# SHANK KINEMATICS AND KINETICS IN PROSTHETIC GAIT: IMPLICATIONS FOR IMPROVED DESIGN OF PROSTHETIC SYSTEMS

by

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# SHANK KINEMATICS AND KINETICS

## IN PROSTHETIC GAIT:

### IMPLICATIONS FOR IMPROVED DESIGN

### **OF PROSTHETIC SYSTEMS**

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#### ABSTRACT

Humans ambulate with bipedal gait and the ankle-foot system is imperative for efficient and symmetrical gait. Individuals with lower-limb amputation lose the anklefoot system and must rely on prostheses to ambulate. Prosthetic systems have traditionally been designed to mimic select aspects of the natural ankle-foot complex, such as roll-over shape and the late-stance ankle power burst. However, as prosthetic users still do not reach the same level of function as individuals without amputation it is clear that there is room for improvement in current prosthetic systems. In other words, there may be other aspects or features of lower limb mechanics during gait that could hold the key to enabling meaningful improvements in prosthesis design to be achieved. In particular, recent research has identified important, and somewhat surprising, features of the natural shank's segmental kinematics and kinetics that may be useful as design criteria for future prosthetic systems. However, before advancements can be made in the designs of prosthetic systems, we must understand the shank segmental kinematics and kinetics during the gait of users with current prosthetic ankle-foot systems. Thus the aim of this study was to characterize the segmental kinematics and kinetics of the residual shank in prosthetic gait and compare to typical gait.

Shank segmental kinematics and kinetics in overground gait were analyzed for four individuals with unilateral transtibial amputation who used the same energystoring-and-returning prosthetic ankle-foot system. The kinematic results revealed that the proximal shank remained horizontal throughout stance in prosthetic gait, similar to typical gait, despite the prosthetic users' lack of active plantar flexion. Kinetic results showed that power flowed into the shank from more proximal segments at the moment

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of push-off on the prosthetic side, as opposed to typical gait and on the intact side where power flowed out of the proximal shank. In addition, the velocity of the proximal shank was higher at push-off on the prosthetic side compared to the intact side or typical gait. These results indicate that, to compensate for the lack of active push-off in the prosthesis, these prosthetic users use more proximal structures to lift the foot off the ground in late stance, instead of actively pushing it off. Analysis of spatiotemporal parameters revealed a shorter stance time and longer step length on the prosthetic side, where, presumably, the prosthetic users terminate the stance phase early to avoid the lowering and downward acceleration of the proximal shank. These results give better insight into the shortcomings of current prosthetic ankle-foot systems and provide design criteria that may be used to improve the design of prosthetic ankle-foot systems and resulting gait function of the prosthetic user.

#### Chapter 1

#### INTRODUCTION AND SPECIFIC AIMS

#### 1.1 Background

#### 1.1.1 Typical gait

Humans locomote with bipedal gait. In gait, the lower extremities move in cyclical motion where one foot swings forward in order to advance the body's center of mass, while the other foot is in contact with the ground, supporting the body. Those different periods are referred to as the swing phase and stance phase, respectively. Each cycle of one foot moving through initial contact with the ground (heel strike), supporting and propelling the body while the other foot swings forward (midstance), pushing off from the ground (toe off) and finally swinging forward and preparing to step down for another round of these events is referred to as the gait cycle (Figure 1). The stance phases of the two lower limbs overlap, so during the loading response at the beginning of stance for each side as well as terminal stance, both feet are on the ground. This period is known as the double-support phase or double stance (Perry, 1992).



Figure 1: Gait cycle, the figure shows stance and swing phase, the main events that take place during stance and the rockers of the stance phase.

The stance phase is typically divided into four rockers i.e. the heel, ankle, forefoot and toe rockers (Perry, 1992; Perry & Burnfield, 2010), referring to the structure over which the shank is rotating at each time (Figure 1). The heel rocker takes place over the period between heel strike and foot flat, during which the body weight is transferred from the trailing leg onto the leading leg, referred to as loading response (Perry, 1992). The ankle rocker follows, which is the period called midstance. The foot on the stance leg is flat on the ground, supporting the body as the other leg swings forward. The period between heel rise and maximal dorsiflexion in the ankle of the stance leg is the forefoot rocker, and finally, during terminal stance from maximal dorsiflexion to toe off, is the toe rocker (Figure 1) (Perry, 1992).

#### 1.1.2 The natural ankle-foot complex

One of the most complex and intricate structures in the human musculoskeletal system is the ankle-foot complex. The natural ankle and foot are comprised of numerous bones and soft tissues which serve many important biomechanical functions during gait, most specifically during stance (Figure 2).



Figure 2: The natural ankle-foot complex; figure licensed from https://stock.adobe.com/.

The main function of the ankle-foot complex is to serve as the body's contact with the underlying surface during gait and enable the upper body to progress over it in a controlled and efficient manner (Kepple, Siegel, & Stanhope, 1997; Neptune, Kautz, & Zajac, 2001; Perry & Burnfield, 2010). The ankle-foot complex must provide stability and proper kinematics to enable knee and hip extension during stance (Owen, 2010; Perry & Burnfield, 2010; Sutherland, Cooper, & Daniel, 1980), the forward movement of the pelvis and trunk (Owen, 2010) and anterior progression of the center-of-pressure (Francis, Lenz, Lenhart, & Thelen, 2013; Sutherland et al., 1980). Through interaction of foot-to-floor and ankle angle excursion over stance, the trajectory of the knee remains nearly horizontal in the sagittal plane throughout stance (Pollen, 2015). Through these kinematic patterns, the individual is able to achieve the optimal step length and necessary position of the limb for transitioning into the next step (Fatone, Gard, & Malas, 2009; Sutherland et al., 1980).

Over the stance phase of gait, the foot acts as both soft, compliant structure and a rigid lever (Ren, Howard, Ren, Nester, & Tian, 2008). The bones and soft tissues of the foot deform right after heel-strike to adapt to the underlying surface and provide shock absorption during weight acceptance (Boonpratatong & Ren, 2010; Ren, Howard, et al., 2008). The foot stiffens throughout the gait cycle to act as a rigid lever to provide body support and propulsion during mid- to late stance (Francis et al., 2013; Kepple et al., 1997; McGowan, Neptune, & Kram, 2008; Neptune et al., 2001). As soft tissue structures in the foot stretch over midstance, they store energy that is later released during push-off (Ren, Howard, et al., 2008). In typical gait, the ankle musculature cycles through phases of power absorption and generation during stance. In early to midstance the plantar flexors, the gastrocnemius and soleus muscles, control forward rotation of the shank over the ankle through eccentric contraction, and in late stance they contract concentrically to provide a large burst of positive power resulting in the foot pushing off the ground and initializing the swing phase (Robertson & Winter, 1980; Winter, 1983, 2009). In fact, the ankle has the highest power output of all the joints in the body (Huang, Shorter, Adamczyk, & Kuo, 2015) and the power generated in the latter half of stance is over five times the amount that was absorbed in early stance (Winter, 1983). An important factor in the power absorption and generating function of the natural ankle-foot complex is the quasistiffness properties of the ankle. The ankle changes stiffness properties over the gait

cycle and behaves as a linear and non-linear spring in turn to allow for power absorption or generation (Versluys et al., 2008). The stiffness properties also change with walking speed, where stiffness increases with increasing walking speed (Hansen, Childress, Miff, Gard, & Mesplay, 2004).

Another interesting property of the natural ankle-foot system in gait is the rollover-shape (ROS). ROS describes the shape the foot, ankle-foot system or knee-anklefoot system conforms to during stance (Figure 3) (Hansen & Childress, 2010; Hansen, Childress, & Knox, 2004).



Figure 3: Roll-over shape (ROS), the shape the foot conforms to during the first three rockers of stance.

In this approach, the stance limb and body are modelled as an inverted pendulum "rolling" over the respective structures. ROS is calculated over the period from heel strike to opposite heel strike (Hansen, Childress, & Knox, 2004) which corresponds to the first three rockers of stance (Perry, 1992; Perry & Burnfield, 2010). The ROS is calculated by finding the center of pressure (COP) under the stance foot and transforming it from a laboratory based coordinate system to the foot, ankle-foot or ankle-hip based coordinate systems (Hansen, Childress, & Knox, 2004). The radius of curvature of the ROS has been shown to be around 30% of leg length in non-disabled individuals (Figure 3) and it is invariant with changing conditions, such as different walking speed, added weight and change in footwear (Hansen & Childress, 2010; Hansen, Childress, & Knox, 2004).

