moment more than doubles, peak positive ankle moment increases 19%, and peak positive knee moment almost doubles, compared to walking on level ground [11]. Conversely, when non-amputees walk down a 21.3° slope, peak positive hip moment remains constant, peak positive ankle moment decreases 44%, and the magnitude of peak negative knee moment increases over four-fold compared to walking on level ground [11]. Not only are the aforementioned joint kinetic changes detected for sloped walking, but the contribution of the muscles acting at the ankle joint to the individual leg work also changes with slope [78]. Specifically, in non-amputees, the ratio of positive ankle work to positive leg work ranges from 0.57 to 1.16 during walking on uphill slopes and level ground [78]. Furthermore, the ratio of negative ankle work to negative leg work ranges from 0.47 to 0.92 during walking on downhill slopes and level ground [78]. Leg joint biomechanics and the ankle's contribution to the leg vary significantly with slope for non-amputees, but it is not yet known how persons with a leg amputation using powered and passive-elastic prostheses adapt to walking on uphill and downhill slopes.

Because the muscles acting at the ankle joint provide push-off work in late stance prior to the step-to-step transition, and contribute a large portion of the total positive leg work and power required for steady-speed walking, populations who have little or no ankle function (e.g. people with an amputation, aging populations, stroke sufferers, etc.) require the contralateral leg or muscles acting at the more proximal joints of the affected leg (AL) to compensate for the lost power [14, 15]. Typically, people with a leg amputation are prescribed a passive-elastic energy storage and return (ESAR) prosthesis that is made of carbon fiber, and functions like a spring with no ability to generate power anew or to articulate. When people with a unilateral transtibial amputation use such passive-elastic prostheses, they have 10-30% higher metabolic demands to walk at the same speeds as non-amputees [34, 88, 89], compensate for lack of ankle push-off work with increased unaffected leg (UL) and decreased AL step-to-step transition work [17, 80, 82]. They also exhibit a slower preferred walking velocity [17, 81] and have increased knee joint adduction moments in their UL [33] compared to non-amputees. Additionally, when walking on level ground using passive-elastic prostheses, people with a unilateral transtibial amputation exhibit an increase in knee flexion in their AL compared to their UL at heel strike and activate their AL biceps femoris more than their UL biceps femoris [35]. The increased AL knee flexion has been attributed to the shape of the prosthetic socket that is flexed to increase patellar tendon loading for patellar tendon bearing socket designs [35]. However, Winter and Sienko [14] showed almost no positive or negative sagittal plane knee power in the first half of the stance phase for persons with a leg amputation walking on level ground with conventional solid-ankle cushioned heel prostheses, implying that co-activation at the knee joint and increased knee joint stiffness could result in less knee power. The advent of ESAR ankle-foot prostheses has resulted in no changes in knee sagittal plane range of motion compared to use of older, conventional solid-ankle cushioned heel prostheses [90] though it is not yet known how advanced prostheses affect knee sagittal plane moments and powers.

In contrast to a passive-elastic prosthesis, a commercially available powered ankle-foot prosthesis (BiOM) has a one degree of freedom ankle articulation (plantar- and dorsi-flexion), generates and provides battery-powered mechanical push-off work in late stance, and has normalized the biomechanics and metabolic demands for walking in people with a transtibial amputation compared to non-amputees on level ground [17, 81]. To tune the BiOM to the user, tuning parameters within the device are adjusted until the user's net prosthetic ankle work is within 2 standard deviations of average non-amputee ankle work values [91], where ankle work equals the product of ankle moment and ankle angle. The BiOM uses an encoder to determine prosthetic ankle position (angle) and provides appropriate torque (moment) and damping with springs in-parallel and in-series with the articulating prosthetic ankle, and a battery-powered motor and transmission [20]. A state space controller based on biological ankle work loops (moment vs. angle curve) during steady speed level-ground walking is used to govern the response of the BiOM [20]. During the stance phase of level-ground walking, the function of the prosthetic ankle is divided into three sub-phases: 1) controlled plantarflexion from heel-strike to foot-flat, 2) controlled dorsiflexion when the tibia progresses over the foot, and 3) powered plantarflexion when the prosthetic ankle generates positive power and provides push-off work, which acts to redirect and accelerate the COM into the next step [20]. During the swing phase (the fourth state space) the prosthetic ankle does not absorb or generate power, but rather returns to a neutral position (ankle angle) to clear the ground and prepare for the next heel strike [20]. Use of this powered ankle-foot prosthesis has normalized the metabolic costs and biomechanics (preferred walking speed, step-to-step transition work) during level-ground walking at speeds of 0.75 to 1.50 m/s for people with a leg amputation compared to non-amputees [17, 80]. However, to our knowledge only one study has investigated how use of a powered prosthesis affects uphill walking and found that the powered prosthesis normalized metabolic rate while walking on level ground and normalized step-to-step transition work on both level ground and a 5° inclined ramp [80]. It remains unclear how use of a powered ankle-foot prosthesis affects leg joint biomechanics and joint work contributions during uphill and downhill walking over a range of slopes compared to use of a passive-elastic prosthesis.

Thus, we sought to determine the effects of using passive and powered ankle-foot prostheses on leg joint biomechanics of people with a transtibial amputation walking at a range of uphill and downhill slopes. We hypothesized that with use of the powered compared to passive-elastic prosthesis: 1) UL total positive and net work would decrease, 2) AL total positive and net work would increase, 3) UL ankle and hip positive and net work would decrease, 4) AL prosthetic ankle positive and net work would increase and hip positive and net work would decrease, and 5) knee work would remain unchanged when walking at each slope.

5.3 Methods

5.3.1 Subject Recruitment

Ten healthy adults with a unilateral transtibial amputation (6M, 4F, mean ± SD: age 42 ±11 yrs, height 1.7 ±0.08 m, and mass without a prosthesis 77.3 ±14.8 kg) (Table 4.1) provided written informed consent according to the Declaration of Helsinki and US Department of Veterans Affairs institutional review board. Subjects self-reported that they were at a Medicare functional classification of K3 or higher, free of neurological, cardiovascular, and musculoskeletal disease other than that associated with a unilateral leg amputation.