#### 1.1.3 Limb loss

Limb loss, the partial or complete absence of a limb, is a condition that is estimated to affect around 2 million people in the United States alone today (Ziegler-Graham, MacKenzie, Ephraim, Travison, & Brookmeyer, 2008). Limb loss can be due to a number of injuries or afflictions, most commonly trauma or dysvascular conditions that result from diseases such as diabetes mellitus or peripheral artery diseases (National Limb Loss Information Statistics, 2007; Ziegler-Graham et al., 2008). A small percentage of limb loss is due to cancer or congenital defects (Ephraim, Dillingham, Sector, Pezzin, & MacKenzie, 2003; National Limb Loss Information Statistics, 2007). With the aging population and improvements in healthcare that lead to subsequent increase in individuals living with diseases that can affect the circulation (Dall et al., 2013; Shaw, Sicree, & Zimmet, 2010), the number of individuals living with amputation has increased in the past few decades and is projected to be over 3.5 million people in the year 2050 (Ziegler-Graham et al., 2008). Individuals with a lower limb amputation typically use prostheses to ambulate. Many prosthetic users have impaired gait patterns and lower self-selected walking speed than individuals without lower limb amputation (Isakov, Keren, & Benjuya, 2000; Nolan et al., 2003). Decreased walking speed can be used as an indicator for functional assessments and the likelihood of hospitalization in many populations (Middleton, Fritz, & Lusardi, 2015; Studenski et al., 2003). Increasing walking speed can lead to greater activity levels and community participation and has been linked to better quality of life in other unilaterally affected groups such as stroke survivors (Schmid et al., 2007). Therefore it is important to optimize function in individuals with amputation for their increased participation and function in society and quality of life.

#### **1.1.4 Prostheses**

Today, individuals with a lower limb amputation are typically prescribed prosthetic ankle-foot systems that aim to replicate, and return, the functions of the natural ankle-foot. Until about 40 years ago, most prostheses were designed mainly to provide body support to enable the user to get around with little regard to restoring the lost ankle-foot function. Prosthetic design has dramatically improved over the past few decades, with increasing emphasis on mimicking the function of the natural ankle-foot complex, particularly recreating the roll-over-shape (ROS) of the human foot and gaining a part of the energy burst from the plantar flexors during push-off (Curtze, Otten, Hof, & Postema, 2011; Hafner, Sanders, Czerniecki, & Fergason, 2002b; Hansen & Childress, 2010). One of the earliest prostheses designed to attempt the replication of ankle function was the Solid-Ankle Cushioned Heel (SACH) foot (Versluys et al., 2009). The SACH foot is simple in design and is inexpensive but fails to replicate the function of the natural ankle-foot complex (Versluys et al., 2009). In the early 1980s, Energy-Storage-and-Return (ESR) feet were introduced (Figure 4). Typical ESR prostheses are keels made of carbon fibers that can store elastic energy when they are compressed in early- to midstance, and return it when released at pushoff, thus supposedly replicating the energy return properties of the ankle musculature (Hafner et al., 2002b; Ren, Howard, et al., 2008; Versluys et al., 2009). ESR prostheses are generally preferred over the SACH feet by prosthetic users as they offer perceived improvements in comfort and performance (Hafner, Sanders, Czerniecki, & Fergason, 2002a) and trends towards increased self-selected walking speed and lower rate of perceived exertion (Hafner et al., 2002a; Hsu, Nielsen, Lin-Chan, & Shurr, 2006). ESR prostheses are the most commonly used types of prostheses today.



Figure 4: Transtibial prostheses with ESR feet; Image licensed from https://stock.adobe.com/.

To provide even greater gain in power production to the prosthetic system, powered (active) prostheses have been developed in the last few years (Cherelle, Mathijssen, Wang, Vanderborght, & Lefeber, 2014; Versluys et al., 2009). While many such prostheses are still in the developmental stage, a few are already being sold commercially. Powered prostheses have a motor that, either through pneumatic or electrically driven methods, enables the prosthesis to "plantar flex" and add power to the system in late stance, thus more closely mimicking the late-stance power generation that is seen in the natural ankle-foot complex (Versluys et al., 2009; Winter, 1983). Those powered prostheses have been shown to increase self-selected walking speed (Gates, Aldridge, & Wilken, 2013; Herr & Grabowski, 2012) and lower metabolic energy cost and mechanical work during walking (Herr & Grabowski, 2012).

Despite recent advancements and design improvements, individuals with lower limb amputation still have lower levels of function than individuals without an amputation and still show asymmetric gait patterns and compensatory movements (Silverman et al., 2008; Versluys et al., 2009). Although ESR prostheses offer advantages over SACH feet, as there are no active components in the ESR prostheses, the carbon fiber keels cannot allow for a higher energy return than what was stored during early stance (Crimin, Mcgarry, Harris, & Solomonidis, 2014). And as for the powered prostheses, despite the advantages of active power input, most of these powered prostheses are bulky and have low power density (power to weight ratio) (Cherelle et al., 2014; Hitt, Sugar, Holgate, Bellman, & Hollander, 2009; Versluys et al., 2008). As a result there is considerable weight added to the distal end of the

amputated leg which can counteract the effects of the power input (Mattes, Martin, & Royer, 2000).

#### **1.1.5** Function in individuals with amputation

Lacking the natural ankle-foot musculature and other structures, individuals who have undergone a lower limb amputation or have congenital limb defects lose many of the important functions previously described, and despite recent advancements, even the most state-of-the-art prosthetic systems fail to completely recoup the lost function. Individuals with transtibial amputation show decreased pushoff power in late stance (Adamczyk & Kuo, 2015; Bateni & Olney, 2002) and increased activity in hip and knee extensors in early- to midstance (Silverman et al., 2008; Voinescu, Soares, Natal Jorge, Davidescu, & Machado, 2012; Winter & Sienko, 1988). Individuals with amputation also have slower self-selected walking speeds (Isakov et al., 2000) and higher metabolic cost associated with ambulation (Genin, Bastien, Franck, Detrembleur, & Willems, 2008; Quesada, Caputo, & Collins, 2016; Schmalz, Blumentritt, & Jarasch, 2002; Waters, Perry, Antonelli, & Hislop, 1976). They show asymmetry in knee strength and loading in gait (Lloyd, Stanhope, Davis, & Royer, 2010) as well as asymmetric spatiotemporal parameters (Isakov et al., 2000; Nolan et al., 2003) and leg loading, with more stress being put on the intact leg (Gailey, Allen, Castles, Kucharik, & Roeder, 2008; Lloyd et al., 2010). These compensatory movements, inefficient energy consumption, and unequal loading can have detrimental effects on body structures and can lead to numerous secondary impairments, such as degenerative changes in the knee and hip on the sound side and

lower back that can develop into osteoarthritis and low back pain (Gailey et al., 2008; Kulkarni, Gaine, Buckley, Rankine, & Adams, 2005).

#### **1.2** Objectives and Specific Aims

As even the most recent prosthetic systems are still suboptimal and fail to fully return a prosthetic user to typical levels of function, shifting the focus from almost strictly attempting to mimic ROS to more comprehensive design criteria may help ultimately improve the design and functionality of lower limb prostheses. In particular, the kinematics and kinetics of more proximal segments such as the shank have been shown to be important factors for achieving natural lower extremity gait patterns and effective energy transfer (Owen, 2010; Pollen, 2015). However, before changes or advancements in prosthesis design criteria can be made, the kinematic and kinetic behavior of the shank in the gait of users of current below-knee prostheses need to be better understood. The shank kinematics and kinetics of typical gait have been documented (Buczek, Kepple, Siegel, & Stanhope, 1994; Owen, 2010; Pollen, 2015) but to the author's knowledge, the kinematics and kinetics of the shank during prosthetic gait have not been characterized to this extent before. Therefore the overall purpose of this study is to characterize and model the shank kinematics and kinetics of individuals with unilateral transtibial amputation during walking while wearing an ESR prosthesis.

# AIM 1: Characterize sagittal plane shank kinematics of individuals with transtibial amputation during the stance phase of gait.

First, the kinematics of shank progression (shank rotation, horizontal and vertical translation of the proximal end of the shank) during the stance phase of gait will be characterized from experimental data of four individuals with a transtibial amputation wearing an ESR prosthesis. Then, a shank kinematic model, previously developed for typical gait, will be verified for prosthetic users. The model will then be used to quantify contributions of the ankle and foot to shank progression in prosthetic gait. All data will be compared to typical gait.

<u>Hypothesis 1</u>: In the absence of active plantar flexion in prosthetic gait, there will be a greater shank rotation over stance and the proximal shank will lower excessively in late stance compared to typical gait.

# AIM 2: Characterize the shank kinetics of individuals with transtibial amputation during the stance phase of gait.

The kinetics of shank progression (shank segmental power) during the stance phase of gait will be characterized from experimental data of four individuals with a transtibial amputation wearing an ESR prosthesis. The segmental power will be calculated using a unified deformable (UD) model to calculate work and power in the prostheses and a 6-degree-of-freedom (6DOF) model to calculate work and power in residual joints. The segmental power and individual power components at the proximal shank at push-off will be specifically considered.

<u>Hypothesis 2</u>: There will be lower power output from the prosthesis compared to a natural ankle-foot complex.

#### Chapter 2

#### **AIM 1: SHANK KINEMATICS IN PROSTHETIC GAIT**

#### 2.1 Background

Much development has taken place in the field of prosthetic design over the last few decades with new concepts emerging that aim to increase the function and societal participation of prosthetic users. The prosthetic systems that have been developed in that time have mainly been designed with the aim of replicating only specific functions and properties of the natural ankle-foot complex. The specific functions that are typically mimicked are the damping effect right after heel-strike such as in the SACH foot (Versluys et al., 2009), roll-over-shape (ROS), which is the shape the foot conforms to during stance (Hansen & Meier, 2010; Hansen, Childress, & Knox, 2004), storage and return of the energy during stance to recreate some of the power burst at the end of stance, as in ESR feet (Versluys et al., 2009), as well as active plantar flexion and power input in terminal stance made available by the introduction of powered prostheses (Cherelle et al., 2014; Versluys et al., 2009). ROS has been an important design and assessment tool for these biomimetic prosthetic systems as it provides insight into the shape and behavior of the natural ankle-foot complex during gait (Hansen & Childress, 2010). One of the advantages of using ROS as a criterion is its invariance under various circumstances, loads and footwear (Hansen & Childress, 2010; Hansen, Childress, & Knox, 2004).

However, prosthetic users still do not reach the same levels of function as individuals without amputation (Adamczyk & Kuo, 2015; Genin et al., 2008; Isakov et al., 2000; Lloyd et al., 2010; Silverman et al., 2008; Voinescu et al., 2012; Winter & Sienko, 1988). Notably, ROS is only calculated over the first three rockers of stance

(Hansen, Childress, Miff, et al., 2004), thus it leaves out the toe rocker, during which many important ankle functions take place (Neptune et al., 2001; Perry & Burnfield, 2010). Focusing on ROS as a primary guideline for prosthetic design can lead to the disregard of more proximal segments, such as the shank, as well. Moreover, ESR feet cannot provide the power output of the natural ankle-foot complex, as the passive nature of the prosthesis means the device cannot return more energy than what was stored in it (Crimin et al., 2014). Although there is some passive conservation and return of energy in the natural foot structures (Ren, Howard, et al., 2008), there is also a considerable amount of power that is actively created in the ankle musculature (Robertson & Winter, 1980; Winter, 1983, 2009). The development of powered prostheses was meant to add energy into the system to better mimic natural gait. However, these active devices typically include batteries and actuators that can add mass to the distal end of the leg (Cherelle et al., 2014; Versluys et al., 2008), which changes the inertial properties and can thus counteract the added power from the prosthesis (Mattes et al., 2000).

Shank segmental kinematics may have important implications for prosthetic design. Shank progression, a function of the shank's sagittal plane rotation (shank angle) and translation (proximal end trajectory), likely plays an important role in achieving optimal step lengths, gait speed and an energetic cost of walking (Owen, 2010). Thus, shank kinematics may be the ideal characterization scheme for improved prosthetic design. It has been shown that replicating typical shank kinematics can improve the kinematics of more proximal segments and alignment of the ground reaction force relative to the knee and hip, an essential feature for minimizing the neuromuscular cost of walking (Owen, 2010). A recent analysis aimed to characterize

the ankle and shank kinematics in typical gait. In said analysis, a sagittal plane kinematic model for calculating shank progression was developed and verified for the gait of non-disabled individuals (Pollen, 2015). The results from the analysis are depicted in Figure 5a which show the sagittal trajectory of the shank segment during the stance phase. The horizontal axis shows the direction of forward progression and the vertical axis shows the height, which correspond to the y- and z-axes in the laboratory coordinate system, respectively. The straight, colored lines show the shank as it moves through the four rockers of stance and the red lines at the proximal (top) and distal (bottom) ends of the shank show the model predicted trajectory of the knee and ankle, respectively (Figure 5a). Results showed that through the interaction of ankle angle and foot-to-floor angle, the proximal shank maintained a fairly horizontal trajectory throughout stance (Figure 5a) (Pollen, 2015). The results also showed a strong agreement between kinematic data and the model outcomes, indicating the validity of the model (Pollen, 2015). To investigate the predictive properties of the model, it was used to mimic prosthetic gait by experimentally limiting the plantar flexion during the toe rocker of the gait data from non-disabled individuals, as most passive prosthetic systems do not plantar flex past neutral. The results predicted that without plantar flexion past neutral, the proximal end of the shank would lower considerably during the toe rocker while the shank over-rotated (Figure 5b) (Pollen, 2015).



a) Observed and Model Predicted Shank Trajectories (Typical)

Figure 5: Sagittal plane shank trajectory. The horizontal axis shows the position in the anterior-posterior direction (y-axis in the laboratory coordinate system) and the vertical axis shows the height (z-axis). a) Experimental and model predicted sagittal plane shank trajectory in typical gait (Pollen, 2015); b) Predicted sagittal plane shank trajectory during stance with ankle plantar flexion restricted past neutral in typical gait (Pollen, 2015). Written permission was obtained for the use of the graphs (see Appendix A).

The purpose of this aim was to experimentally evaluate the kinematics of shank progression in prosthetic gait and compare it to the model predictions from prior work. It was hypothesized that the proximal shank would lower and over-rotate at the end of stance as predicted by the limited plantar flexion model (T. Pollen, 2015). Understanding shank kinematics in prosthetic gait may facilitate the development of novel prosthetic systems that better replicate the shank kinematics in typical gait.

#### 2.2 Methods

#### 2.2.1 Data acquisition

Kinematic and kinetic data from four individuals with transtibial amputation (all males; ages  $45 \pm 3.16$  years; height  $1.76 \pm 0.07$  m; body mass  $91.4 \pm 24.8$  kg) were included in this study, drawn from an existing database. They were all high functioning, with an average self-selected walking speed of 1.15 m/s and all were able to ambulate without support on level ground. All used the Vari-Flex® ESR prosthesis (Össur, Reykjavik, Iceland) to ambulate in everyday life. Before beginning the data collection, all subjects signed an informed consent form (see Appendix B). For the data collection the subjects wore their prosthesis, a prosthetic cover and athletic shoes. To track lower extremity motion, reflective markers were placed on the subjects' bodies according to the six-degree-of-freedom method (Figure 6a) (Holden & Stanhope, 1998). The placement of the ankle calibration markers on the lateral malleolus had to be approximated on the prosthetic side based on measurements on the sound side as the anatomical landmarks were missing.



Figure 6: Visual 3d a) model and b) walking trial.

For each trial, the subjects were instructed to walk at their self-selected walking speed (1.15 m/s on average) that had been determined prior to testing using a 10-Meter-Walk-Test (10MWT) (Palmer, Matlick, & Council, 2015; RehabMeasures, 2014). They walked over a strain gauge force platform (AMTI, Watertown, MA) embedded along a straight-line walkway that collected kinetic data at a sampling rate of 1200 Hz. Kinematic data were collected using a six-camera motion capture system (Motion Analysis Corp., Santa Rosa, CA) at a sampling rate of 240 Hz. During the data collection, speed was measured via two photocell beams located three meters apart and after each trial, verbal feedback was provided to cue subjects towards the

targeted speed. Acceptable trials were ones where the subject walked within  $\pm 0.2$  standard deviations of the targeted speed and each foot was placed completely on one of the force plates (Figure 6b).

#### **2.2.2 Data analysis**

The collected data were processed and analyzed using Visual 3D (C-Motion Inc., Germantown, MD) and were low-pass filtered using a zero-lag fourth-order Butterworth filter, kinematic data at the cut-off frequency of 6 Hz and kinetic data at 25 Hz. A right-handed laboratory coordinate system was established such that the positive x-axis pointed to the subject's right, positive y-axis pointed towards the direction of forward progression and the z-axis pointed straight up (Figure 6b). To facilitate gait analysis the stance phase was divided into four rockers based on gait events during stance. The heel rocker was between heel strike and foot flat, which was defined as when the vertical velocity of the foot's distal end approached 0 m/s. The ankle rocker was between foot flat and heel off, defined as the moment when the knee was in maximal extension. This definition for heel off was based on the fact that the individuals with amputation lack active plantar flexion and thus in order to start the plantar flexion in push-off they must bend the knee. The forefoot rocker was between heel off and maximum ankle dorsiflexion and the toe rocker covered the push-off phase between the maximum dorsiflexion and toe off.