5.3.2 Experimental Protocol

5.3.2.1 Tuning of the Powered Prosthesis

First, a certified prosthetist from BionX Medical Technologies aligned the powered prosthesis (BiOM T2, BionX Medical Technologies, Inc. Bedford, MA, USA) to each subject. We then placed reflective markers on subjects' lower limbs according to a modified Helen Hayes marker set. We placed markers over joint centers and clusters of at least 4 markers over each segment. We also placed reflective markers on the AL at the approximate locations of the prosthetic foot 1st and 5th metatarsal heads, posterior calcaneus, and medial and lateral malleoli, matching the locations on the UL. We placed "malleoli" markers for the powered prosthesis on the encoder, which coincided with the center of rotation in the sagittal plane (Figure 4.1). We placed "malleoli" markers for the passive-elastic prosthesis on the medial and lateral edges of the carbon fiber prosthesis at the most dorsal point of the keel (Figure 4.1). Subjects then walked using the BiOM at 1.25 m/s on a dual-belt force-measuring treadmill (Bertec Corp., Columbus, OH, USA) for at least 45 seconds at slopes of 0°, ±3°, ±6°, and ±9° while we simultaneously measured kinematics at 100 Hz (Vicon, Oxford, UK) and ground reaction forces at 1000 Hz.

To objectively tune the BiOM, we calculated prosthetic ankle angles, moments, powers, and work normalized to body mass including prosthetic mass using Visual 3D software (C-Motion, Germantown, MD) after each 45-second trial and compared these data with averages from 20 non-amputees walking at the same speed and slopes [83]. We then iteratively tuned the BiOM for each slope using a tablet with software provided by the manufacturer (BionX Medical Technologies, Bedford, MA) until the AL prosthetic ankle range of motion, peak moment, peak power, and net work normalized to mass matched the UL and non-amputee averages within one standard deviation (Figure 4.2).

5.3.2.2 Kinetic and Kinematic Data Collection

Subjects walked while using the powered prosthesis for approximately 10 hours over 5 experimental sessions on the treadmill at the same speed and slopes prior to the session where we measured kinetic and kinematic data for analyses. All experimental sessions were at least 24 hours and no more than 2 weeks apart. The first two sessions were each 2-3 hours long and dedicated to tuning the powered prosthesis at each slope. During the third through fifth sessions (each 1.5 hours in length) we measured metabolic energy expenditure from subjects using the powered prosthesis tuned at each slope, the powered prosthesis tuned for level ground, and a passive-elastic prosthesis during walking at each slope (see Chapter 3 Methods). Then, in the sixth session (approximately 2.5 hours in length), we simultaneously measured kinematics at 100Hz and ground reaction forces (GRFs) at 1000 Hz while subjects walked 1.25 m/s on a dual-belt force-measuring treadmill (Bertec Corp., Columbus, OH) at slopes of 0°, ±3°, ±6°, and ±9°. We filtered ground reaction forces using a fourth order recursive Butterworth filter with a 30 Hz cutoff and filtered kinematic data using a sixth order recursive Butterworth filter with a 7 Hz cutoff. Vertical ground reaction force data from each leg were used to determine ground contact with a threshold of 20N. We calculated sagittal plane joint powers using Visual3D software (C-Motion, Germantown, MD). Using a custom Matlab (MathWorks, Natick MA) script, we integrated joint power with respect to stride time to determine joint work over a stride (heel-strike to heel-strike of the same foot). We summed ankle, knee, and hip joint work over a stride to calculate leg work over a stride. We averaged at least 5 strides for each subject at each condition and calculated an ensemble average of all 10 subjects.

5.3.3 Statistical Analyses

Because our hypotheses are based on changes in leg or joint work at each slope, and on the effects of using each prosthesis on leg and joint work, we used one-way repeated measures ANOVAs with prosthetic foot type (powered BiOM or passive-elastic ESAR) as the independent variable and leg or joint positive, negative, or net work as the dependent variable with a significance level of 0.05 at each slope. We removed data outliers from statistical analyses if they fell outside the first or third interquartile range (R Studio, Boston, MA).

5.4 Results

We tuned the powered prosthesis so that prosthetic ankle joint range of motion, peak moment, peak power, and net work matched non-amputee averages within one standard deviation on level ground and slopes of ±3° and +6°. For the remaining slopes (-6°, ±9°) we matched the prosthetic ankle joint mechanics to those of the UL as closely as possible.

We found no effect of prosthetic foot type on UL positive, negative, or net work ($p>0.05$, Figure 5.1). We found no effect of prosthetic foot type on AL positive work ($p>0.05$) but did find a effect on AL net work at uphill slopes of +6° and +9° ($p<0.05$, Figure 5.1). We also found no effect of prosthetic foot type on AL negative work at all slopes ($p<0.001$, Figure 5.1). AL net work was more positive on +6°, and +9° slopes by 146%, and 82%, respectively with use of the BiOM compared to a passive-elastic prosthesis (Figure 5.1).

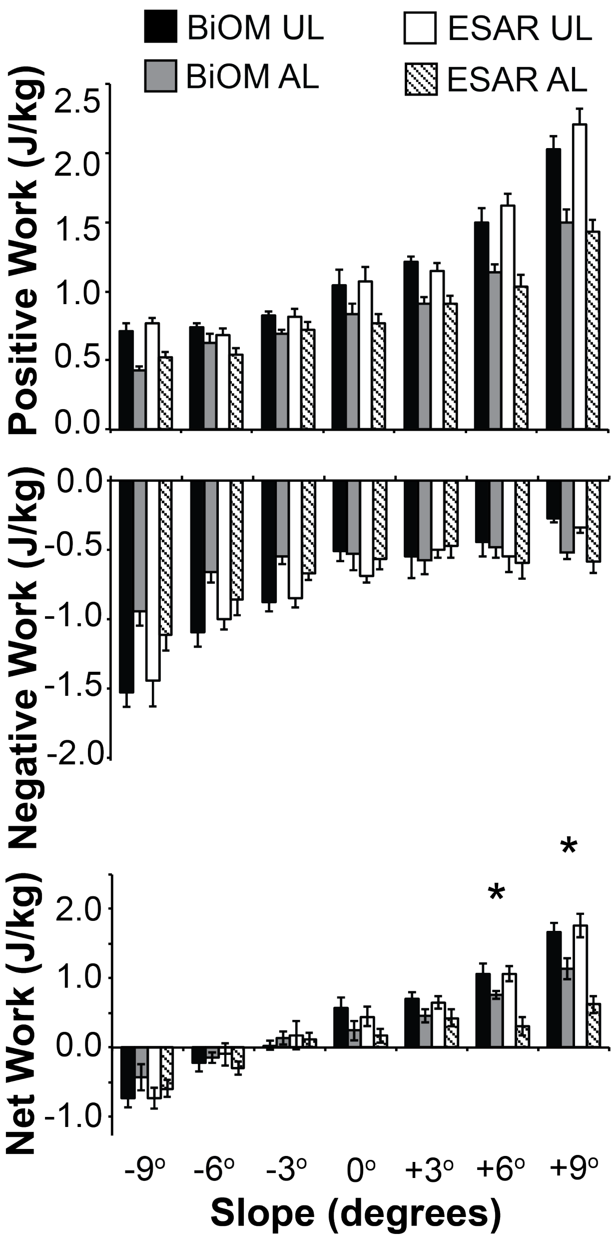

Figure 5.1. Leg positive (top), negative (middle), and net (bottom) work over an entire stride for subjects using the BiOM powered (unaffected leg (UL) is black, affected leg (AL) is grey), and passive-elastic ESAR (UL is white, AL is hashed) prosthesis during walking at 1.25 m/s at a range of slopes. * indicates a significant difference in AL work between the BiOM powered and passive-elastic ESAR prosthesis.

We found no effect of prosthetic foot type on UL ankle or hip positive, negative, or net work ($p>0.05$, Figure 5.2). However, we did find a effect of prosthetic foot type on AL positive ankle work at uphill slopes of +6° and +9° ($p<0.05$, Figure 5.2) and on AL net ankle work at all slopes

($p \leq 0.001$, Figure 5.2). AL positive ankle work increased 89%, and 55% with use of the BiOM compared with a passive-elastic prosthesis at +6°, and +9°, respectively (Figure 5.2). Additionally, at +3, we found a trend for AL positive ankle work to be 44% greater with the use of the BiOM compared to an ESAR prosthesis ($p=0.0575$, Figure 5.2). AL net ankle work was greater (i.e. more positive) at all slopes with use of the BiOM compared with a passive-elastic prosthesis. Specifically, at downhill slopes of -9°, -6°, and -3° AL net ankle work was 94%, 109% and 155% more positive, respectively, with use of the BiOM compared to a passive-elastic prosthesis (Figure 5.2). At 0°, +3°, +6°, and +9°, AL net ankle work increased 350%, 339%, 674%, and 942%, respectively with use of the BiOM compared to a passive-elastic prosthesis (Figure 5.2). In other words, the BiOM produced almost ten times as much net ankle work as a passive-elastic prosthesis at the steepest uphill slope of +9°. We also found a significant effect of prosthetic foot type on AL negative ankle work at -9°, -3°, +6° and +9° ($p<0.05$, Figure 5.2). Specifically, AL negative ankle work was 35-45% less negative with use of the BiOM compared to a passive-elastic prosthesis (Figure 5.2).

We did not find a effect of prosthetic foot type on AL hip positive, negative, or net work ($p>0.05$, Figure 5.2). Similarly, and in support of our fifth hypothesis, we found no effect of prosthetic foot type on UL or AL knee positive, negative, or net work ($p>0.05$, Figure 5.2).