#### 2.2.3 Characterization of shank kinematics

Shank rotation and translation in prosthetic gait were characterized via the shank angle in the laboratory coordinate system and horizontal and vertical translations of the proximal end of the shank, as measured by the knee joint center, respectively. Specifically, the change in shank angle (excursion) and change in the horizontal and vertical translation of the shank's proximal end (displacements) were quantified during each rocker and over the entire stance phase. Excursions and displacements were calculated as the difference between the end position and start position, both for each individual rocker and the entire stance phase. Each parameter was calculated for at least two trials per subject (the range was 2 to 5), averaged across trials and then combined across subjects. To normalize across trials and subjects, the stance phase was divided into 101 datapoints and horizontal position of the ankle joint center was set to zero at heel strike for each trial.

#### 2.2.4 Kinematic model

To analyze shank segmental kinematics in gait, the sagittal plane gait model described by Pollen (Figure 7; Equations 1-5) that related proximal and distal shank position to distal foot position, foot and ankle angles as well as foot and shank geometry was used (Pollen, 2015).



Figure 7: Sagittal plane kinematic model of shank progression used to predict ankle and knee joint center trajectories during the stance phase of gait.

$$y_{ankle} = y_{MH2} - sin(\theta) * L_{foot}$$
(1)

$$z_{ankle} = z_{MH2} + \cos(\theta) * L_{foot}$$
(2)

$$y_{knee} = y_{ankle} - sin(\theta_0 - \phi) * L_{shank}$$
(3)

$$z_{knee} = z_{ankle} + \cos(\theta_0 - \phi) * L_{shank}$$
(4)

$$\theta_{shank} = \theta_0 - \phi \tag{5}$$

where  $L_{foot}$  and  $L_{shank}$  were the foot and shank lengths, respectively;  $y_{MH2}$  and  $z_{MH2}$ were the coordinates of the second metatarsal head;  $\theta$  was the foot-to-floor angle, defined by the rotation of the foot coordinate system relative to the laboratory coordinate system;  $\theta_0$  was foot-to-floor angle normalized to 0° in the static pose (quiet standing, feet flat on floor);  $\phi$  was the ankle (shank-to-foot) angle, defined by the angle between the long axis of the shank (ankle to knee joint center) and the long axis of the foot (second metatarsal head to ankle joint center); and  $\theta_{shank}$  was the sagittal plane shank angle, measured from the lab's vertical (z) axis. The model was previously developed and validated for typical individuals (Pollen, 2015; Pollen, Arch, & Stanhope, 2015).

#### 2.2.5 Shank kinematics and model validation for prosthetic gait

Shank kinematics were quantified by using both experimental measurements and model calculations. To validate the model for prosthetic gait, model-predicted shank kinematics (shank angle, vertical and horizontal positions of the shank's proximal end) were compared to means from the experimental gait analysis data over the whole stance phase. Then both the experimental and model based data were used to plot the sagittal plane shank trajectory throughout stance.

Next, the contributions of the prosthetic ankle and foot angles and ankle translation to shank progression in prosthetic gait were considered. Foot and ankle rotations were quantified via the foot-to-floor angle and ankle angle, respectively. Ankle translations were quantified via the vertical and horizontal translations of the foot's proximal end, as defined by the ankle joint center. Specifically, the change in ankle and foot-to-floor angles (excursions) and change in vertical and horizontal translations of the foot's proximal end (displacements) were quantified during each rocker as well as over the entire stance phase. Additionally, the percent contributions of the ankle and foot-to-floor angles to shank rotations were calculated by dividing the ankle and foot excursions by the shank excursion for each rocker. Ankle excursions

were negated for the percent contribution calculations as, given the established coordinate systems, negative excursions signified rotations of the foot and shank in the direction of forward progression, but ankle dorsiflexion, which would also rotate the system in the direction of forward progression, was positive. The percent contribution of the ankle translations to shank translations were also calculated in a similar fashion for each of the rockers.

#### 2.2.6 Statistical analysis

Due to the low number of prosthetic users included in the study, no statistical analysis was performed on the data. Instead, in order to assess the model validity for the prosthetic gait, visual and quantitative comparisons were made between the experimental data and the model results.

#### 2.3 Results

Experimental data and model results for shank angle excursion and proximal shank translational trajectories for one representative subject are depicted in Figure 8. Visual inspection of the prosthetic gait data showed close agreement between the experimental and model-predicted data for prosthetic users. The maximum mean difference between the model based shank angle and experimental data was  $0.32^{\circ} \pm 0.40^{\circ}$ . As for the proximal shank trajectories, the maximum mean difference between the model and experimental data was  $0.015 \pm 0.003$  m in the horizontal direction and  $0.0094 \pm 0.005$  m in the vertical direction.



Figure 8: Shank kinematics for one representative subject, the graphs shows the close agreement between experimental and model based data for a) shank angle, b) horizontal trajectory and c) vertical trajectory of the proximal end of the shank.

Shank kinematics in prosthetic gait for one representative subject are depicted in Figure 9. The figure shows the results from both experimental kinematic data as well as the calculated results from the model. The colored lines show the orientation of the shank throughout stance during each of the four rockers. The dashed lines at the proximal end (top) and distal end (bottom) of the shank show the model calculated position of the knee and ankle, respectively.



Figure 9: Shank trajectories in prosthetic gait. Experimental and model-based sagittal plane shank trajectory in prosthetic gait, colored by rockers during stance for one representative subject.

Mean angular excursions for ankle, foot and shank during each rocker are summarized in Table 1 (mean  $\pm$  standard deviation). Percent contributions of the ankle and foot-to-floor excursions to shank excursion for prosthetic gait are summarized as well.

Table 1: Angular excursions for ankle, foot and shank at each rocker and total during stance (mean  $\pm$  standard deviation) as well as percent contributions of the ankle and foot-to-floor excursions to shank excursion for prosthetic gait. Negative excursions signify ankle plantar flexion and rotations of the foot and shank in the direction of forward progression. A negative percentage indicates the contribution opposed the resultant angle of the shank.

	Angle	Heel Rocker (°)	Ankle Rocker (°)	Forefoot Rocker (°)	Toe Rocker (°)	Total (°)
	Ankle	-1.57 ± 3.46	8.50 ± 3.54	5.41 ± 3.60	-9.04 ± 3.71	3.30 ± 7.15
	Foot-to- floor	$\begin{array}{r} -23.83 \pm \\ 1.98 \end{array}$	-2.32 ± 2.55	-6.57 ± 3.36	-34.47 ± 5.54	-67.18 ± 7.23
	Shank	$\begin{array}{c} -22.00 \pm \\ 4.08 \end{array}$	-10.99 ± 3.89	-12.11 ± 3.74	$\begin{array}{r} -25.50\pm\\ 4.68\end{array}$	-70.60 ± 8.23
Percent	Ankle	-7%	77%	44%	-36%	5%
Contributions	Foot-to- floor	108%	21%	54%	135%	84%

The ankle angle showed a very slight plantar flexion during the heel rocker, dorsiflexion in the ankle and forefoot rockers and during the toe rocker there was 9.04  $\pm$  3.71° of plantar flexion (Table 1). The shank had a total excursion of 70.6  $\pm$  8.23° (Table 1) with most of the rotation taking place during the heel and toe rockers. Considering the interplay between the ankle and foot-to-floor angles and their contributions to the shank angle in prosthetic gait showed that they counteracted each
other during the heel and toe rockers. Over stance the foot-to-floor angle had a much greater effect on the shank angle than the ankle angle.

Table 2 summarizes the mean horizontal and vertical translational displacements of the proximal end of the foot and proximal end of the shank at each rocker as well as the total displacement during stance (mean  $\pm$  standard deviation). The table also lists the relative contributions of the proximal end of the foot's displacements to the proximal end of the shank's displacements.

Table 2: Horizontal and vertical translational displacements of the proximal end of the foot and proximal end of the shank at each rocker and total during stance (mean  $\pm$  standard deviation) as well as relative contributions of the proximal end of the foot's displacements (horizontal, vertical) to the proximal end of the shank's displacements for prosthetic and typical gait. Positive horizontal displacements signify anterior translation, positive vertical displacements signify a rise in the foot/shank's position. A negative percentage indicates the foot's displacement was in the opposite direction of the shank's displacement.