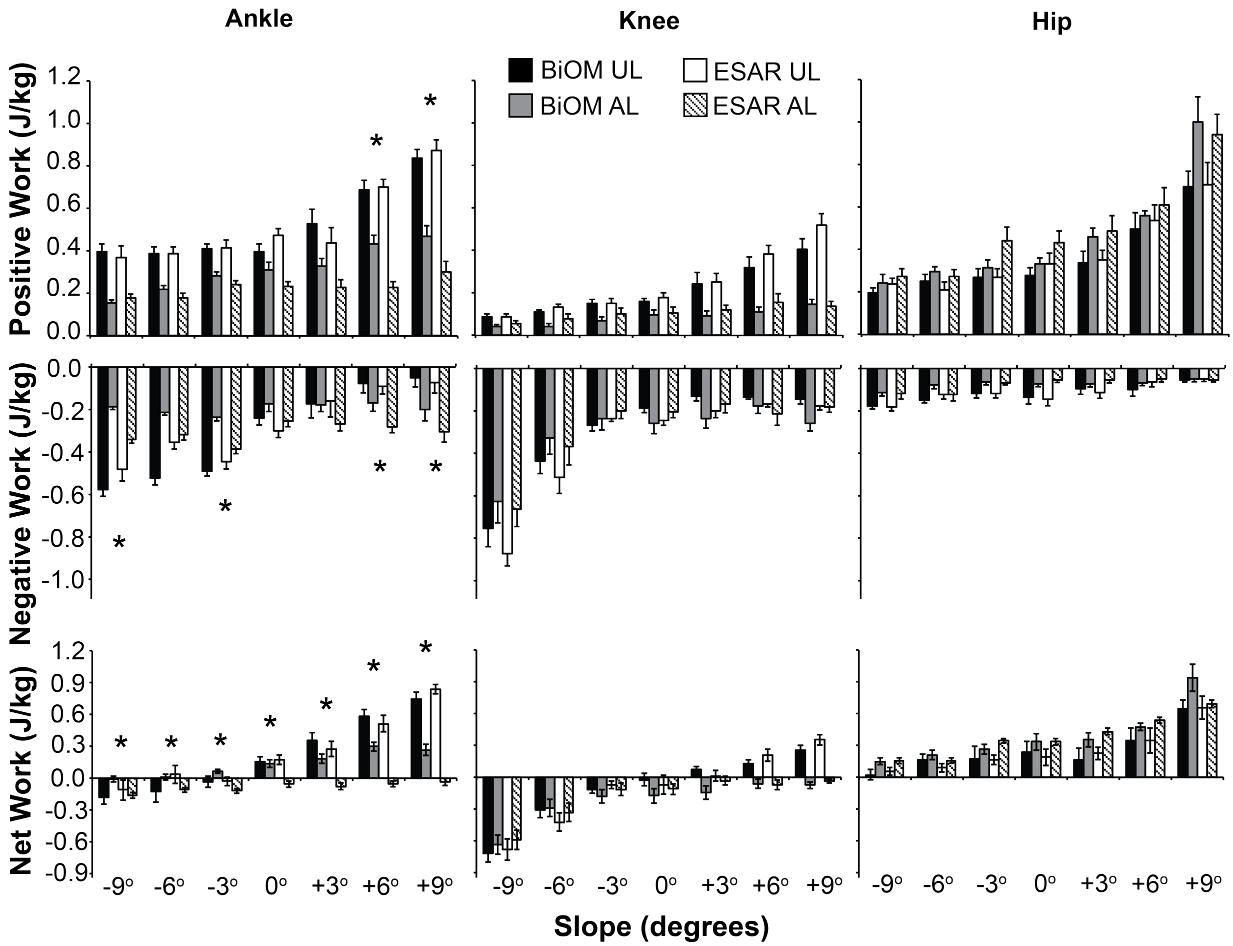

Figure 5.2. A) Ankle, B) Knee, and C) Hip positive (top), negative (middle), and net (bottom) work over an entire stride for subjects using the BiOM powered (unaffected leg (UL) is black, affected leg (AL) is grey), and a passive-elastic ESAR (UL is white, AL is hashed) prosthesis during walking 1.25 m/s at a range of slopes. * indicates a significant difference in AL work between use of the BiOM powered and passive-elastic ESAR prosthesis.

5.5 Discussion

The net work produced about a joint can be calculated by 1) the product of joint moment and joint angle, or 2) the integral of joint power with respect to time. Thus, by increasing only the peak positive moment, range of motion, or peak positive power, net work will increase. We integrated sagittal plane joint power with respect to time to determine joint work. Power, and thus the net work done at the ankle, knee, and hip, was more positive when walking uphill and more negative when walking downhill when subjects used either prosthesis (Figure 5.2). Similar to non-amputees [11, 71], peak hip and ankle power increased with increasing slope (Figures 5.3 and 5.5) and the magnitude of negative knee power increased with decreasing slope (Figure 5.4) when subjects used either prosthesis. Furthermore and similar to Winter and Sienko [14], subjects exhibited little to no knee power in the first half of stance (Figure 5.4).

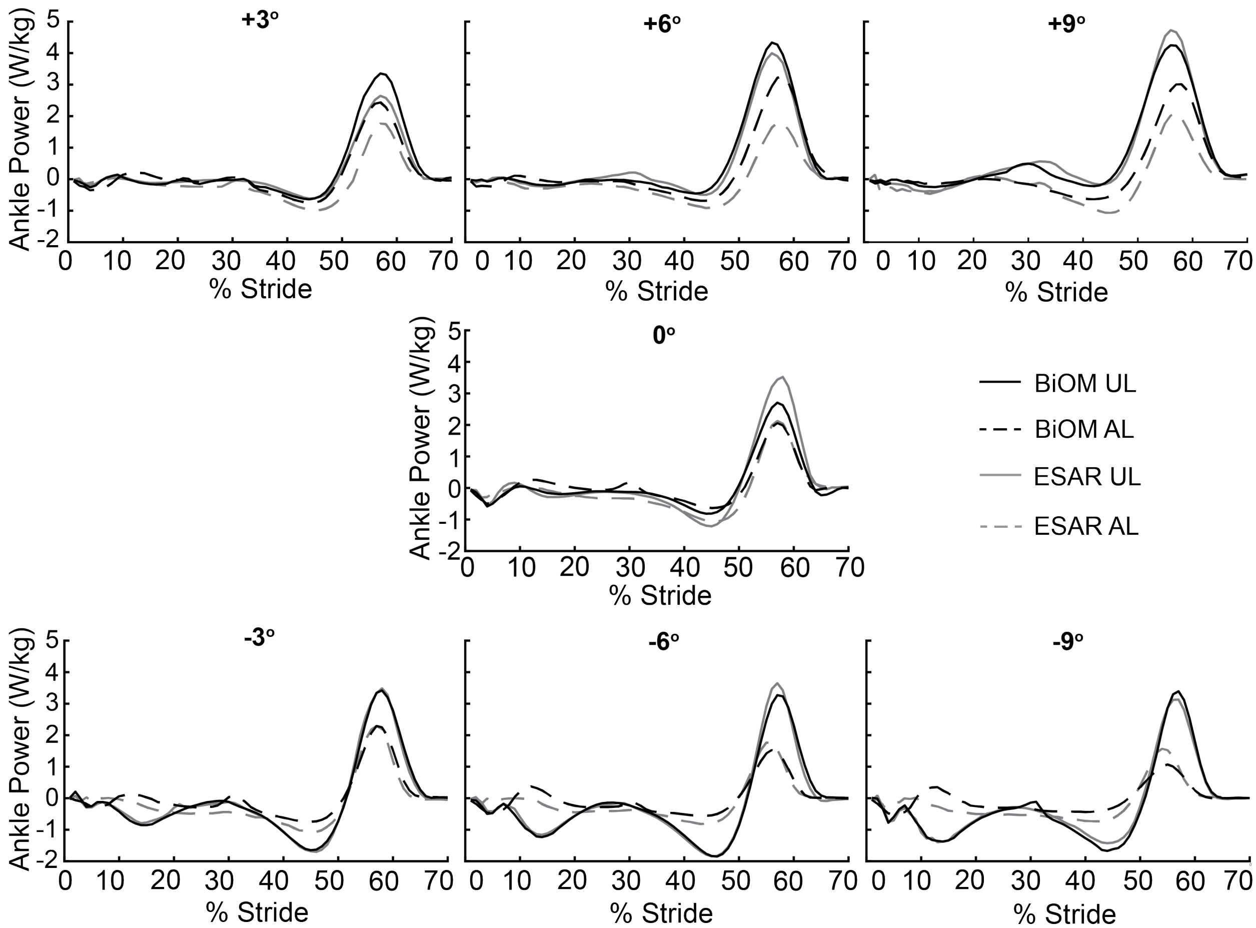

Figure 5.3. Affected leg (AL) prosthetic and unaffected leg (UL) ankle power (ensemble average) of subjects using the BiOM powered (black) and passive-elastic ESAR (grey) prosthesis during walking 1.25 m/s on uphill slopes (top), level ground (middle), and downhill slopes (bottom). Solid lines represent subjects' UL data and dashed lines represent subjects' AL data.

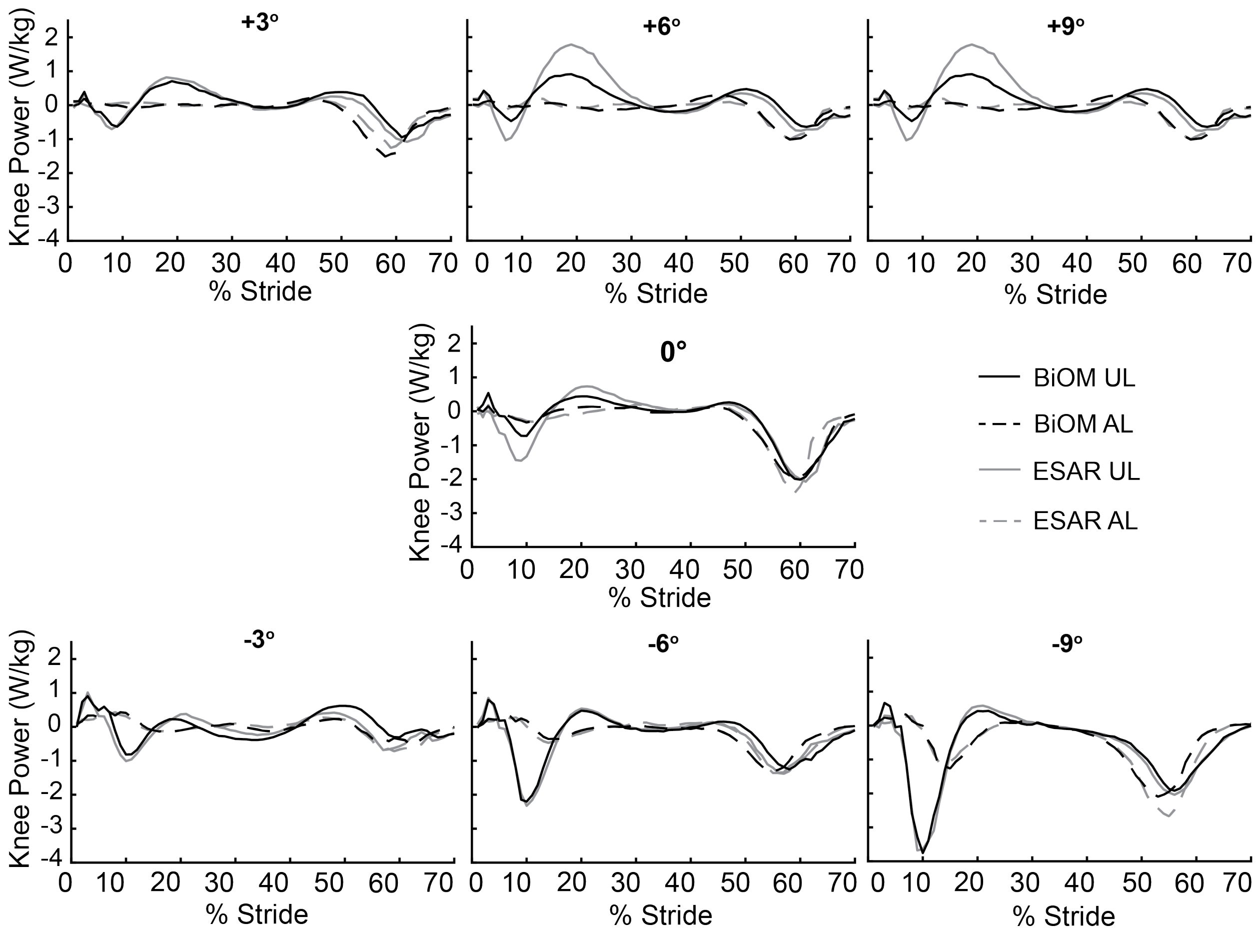

Figure 5.4. Knee power (ensemble average) of subjects using the BiOM powered (black) and passive-elastic ESAR (grey) prosthesis during walking 1.25 m/s on uphill slopes (top), level ground (middle), and downhill slopes (bottom). Solid lines represent subjects' UL data and dashed lines represent subjects' AL data.