	Displacement Direction	Heel Rocker (m)	Ankle Rocker (m)	Forefoot Rocker (m)	Toe Rocker (m)	Total (m)
Proximal Foot (Ankle)	Horizontal	0.055	0.004	0.013	0.106	0.178
	Horizontai	$\pm 0.013$	$\pm 0.019$	$\pm 0.022$	$\pm 0.025$	±0.040
	Vertical	-0.016	0.004	0.011	0.078	0.077
		$\pm 0.022$	$\pm 0.019$	$\pm 0.019$	$\pm 0.015$	$\pm 0.038$
Proximal Shank	Horizontal	0.217	0.094	0.106	0.252	0.669
		$\pm 0.042$	$\pm 0.046$	$\pm 0.052$	$\pm 0.073$	$\pm 0.109$
	Vertical	0.006	0.002	-0.012	-0.040	-0.043
		$\pm 0.041$	$\pm 0.044$	$\pm 0.044$	$\pm 0.033$	$\pm 0.081$
Percent Contribution (Ankle to Shank)	Horizontal	25%	5%	12%	42%	
	Vertical	-287%	164%	-99%	-198%	-

The proximal end of the shank had anterior translation of  $0.669 \pm 0.109$  m, the motion again mostly happening during the heel and toe rocker (Table 2). The shank rose slightly during the heel and ankle rockers but lowered again during the forefoot and toe rocker for a total of  $-0.043 \pm 0.081$  m with most of the lowering happening during the toe rocker or  $-0.040 \pm 0.033$  m (Table 2). In the horizontal translations, the ankle translations positively contributed to the proximal shank translations throughout stance. In contrast, with the exception of the ankle rocker, the ankle and proximal shank counteracted each other in the vertical direction. This counteraction was most pronounced during the heel rocker, where it was 287% during this portion of stance, and only 198% during the toe rocker phase (Table 2).

## 2.4 Discussion

This analysis aimed to gain better insight into the kinematics of the shank segment in prosthetic gait. The kinematics of the proximal end of the shank were characterized and compared to the results from typical individuals and model predictions of prosthetic gait. The primary finding of this study was that the proximal shank was shown to maintain a relatively horizontal trajectory throughout stance in prosthetic gait, similar to typical gait and in contrast to the previously reported prediction (Pollen, 2015). This is, to the author's knowledge, the first analysis of this kind to be reported.

The previous study that investigated shank kinematics in typical gait revealed an interesting interplay in the natural ankle-foot system that, through an interaction of ankle and foot-to-floor angles as well as translations of segmental ends, the proximal end of the shank segment maintained a horizontal trajectory throughout most of stance

(Figure 5a) (Pollen, 2015). However, when using the kinematic model to predict the behavior of the proximal end of the shank while limiting the plantar flexion of the ankle, as is the case of passive prosthetic foot systems, the shank over-rotated and the proximal end dipped down at the end of stance (Figure 5b) (Pollen, 2015). Based on this model prediction, it was hypothesized that the shank would over-rotate and translate downward in prosthetic gait. Comparing the results from the four prosthetic users in this study to those predictions showed that, in fact, the shank did not rotate and lower vertically to the extreme that the model predicted (Figure 9). Instead, experimental data from the prosthetic users showed shank kinematics in prosthetic gait much more closely resembled the unaltered typical gait data. This discrepancy with the model prediction might be explained by spatiotemporal asymmetries commonly seen in prosthetic gait, namely stance times on the amputated side having been reported to be 5% shorter compared to the intact side (Adamczyk & Kuo, 2015; Isakov et al., 2000; Nolan et al., 2003). If this explanation holds, it can thus be theorized that in order to avoid the potentially detrimental consequences of the shank lowering and over-rotating and the resultant lowering of the body's COM, users of passive prosthetic devices may terminate the stance phase early, as it would take a considerable amount of energy to hoist the COM back up for the next gait cycle. The four prosthetic users included in this study did indeed show patterns of spatiotemporal asymmetries. Step length showed the greatest asymmetry, with a 6% difference between the amputated and sound limbs. This theory should be further investigated in future studies.

The kinematic model that was used in this study had been previously verified for typical gait, showing high agreement between the experimental and model

predicted data (Pollen, 2015). The current analysis indicated that the same is true for prosthetic gait, as the difference between the experimental and model based values of the shank angle as well as the translations of the segmental ends were minimal (Figure 8).

The main limitation of this study was the low number of subjects which prevented statistical analyses of these data. However, the results did show fairly strong trends and did indeed give a good idea of the prosthetic function. During the data collection, as the anatomical landmarks typically used for the placement of the ankle markers were missing on the prosthetic side, the marker placement had to be approximated on the prosthetic side, based on the intact side. However that might have caused some inaccuracy in the experimental data. Another limitation is that only one type of prostheses was used, as well as only one, self-selected gait speed. Analyzing the shank kinematics over a wider variety of prosthetic feet and walking speeds as well as having the speed standardized between subjects may yield different results. Additionally, all of the subjects were high functioning and had the same level of amputation, unilateral transtibial. Including prosthetic users over a wider range of functional and amputation levels can give better insight into the shank kinematics of prosthetic gait.

The results from this analysis show that prosthetic users appear to maintain similar shank kinematics (horizontal trajectory of the proximal end of the shank) as individuals without amputation. However, prosthetic users may adapt to the prosthesis' lack of plantar flexion by terminating the stance phase early to prevent excessive shank rotation and the lowering of the shank's proximal end. Enabling the prosthetic user to maintain the horizontal trajectory of the knee longer in stance might

lead to a more symmetric gait. Designers of future prosthetic systems should aim to better replicate the typical shank kinematics through further development of biomimetic systems or through novel non-biomimetic approaches. This can help prosthetic users achieve higher levels of function.

# Chapter 3

#### **AIM 2: SHANK SEGMENTAL POWER**

#### 3.1 Background

While there is much knowledge to be gained about how the body moves from kinematic analysis, it does not offer complete insight into the muscle work that takes place to achieve the movement. Kinetic analysis of the forces, moments and resulting powers can yield better understanding of how the muscles work. Power is the amount of work performed per unit of time. It can also be defined as the amount of energy created, absorbed or transferred between systems at each instant in time (Winter, 1987). Segmental power, as it refers to body segments, is the rate at which energy is created or absorbed by muscles or flows between body segments (Winter, 1987, 2009).

There are two main ways to calculate segmental power. One method calculates segmental power through inverse dynamics analysis as the sum of the power flowing in (positive) or out (negative) of the segment's proximal and distal ends and the other calculates the segmental power as the sum of the total rate of change of energy in the segment (Figure 10) (Robertson & Winter, 1980).

$$P_{t,p} = \vec{F}_{p} \cdot \vec{v}_{p}$$

$$P_{r,p} = \vec{M}_{p} \cdot \vec{\omega}$$

$$P_{shank} = P_{r,p} + P_{t,p} + P_{r,d} + P_{t,d}$$

$$=$$

$$\frac{dE_{shank}}{dt} = PE_{shank} + TKE_{shank} + RKE_{shank}$$

$$P_{t,d} = \vec{F}_{d} \cdot \vec{v}_{d}$$

Figure 10: Segmental power; The figure shows how power is either calculated using the flow of power at the proximal and distal ends of the segment or with the rate of change of segmental energy. The flow of translational power,  $P_t$ , from joint forces (blue) and rotational power,  $P_r$ , muscle moments (red) flowing into the proximal and distal end of the shank are also depicted.

In the former method, the total power is calculated as the result of a passive flow of energy between segments through joint forces, referred to as translational power and active energy transfer through moments created by the pull of the muscles attached to the segment, or rotational power (Figure 10) (Robertson & Winter, 1980; Zajac, Neptune, & Kautz, 2002). Several assumptions have to be made for inverse dynamic analysis, such as the assumption of rigid body segments (Zajac et al., 2002). The rigid body assumptions cannot, however, apply to all body segments. The foot especially cannot be modeled as rigid as it does deform during stance for shock absorption and propulsion (Ren, Howard, et al., 2008). Therefore, a kinetic model that calculates the distal foot power has been developed (Takahashi & Stanhope, 2013). Another example of the violation of the rigid body assumption is a prosthetic foot, which, in the case of ESR feet for example, also deforms during stance (Versluys et al., 2008). Thus, a model to calculate the segmental power for the prosthetic segment and residual shank in individuals with transtibial amputation has also been developed (Takahashi, Kepple, & Stanhope, 2012). That method models the prosthetic segment and the residual shank as one continuous segment where a proximal rigid component (the shank) is connected in series with a deformable distal component (Takahashi et al., 2012).