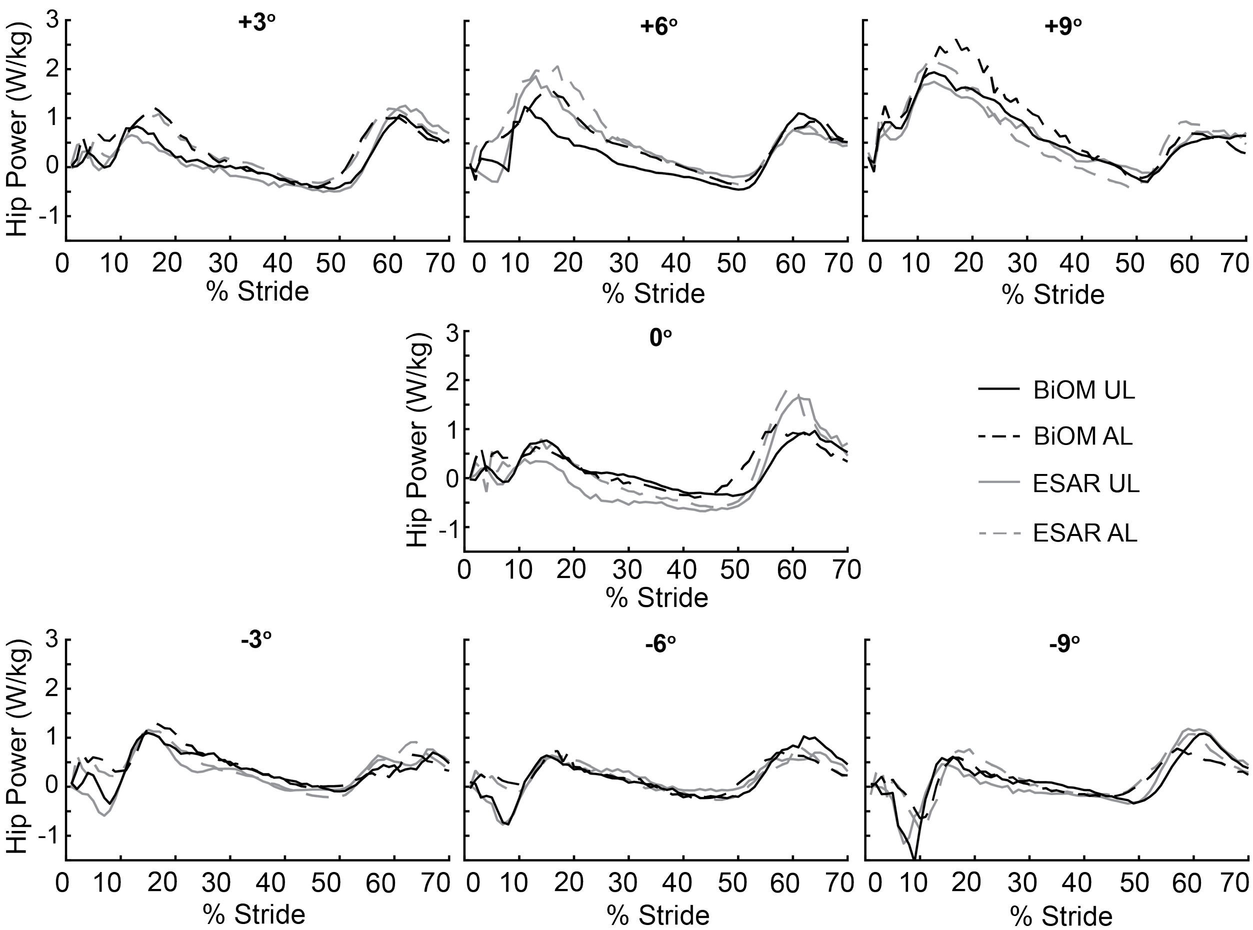

Figure 5.5. Hip power (ensemble average) of subjects using the BiOM powered (black) and passive-elastic ESAR (grey) prosthesis during walking 1.25 m/s on uphill slopes (top), level ground (middle), and downhill slopes (bottom). Solid lines represent subjects' UL data and dashed lines represent subjects' AL data.

In contrast to our first hypothesis, we found no UL net work change when using the BiOM compared to a passive-elastic prosthesis. Our second hypothesis that AL total positive work and net work would increase with use of the BiOM compared to a passive-elastic prosthesis was only partially supported. We did not find an effect of prosthetic foot type on AL positive work. However, we did find that AL net work was more positive (i.e. increased) at uphill slopes of +6° and +9° with use of the BiOM compared to a passive-elastic prosthesis. We calculated net work as the sum of positive and negative ankle, knee, or hip joint work and summed all joints to determine leg work. Thus, changes in net work were due to either changes in positive or negative work, or a combination of the two. Though not independently significant, the shift of AL net work to more positive at uphill slopes of +6° and +9° was due to both an increase in AL positive work and a decrease in AL negative work.

Our third hypothesis was unsupported because we found no significant effect of prosthetic foot type on UL ankle or hip positive or net work. Further, our fourth hypothesis that AL ankle positive and net work would increase and AL hip positive and net work would decrease with use of the BiOM compared to a passive-elastic prosthesis was partially supported. There were no significant differences in AL hip positive or net work between prosthetic foot types. However, in support of our fourth hypothesis, AL positive ankle work increased with use of the BiOM compared with a passive-elastic prosthesis at uphill slopes of +6° and +9°. In further support of our fourth hypothesis, AL net ankle work was more positive at all slopes with use of the BiOM compared to a passive-elastic prosthesis. In fact, not only did use of the BiOM provide greater prosthetic ankle net work than a passive-elastic prosthesis but at the steepest uphill slope of +9°, the BiOM prosthetic ankle net work was almost ten times greater than that of a passive-elastic prosthesis. At the steepest uphill slopes of +6° and +9° where we found significantly more AL positive ankle work, we also found a significant increase in AL net work with use of the BiOM compared to a passive-elastic prosthesis. It is possible that at moderate uphill and downhill slopes the substantial increase in AL prosthetic ankle positive and net work with use of the BiOM was absorbed in the knee or in the socket-limb interface and thus did not translate into significantly higher AL net work.

Finally, in support of our fifth hypothesis, UL and AL sagittal plane knee work remained unchanged with use of the BiOM compared to a passive-elastic prosthesis. This result is in line with Postema et al. [90], who found no difference in sagittal plane AL knee range of motion with the use of ESAR prostheses compared to conventional solid-ankle cushioned heel prostheses. However, this result is incongruent with Isakov et al. [35], who found an increase in sagittal plane AL knee flexion at heel strike compared to subjects' UL and an increase in AL biceps femoris activation during the first half of the stance phase compared to the UL, indicating greater work absorption by the AL.

We iteratively tuned the BiOM prosthesis to match the average of ankle sagittal plane range of motion, peak moment, peak power, and net work from twenty non-amputees at each slope [83]. Though subjects used the same tuning parameters, prosthetic components, and alignment that were established in the tuning session for all experimental sessions, subjects may have modified the way they walked while using the BiOM during the final experimental session compared to the tuning session. It is possible that after acclimation to walking while using the powered prosthesis, the tuning parameters should be further adjusted as the wearer modifies his or her gait. Previous studies that have analyzed the use of the BiOM prosthesis on level ground or up a 5° incline were completed in fewer experimental sessions (1-3 sessions in previous studies versus 6 sessions for the present study), used over-ground measurements, and relied on data collected from the BiOM's on-board microprocessor rather than from independent treadmill-based motion capture and ground reaction force data [17, 80]. Furthermore, the previous studies used force plates mounted in a walkway to collect ground reaction force data [17, 80] and thus force plate targeting while walking could attribute to differences between our data and those of

Herr and Grabowski and Esposito et al. [17, 80]. The difference in tuning strategies and protocol length may also potentially explain the differences in our results compared to others. Based on our ankle joint mechanics data (Figures 5.2-5.3), we were unable to match the BiOM prosthetic ankle net work to UL ankle net work on the final day of our protocol. However, we still found a significant increase in AL net ankle work at all slopes and a significant increase in AL positive ankle work at uphill slopes of +6° and +9° with use of the BiOM compared to a passive-elastic prosthesis. Based on previous studies of individuals with impaired or no ankle function [14, 79, 82, 92, 93], a reduction in ankle push-off work results in a slower preferred walking speed, increased gait asymmetry, increased metabolic energy expenditure, and increased AL hip positive work production. Thus, by increasing AL ankle positive and net work, persons with a transtibial amputation using a powered ankle-foot prosthesis may increase preferred walking speed, improve gait symmetry, decrease metabolic demand, and decrease reliance on the work performed by the AL hip joint and UL. The changes in leg joint mechanics when using a powered compared with a passive-elastic prosthesis during walking may therefore result in improved functional mobility and thus quality of life especially when navigating uphill slopes [73-75].

We found that use of the BiOM powered compared to passive-elastic prosthesis increased AL positive prosthetic ankle work, AL net prosthetic ankle work and AL net work. Furthermore, at all slopes ankle net work was greater with use of the BiOM powered compared to a passive-elastic prosthesis, but there were no differences in hip or knee positive or net work for either leg. Previous modeling and experimental studies [79, 82] have found that people with a unilateral transtibial amputation walking while using a passive-elastic prosthesis compensate for reduced ankle push-off work with increased UL and AL hip positive work. In the current study, we found that prosthetic ankle positive and net work increased, but UL and AL hip work did not change. Thus, it is possible that the increased prosthetic ankle push-off work provided by the BiOM was not enough for the user to overcome the compensation strategy typically adopted by individuals with a transtibial amputation using passive-elastic prostheses or that our subjects did not utilize the compensation strategy – as evidenced by no change in hip work though affected leg ankle positive and net work increased with use of the BiOM – seen in previous studies when using a passive-elastic prosthesis during walking [79, 82, 94]. The unchanged UL and AL hip power exhibited by our subjects could also be due to the type of passive-elastic prosthesis used by each subject or to the activity level of the subjects. The type of passive-elastic prosthesis used by subjects in Silverman et al., and Adamczyk & Kuo is not clear [79, 82]. But, in an often-cited study regarding compensatory strategies adopted by persons with a leg amputation during walking, the subjects used prostheses (solid ankle cushioned heel) that were not designed to restore push-off energy to the wearer [14]. These studies also included some subjects who had a leg amputation due to vascular disease [14, 79]. Persons who undergo a leg amputation due to vascular disease as opposed to a traumatic or congenital amputation typically require much higher metabolic energy to walk at the same speeds as non-amputees on level ground and have a slower preferred walking speed [88]; both of which could be attributed to the redistribution of positive push-off work from the ankle to the hip, similar to elderly populations [95].

Another possible explanation for the increase in AL and prosthetic ankle net work, but no change in UL or AL hip work with use of the powered compared to a passive-elastic prosthesis, could be that the provided prosthetic ankle push-off work is similar to the work provided by the uni-articular soleus, rather than the bi-articular gastrocnemius because an ankle-foot prosthesis does not span the knee. One previous musculoskeletal modeling study [96] predicts that horizontal trunk propulsion/acceleration is primarily provided by the soleus and rectus femoris during late stance in non-amputees walking on level ground. However, in experimental studies of non-amputees that measured muscle activation of the plantar-flexors during level-ground walking, ankle push-off work was primarily due to medial gastrocnemius activation, while the soleus played a small role [23, 95]. Therefore, it is possible that a powered ankle-foot prosthesis that only replaces the function of the uni-articular soleus is incapable of fully replicating biological ankle function in walking. Future studies should investigate the mechanical energy loss and transfer from the prosthetic ankle to the residual limb, and investigate the role of the socket-limb interface in this energy transfer.