The second method to calculate segmental power is done by using the rate of change, or time derivative, of the total energy of the segment, which is the sum of the segment's potential, translational kinetic and rotational kinetic energy and uses the motion of segments and anthropometric estimates of segment lengths and mass (Figure 10) (Robertson & Winter, 1980; Zajac et al., 2002). Those two methods of calculating power should yield the same results, as they are simply two different ways of calculating the same thing (Zajac et al., 2002). Nonetheless, traditionally there has been some discrepancy, or power imbalance, between these two methods (McGibbon & Krebs, 1998). A recent analysis, however, reported a way to remove the imbalance by introducing an error correction term which accounts for the relative motion that takes place between segment ends within a joint (Ebrahimi et al., 2017). That was analogous to defining the proximal end of the next segment below the segment in question as the distal end (Ebrahimi et al., 2017).

Power analysis can give better insight into many mechanisms in the body during gait, for instance the ankle function. Many important kinetic functions take place in the ankle during stance. There is a small phase of power absorption right after

heel-strike (Winter, 1987). During the majority of the stance phase, after the weight acceptance period, the plantar flexor muscles contract eccentrically in order to control the forward rotation of the shank, creating an increasing plantar flexion moment that results in a phase of power absorption (Takahashi & Stanhope, 2013; Winter, 1987). During late-stance, after heel-off, the increasing plantar flexion moment causes a plantar flexion of the ankle which results in a high positive power spike that enables the foot to push off from the ground and start the swing phase (Takahashi & Stanhope, 2013; Winter, 1987).

Push-off, often defined as the peak ankle power (Winter, 1983), is an important event in gait. The power produced by the ankle plantar flexors at push-off is the highest power seen in the body during typical gait (Huang et al., 2015; Neptune et al., 2001). Decreased push off power, whether unilaterally or bilaterally, can lead to increased metabolic cost during walking (Doets, Vergouw, Veeger, & Houdijk, 2009; Houdijk, Pollmann, Groenewold, Wiggerts, & Polomski, 2009), increased flexion moment in the hip on the affected side at early stance (Judge, Davis, & Ounpuu, 1996; Mueller, Minor, Sahrmann, Schaaf, & Strube, 1994; Prince, Corriveau, Hébert, & Winter, 1997; Winter & Sienko, 1988) and greater collision energy loss on the unaffected side (Adamczyk & Kuo, 2015; Huang et al., 2015). Individuals who lack the natural ankle-foot complex show decreased ankle push-off power in late stance (Ferris, Aldridge, Rábago, & Wilken, 2012; Winter & Sienko, 1988). Prosthetic users must make up these deficiencies through the function of the prosthesis or by compensating elsewhere (Silverman et al., 2008; Voinescu et al., 2012; Weinert-Aplin et al., 2017; Winter & Sienko, 1988). It has been reported that at push-off in typical gait as well as on the intact side in prosthetic gait, power flows from the shank towards

the thigh, i.e. the proximal shank power was negative as power is generated by the plantar flexors (Weinert-Aplin et al., 2017). In contrast, at push-off on the prosthetic side in prosthetic gait, power flowed into the shank from the thigh (Weinert-Aplin et al., 2017).

Recent prosthetic systems have been designed with the replication of anklefoot kinetics and ROS in mind (Cherelle et al., 2014; Ren, Jones, & Howard, 2008; Versluys et al., 2009), such as with ESR feet and powered prostheses that are designed to return the energy stored during stance (ESR) or generate the power spike seen in push-off (powered). Yet, as they have still not been perfected (Crimin et al., 2014; Hitt et al., 2009; Versluys et al., 2009), there is a reason to consider new design criteria, such as the kinetics of other body segments. A recent analysis that investigated the proximal shank kinetics at push-off in typical gait revealed that the segmental powers at the proximal shank were mainly due to the translational power component, i.e. the dot product of the joint force and the velocity of the segment's end (Pollen, 2015). The velocity was mainly in the anterior (y) direction, indicating that the majority of the power was directed into pushing forward (Pollen, 2015). However, while this previous work provided important insights into the shank's segmental power in typical gait, the segmental power profile of the residual shank for users of passive prostheses is not well understood, particularly during push-off.

Therefore, the aim of this study was to characterize the lower extremity segmental powers during gait for users of a unilateral, transtibial, energy-storing-andreturning prosthesis, with particular attention paid to the segmental kinetics of the proximal shank at push-off. It was hypothesized that there would be a lower power output from the prosthetic segment compared to the intact side. Additionally, it was

hypothesized that when considering the individual power components at push-off in prosthetic gait, the translational power term would be higher than the rotational power term, and that the joint force and velocity at the shank's proximal end would be less on the prosthetic side compared to typical, as passive prostheses lack active push-off. Results from this study may provide important insights that can be used to optimize the design of prostheses.

# 3.2 Methods

### **3.2.1** Data acquisition and analysis

The same data as was described in Section 2.2 were used for the kinetic analysis.

## **3.2.2** Kinetic models

Several models were used to calculate the segmental powers for prosthetic gait based on the nature of each segment.

On the prosthetic segment the segmental power was calculated using a unified deformable model (UD) described by Takahashi et al. (2012) (Figure 11a). In their approach, the UD segment was defined as being comprised of the prosthetic foot and socket as well as the residual shank and was modelled as a proximal rigid component connected in series to a deformable distal component (Takahashi et al., 2012). The UD segmental power was calculated using Equations 6 and 7

$$P_{UD} = \vec{F}_{grf} \cdot \vec{v}_{UD,d} + \vec{M}_{free} \cdot \vec{\omega}_{UD}$$
(6)

where  $\vec{F}_{grf}$  is the ground reaction force,  $\vec{M}_{free}$  is the free moment,  $\vec{\omega}_{UD}$  is the angular velocity of the UD segment. The  $\vec{v}_{UD,d}$  velocity term is calculated with Equation 7

$$\vec{v}_{UD,d} = \vec{v}_{UD,cm} + (\vec{\omega}_{UD} \times \vec{r}_{UD\_cop}) \tag{7}$$

where  $\vec{v}_{UD,cm}$  is the translational velocity of the UD segment's center of mass (COM) and  $\vec{r}_{UD\_cop}$  is the displacement of the center of pressure (COP) with respect to the COM of the UD segment (Takahashi et al., 2012).

For the foot on the intact side, a deformable segment model was used to account for the deformation of the distal foot (Equations 8 and 9) (Figure 11b)

$$P_{ft,d} = \vec{F}_{grf} \cdot \vec{v}_{ft,d} + \vec{M}_{free} \cdot \vec{\omega}_{ft}$$
(8)

where, again,  $\vec{F}_{grf}$  is the ground reaction force,  $\vec{M}_{free}$  is the free moment, and  $\vec{\omega}_{UD}$  is the angular velocity of the foot segment. The  $\vec{v}_{ft,d}$  term is calculated with Equation 9

$$\vec{v}_{ft,d} = \vec{v}_{ft,cm} + (\vec{\omega}_{ft} \times \vec{r}_{ft\_cop}) \tag{9}$$

where  $\vec{v}_{ft,cm}$  is the translational velocity of the foot's center of mass (COM) and  $\vec{r}_{ft\_cop}$  is the distance between the center of pressure (COP) and the COM of the foot (Siegel, Kepple, & Caldwell, 1996; Takahashi & Stanhope, 2013).



Figure 11: a) Unified deformable (UD) model for calculating segmental power in the prosthesis and residual shank, and b) distal foot power model for calculating segmental power in the distal foot on the intact side.

To calculate segmental powers on the remaining segments (thigh on prosthetic side, shank and thigh on sound side, and pelvis), a six-degree-of-freedom (6DOF) kinetic model was used, as described by Ebrahimi et al. (2017). In the 6DOF model a "power correction" term is used to account for the relative movement between segments. By using that kinetic method, it is possible to achieve the same results as when power is calculated using the rate of change of energy (Ebrahimi et al., 2017).

The segmental power for segment m is calculated using Equation 10 with the 6DOF method

$$P_{m} = \vec{M}_{p,m} \cdot \vec{\omega}_{m} + \vec{M}_{d,m} \cdot \vec{\omega}_{m} + \vec{F}_{p,m} \cdot \vec{v}_{p,m} + \vec{F}_{d,m} \cdot \vec{v}_{d-JC,m}$$

$$= (I_{m}\vec{\alpha}_{m} + \vec{\omega}_{m} \times I_{m}\vec{\omega}_{m}) \cdot \vec{\omega}_{m} + (m_{m}\vec{a}_{m} - m_{m}\vec{g}) \cdot \vec{v}_{m} - (\vec{r}_{cop,m} \times \vec{F}_{grf}) \cdot \vec{\omega}_{m} + (\vec{r}_{cop,m-1} \times \vec{F}_{grf}) \cdot \vec{\omega}_{m} - \vec{F}_{grf} \cdot (\vec{\omega}_{m} \times \vec{r}_{p,m}) + \vec{F}_{grf} \cdot (\vec{\omega}_{m} \times \vec{r}_{d-AR,m}) + \vec{F}_{grf} \cdot (\vec{\omega}_{m} \times \vec{r}_{m/m-1}) \quad (10)$$

where  $\vec{M}_{p,m}$  and  $\vec{M}_{d,m}$  are the proximal and distal joint moments for the segment, respectively;  $\vec{F}_{p,m}$  and  $\vec{F}_{d,m}$  are the proximal and distal joint forces for the segment, respectively;  $\vec{v}_{p,m}$  is the translational velocity of the proximal end of the segment and  $\vec{v}_{d-JC,m}$  is the translational velocity of the distal end of the segment, defined as the proximal end of the more distal segment;  $I_m$  is the segment's moment of inertia;  $\vec{\alpha}_m$ and  $\vec{\omega}_m$  are the angular acceleration and angular velocity, respectively;  $m_m$  is the segment's mass;  $\vec{a}_m$  and  $\vec{v}_m$  are the acceleration and velocity of the segment's COM, respectively;  $\vec{r}_{cop,m}$  is the vector from the segment's proximal end to the COP;  $\vec{F}_{grf}$ was the ground reaction force;  $\vec{r}_{p,m}$  is the vector from the segment's COM to its proximal end,  $\vec{r}_{d-AR,m}$  is the vector from the COM to its distal end and  $\vec{r}_{m/m-1}$  the vector from the segment's distal end to the proximal end of the next segment below (Ebrahimi et al., 2017)

## **3.2.3** Outcome variables

Key outcome variables analyzed in this study were the segmental power for the pelvis segment, the foot, shank and thigh on the intact side and the thigh and the unified deformable segment (residual shank and prosthesis) on the prosthetic side.

Additionally, to assess proximal shank kinetics at "push-off" in prosthetic gait and evaluate the hypothesis, the power at the proximal shank was calculated at the instance of peak ankle power (push-off) on the prosthetic and intact side. The individual components of the segmental power, the rotational and translational power terms, as well as the components that make up those, the the proximal moment and angular velocity of the shank and the joint force and velocity at the proximal end of the shank were reported as well. As the motion in gait mainly takes place in the sagittal plane (Srinivasan, Raptis, & Westervelt, 2008) only the anteroposterior (y) and vertical (z) components were analyzed.

## **3.2.4** Statistical analysis

Due to the low number of prosthetic user subjects (n=4), statistical analysis was not performed.

# 3.3 Results

## **3.3.1** Segmental power

During the first 20% of stance, the segmental power was negative for all segments in all subjects with the exception of the pelvis segment during stance on the prosthetic side in one subject (Figure 12, Appendix C). During midstance (roughly 20-80%) the power mostly stayed around 0 W/kg. At terminal stance, 80-100% of stance, the pelvic power went back to negative while the rest of the segments showed relatively large positive powers (Figure 12, Appendix C).



Figure 12: Segmental power in stance for one representative subject, when the signal is positive power is flowing into the segment; Note that for the pelvis segment, intact and prosthetic side refer to the stance side.

# 3.3.2 Shank proximal power at push-off

Table 3 shows the power at the proximal end of the shank segment at the moment of push-off, defined as peak ankle power for the prosthetic users.

	Total power, P=Pt+Pr (W/kg)		Translatio Pt=Fp·v	onal power, <sub>p</sub> (W/kg)	Rotational power, Pr=M <sub>P</sub> ·ω (W/kg)		
Subject	Intact side	Prosthetic side	Intact side	Prosthetic side	Intact side	Prosthetic side	
1	-0.869	0.552	-1.257	1.093	0.388	-0.541	
2	-0.346	0.680	-0.370	1.014	-0.030	-0.362	
3	-0.499	0.065	-0.053	0.330	-0.447	-0.265	
4	0.865	0.564	0.857	0.859	0.008	-0.294	
Average	-0.212	0.465	-0.206	0.824	-0.020	-0.366	
SD	0.751	0.273	0.872	0.343	0.341	0.124	

Table 3: Proximal shank power (W/kg), total power and individual power components at the moment of push-off.

The results showed that for three of the four subjects, power flowed out from the shank at push-off on the intact side. Only Subject 4 was different. Furthermore, for all subjects, the power entered the shank at push-off on the prosthetic side (Table 3). Looking at the individual components of the power, the joint force power,  $P_t$ , was the dominating component for three subjects (Subjects 1, 2 and 4) (Table 3).

As seen in Table 4, inspection of the components that make up the segmental joint force power,  $P_t$  of the shank's proximal end revealed that the vertical (z) component of the joint force dominated on both the prosthetic and intact sides. In support of our hypothesis, the magnitude of the joint force was lower on the prosthetic side than the intact side for all subjects. Additionally, the horizontal (y) component of the velocity was much larger than the vertical (z) component for the proximal end of the shank (Table 4). In contrast to our hypothesis, however, the velocity at the shank's proximal end was, in most cases, higher on the prosthetic side than the intact side (Table 4).

		Proximal shank force, F <sub>P</sub> (N/kg)			Proximal shank velocity, vp (m/s)		
	Subject	Y	Ζ	Y/Z	Y	Ζ	Y/Z
Intact side	1	-1.256	-5.774	22%	1.706	-0.154	-1104%
	2	-0.656	-4.027	16%	2.063	-0.235	-879%
	3	-0.605	-4.298	14%	1.884	-0.251	-751%
	4	0.401	-0.312	-128%	1.971	-0.237	-831%
	MEAN	-0.529	-3.603	-19%	1.906	-0.219	-891%
	STDEV	0.687	2.324		0.152	0.044	
Prosthetic side	1	-0.223	-3.366	7%	2.247	-0.473	-475%
	2	-0.308	-3.441	9%	2.631	-0.529	-498%
	3	-0.178	-2.957	6%	1.803	-0.173	-1044%
	4	0.084	-2.444	-3%	1.897	-0.279	-679%
	MEAN	-0.156	-3.052	5%	2.145	-0.364	-674%
	STDEV	0.169	0.457		0.376	0.166	

Table 4: Joint force (N/kg) and velocity (m/s) at the proximal shank at the instant of push-off for the prosthetic users, the table shows the anteroposterior horizontal component (Y), vertical component (Z) and the ratio of the two (Y/Z).

To clarify this further, a post-hoc analysis on the push-off power at the proximal shank in the gait of ten typical individuals was conducted. The data were drawn from the same database as the data that were used to characterize the typical shank kinematics described in Chapter 2 (Pollen, 2015). These ten individuals had the mean age of  $24.2 \pm 3.0$  years and had an average height  $1.71 \pm 0.07$  m, body mass  $69.6 \pm 11.3$  kg and none of them had any history of musculoskeletal injury or disease. During the data collection, the typical individuals walked over a strain gauge force platform at a scaled velocity of 0.8 body heights per second (BH/s). The results from the post-hoc analysis revealed an average proximal shank power of  $-1.073 \pm 0.421$  W/kg at the moment of push-off. Investigation into the individual components of the power revealed that the majority of the power came from the translational power term. The joint force at push-off at the proximal shank was on average  $-1.076 \pm 0.174$  N/kg and  $-5.506 \pm 0.412$  N/kg in the horizontal (y) and vertical

(z) directions, respectively. The velocity of the proximal shank was  $1.952 \pm 0.121$  m/s in the y-direction and  $-0.198 \pm 0.066$  m/s in the z-direction at push-off. Those values are lower than the ones seen on the prosthetic side in the prosthetic gait and more closely matched the intact side. Note that when comparing prosthetic gait to typical, as the walking speeds were not standardized between the groups, only the general trends and signs (positive or negative) of the values were inspected and direct comparison of the values was not done.

## 3.4 Discussion

In this study, the segmental power in prosthetic gait was calculated. The mechanics of push-off at the end of stance were also analyzed by considering the direction and magnitude of the proximal shank's segmental power at the instance of peak ankle power.