Use of a powered prosthesis decreases frontal plane knee moments, which have been associated with knee osteoarthritis, in the UL during walking on level ground compared to use of a passive-elastic prosthesis [33] [97]. We calculated sagittal plane joint power, but did not include contributions to or changes in frontal or transverse plane powers. It is possible that the reduction in frontal plane knee moments is still observed when using the powered prosthesis to walk uphill and downhill. Future studies should investigate frontal plane joint moments when persons with a unilateral transtibial amputation walk uphill and downhill using passive-elastic and powered prostheses. Further, while we attempted to match prosthetic sagittal plane ankle biomechanics to UL sagittal plane ankle biomechanics by adjusting tuning parameters in the BiOM powered prosthesis, we were not able to match values at all slopes, and the prosthetic ankle work and peak power achieved during tuning were not observed on the final day of data collection. Future studies should investigate the effects of systematically varying powered ankle-foot prosthetic tuning parameters on the biomechanics of level-ground and sloped walking to determine the effects of tuning. Further, future studies should determine appropriate acclimation times and the effects on tuning. Acclimation could influence the way current devices are tuned to an individual patient in the clinical setting and thus the efficacy of using a powered ankle-foot prosthesis during daily activities.

5.6 Conclusion

Previous studies of persons with a leg amputation using passive-elastic and powered prostheses have primarily focused on level-ground walking. We found that with use of the BiOM powered compared to a passive-elastic prosthesis, affected leg sagittal plane prosthetic ankle positive work increased at uphill slopes of +6° and +9° and net prosthetic ankle work increased at all slopes (-9° to +9°). Similarly, with use of the BiOM powered compared to a passive-elastic prosthesis, affected leg net work increased at uphill slopes of +6° and +9°. There were no differences in affected leg knee or hip positive or net work nor did unaffected joint or leg work change when using the BiOM powered compared to a passive-elastic prosthesis. Greater prosthetic ankle positive and net work through use of a powered prosthesis could improve symmetry between the legs of people with a transtibial amputation and thus improve preferred walking speed, functional mobility and quality of life.

6 CONCLUSION

In my dissertation, I contributed to our understanding of the basic biomechanics and metabolic costs of walking on uphill and downhill slopes for people with and without a leg amputation. Specifically, I found that, for healthy adults without a leg amputation, individual leg mechanical power during step-to-step transitions accounts for ~65% of the metabolic power needed to walk at 21 combinations of slope and velocity, and that quadratic models best described the relationships between each leg's mechanical power and whole body metabolic power. Then, I quantified the biological ankle joint's contribution to total individual leg positive and negative work on a range of slopes and speeds and found that the muscles surrounding the ankle joint are responsible for between 47% and 116% of individual leg positive and negative work, indicating the importance of the ankle joint to maintain steady-state walking at various speeds and slopes. Thus, replicating biological ankle function over a range of speeds and slopes with prosthetic or assistive devices will require dynamic changes in positive and negative ankle work, as well as peak moments, powers, and range of motion of the prosthetic ankle.

When persons with a unilateral transtibial amputation walk on level ground with a passive prosthesis, the mechanical energy requirements are the same as non-amputees but they lack the ability to generate power at the prosthesis. The lack of, or reduction in, ankle push-off work results in approximately 10-30% higher net metabolic costs to walk at the same speed as non-amputees. Powered ankle-foot prostheses have been developed that normalize the metabolic demand required, preferred walking speed, and knee external adduction moments during level-ground walking for persons with a leg amputation walking at a range of speeds. I investigated how these powered prostheses affect the metabolic demand, step-to-step transition work, individual leg, and joint work of people with a transtibial amputation walking at a range of uphill and downhill slopes.

The use of a powered ankle-foot prosthesis reduced the net metabolic power required to walk up a +3° and +6° slope and trended lower at +9° but did not change the net metabolic power required to walk on level ground or on downhill slopes of -3°, -6°, and -9° compared with use of a passive-elastic prosthesis. Use of the powered ankle-foot prosthesis also improved leg net work symmetry at uphill slopes of +6° and +9°, potentially highlighting the limitations of passive-elastic prostheses to accommodate walking up steeper slopes. Similarly, affected leg sagittal plane prosthetic ankle positive work and affected leg net work significantly increased at uphill slopes of +6° and +9° with use of the powered compared to a passive-elastic prosthesis. Furthermore, net prosthetic ankle work increased at all slopes with use of the powered compared to a passive-elastic prosthesis. Contrary to previous studies of persons with a leg amputation walking on level ground, I found no differences in affected leg knee or hip positive or net work, nor did unaffected joint or leg work change when using the powered compared to a passive-elastic prosthesis.

Previous studies of persons with a leg amputation using passive-elastic and powered prostheses have primarily focused on level-ground walking. This set of studies in my dissertation is the first to document the role of the biological ankle in sloped walking over a range of speeds and to show how use of passive-elastic and powered ankle-foot prostheses affects metabolic cost, individual leg, and joint biomechanics during walking on a range of uphill and downhill slopes.

These results provide a foundation for the design and control of powered leg assistive devices for robust constraints such as varied slopes and velocities during walking. The development and implementation of such devices has the potential to restore normative metabolic costs, leg symmetry, and normative biomechanics for users. Furthermore, use of powered ankle-foot prostheses has the potential to increase quality of life, level of functional mobility, and reduce comorbidities (such as knee osteoarthritis and low back pain) commonly incurred by persons with no ankle function. This dissertation provides insight into the effects of amputation and prosthetic foot type on walking and possibly provides support for the prescription of powered devices for all individuals with a leg amputation.

7 PUBLICATION INFORMATION

All chapters of my dissertation either have been published or are in preparation for publication. Chapter 1, *The Correlation Between Metabolic and Individual Leg Mechanical Power During Walking at Different Slopes and Velocities*, was published in the Journal of Biomechanics in 2015. Chapter 2, *The Ankle Joint's Contribution to Individual Leg Work During Uphill and Downhill Walking Over a Range of Speeds*, has been submitted to the Royal Society Open Science and is under review. We expect publication of Chapter 2 in 2017. Chapter 3, *Biomechanical and Metabolic Effects of Using a Powered Compared to Passive-Elastic Ankle-Foot Prosthesis During Uphill and Downhill Walking*, is being prepared for submission to the Journal of Applied Physiology in 2017. Chapter 4, *Effects of Passive and Powered Ankle-Foot Prostheses on Leg Joint Work During Sloped Walking*, is being prepared for submission to Frontiers in Bioengineering and Biotechnology in 2017.

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