The shape of the segmental power curves for the prosthetic users mostly resembled those of typical gait (Ebrahimi et al., 2017; Robertson & Winter, 1980), however the amplitude was generally lower (Figure 12, Appendix C). With the exception of one prosthetic user (Figure 12), the pelvis segmental power was negative in the first 15-20% of stance. The negative power at the beginning of stance for all segments indicated that power was leaving all lower extremity segments, possibly to redirect the body's center of mass upwards. The power for most of the segments stayed close to 0 W/kg between roughly 20-80% of stance (Figure 12, Appendix C), indicating small net flow of energy between segments during midstance. The pelvic power was negative at the end of stance for all subjects except in one case (Figure 12), indicating that power was flowing out of the segment, possibly towards lower

segments that all showed a positive power spike at terminal stance (Figure 12, Appendix C).

The results from the push-off analysis indicated that at the instance of push-off, i.e. peak ankle power, the power flowed out of the proximal shank on the intact side but into the shank on the prosthetic side (Table 3). This is in accordance with previous studies (Weinert-Aplin et al., 2017). Additionally, it was revealed that the majority of the proximal shank power came from the joint force and velocity (Table 3) which was similar to what has been reported in typical gait (Pollen, 2015). Studies indicate that on the intact side, the outflow of power is due to the high power generation of the plantar flexors (Neptune et al., 2001; Weinert-Aplin et al., 2017; Winter, 1987). In contrast, on the prosthetic side, power must be generated in other segments and then directed into the shank segment to compensate for the lack of active push-off (Weinert-Aplin et al., 2017; Winter & Sienko, 1988). Thus, the prosthetic users may be lifting the prosthetic limb off the ground, instead of pushing it off as is done in typical gait.

Considering the individual components of the proximal shank translational power, the force was generally lower on the prosthetic side (Table 4), which was in accordance with previous literature (Silverman et al., 2008). The results for the velocity of the proximal shank were less expected. The proximal shank velocity was mainly directed in the anterior direction on both the intact and prosthetic side (Table 4), similar to what was reported in typical individuals (Pollen, 2015). The ratio between the horizontal (anteroposterior) and vertical components was larger on the intact side than on the prosthetic side (Table 4). What was unexpected was that the velocity, both in the horizontal and vertical direction was generally higher on the

prosthetic side than on the intact side (Table 4) or in typical gait. This is not what was expected to be seen, as it was hypothesized that with lower push-off power on the prosthetic side, the velocity of the proximal shank would be lower than on the intact side or in typical gait. This discrepancy might be explained by the nature of gait, which has been theorized by some to be a series of controlled falls, slowed down by the action of the plantar flexors during the second half of stance (Perry, 1992). As prosthetic users lack plantar flexors, the fall might be less controlled, making the proximal shank reach higher velocities before the foot is lifted off the ground and into the swing phase. This might be connected to the asymmetrical spatiotemporal parameters seen in prosthetic gait (Adamczyk & Kuo, 2015), much like was observed in the proximal shank kinematics discussed in Chapter 2 of this analysis. That is, in addition to ending stance phase early in order to avoid further drop of the proximal shank, the prosthetic user also avoids further downward acceleration of the proximal shank. This theory should be investigated further. Note that the difference in walking speed exhibited by the two groups (the prosthetic users walked at 0.6 body heights/s while the typical individuals walked at 0.8 body heights/s) may have accounted for some of the difference in the results displayed by them. However, as there is also asymmetry between the intact and prosthetic sides in the prosthetic users, this difference in walking speed does not explain the whole difference.

There were some limitations in this study. First, this analysis was only conducted on four subjects, which made any meaningful statistical analysis of the data impossible. Since the results from the segmental power analysis showed some variability between subjects, including more subjects might give a better idea of the segmental power. As the prosthetic users lack the anatomical landmarks necessary for

the ankle marker placement on the prosthetic side, the ankle marker placement on the had to be approximated based on the sound side. This might have lead to slight inaccuracy in the experimental data. The fact that only one type of prosthetic foot was used is another limitation. While it reduces variability somewhat it does not give a general idea of the kinetic behaviour for other prostheses. Additionally, the subjects were quite uniform in regard to functional level and level of amputation. Another limitation is that data were only collected for self-selected walking speed which was not standardized between users. Using more than one type of prostheses, more walking conditions and including prosthetic users over a wider range of function levels might give better insight into the segmental kinetics.

Studies of joint and segmental kinetics in gait, similar to the one reported here have been published. However, to the author's knowledge, none have analyzed the individual components of the segmental power in prosthetic gait as was done here, particularly the velocity of the proximal shank at push-off. This kinetic analysis yielded insight into the movement strategies that take place in the body to compensate for the limited kinetic function associated with passive prosthetic feet. Prosthetic users showed altered gait kinetics as a result of limited push-off power in the prostheses, particularly with power flowing into the shank at push-off on the prosthetic side, as opposed to the intact side where power was flowing out. The implications of these results give a better idea of the shortcomings that designers of prosthetic systems have yet to overcome in order to ensure the users' full function. Knowledge of the kinetics of prosthetic gait can give designers of prosthetic feet an idea of the qualities new prosthetic feet should possess in order to overcome the shortcomings of current prosthetic feet should possess in order to overcome the shortcomings of prosthetic proximal

shank kinetics at push-off might have beneficial effects on prosthetic gait and allow users to achieve more symmetric gait patterns.

### **Chapter 4**

#### CONCLUSION

In this analysis, the kinematics and kinetics of the proximal shank in prosthetic gait were determined. The results have yielded better insight into the current performance of commonly used prosthetic feet.

The main conclusion is related to the compensatory mechanism that took place in prosthetic gait. The prosthetic users displayed asymmetric spatiotemporal parameters, such as a shortened stance phase and an increased step length on the prosthetic side. From the results reported here it can be theorized that this is done to avoid the lowering of the proximal shank and further downward acceleration that otherwise would take place towards the end of the stance phase. These theories need to be investigated further. However, the results from the proximal shank analysis reported here did give interesting new insight into the possible reasons for the compensatory mechanisms and asymmetry displayed by prosthetic users that have previously gone unnoticed due to the focus mainly being on the foot and ankle properties. The clinical implications of these results are the possibility of designing new prosthetic systems that aim to match the proximal shank mechanics in the hope of reducing asymmetry in prosthetic gait. This might even be achieved through nonbiomimetic methods, for instance, including a shank segment that lengthens at the second half of stance as opposed to an ankle that plantar flexes and thus matches the proximal shank kinematics of typical gait. Maintaining the horizontal trajectory of the proximal shank in this way might reduce the spatiotemporal asymmetry in prosthetic gait.

The biggest limitation of this study was the small number of subjects included, which prevented any statistical analysis. However, there were some apparent trends that gave a good idea of the properties that were investigated. This can be improved upon by including more subjects in future studies. During the data collection, accuracy may have diminished somewhat because the placement of the calibration markers on the lateral and medial malleoli had to be approximated on the prosthetic side based on the intact side, as the prosthetic users lack the malleoli. Using precise measurements of the position of the markers on the intact side to place the markers on the prosthetic side did help with a more accurate placement. Furthermore, the gait speed used for the data collection was not standardized between subjects as they were instructed to walk at their self-selected walking speed. However, as each subject walked at their habitual walking speed, the gait patterns displayed were not altered by the prosthetic users trying to walk unusually fast or slow. The subjects were all high functioning individuals with unilateral transtibial amputation. Although this contributed to the uniformity of the subjects and thus reduced the variability in the results somewhat, this also prevented generalization of the results to other groups of prosthetic users. Including users over a wider range of function and with different levels of amputations, e.g. individuals with transfermoral or bilateral amputations can give more generalizable results. Additionally, all subjects used the same type of prosthesis, i.e. the Vari-Flex® ESR prosthesis (Össur, Reykjavik, Iceland). Analyzing different types of ESR or even powered prostheses might yield more comprehensive results and expand on the theories put forth in this study.

The purpose of this study was to characterize the shank kinematics and kinetics in the gait of individuals with unilateral transtibial amputation using an ESR

prosthesis. In the first aim, the kinematics of shank progression were characterized experimentally and using a kinematic model. It was hypothesized that the proximal shank would lower towards the end of stance as had been predicted in a previous analysis. However, this hypothesis was not supported, as the proximal shank mostly maintained a horizontal trajectory throughout the stance phase in the prosthetic gait. In the second aim, shank segmental power was calculated. It was hypothesized that there would be a lower power output from the prosthesis compared to the intact ankle-foot complex. Results showed that at push-off, power was flowing into the proximal shank on the prosthetic side but out of the proximal shank on the intact side. This indicated that there is a lower power output form the more distal prosthetic segment that must be compensated for in more proximal structures, which supports the hypothesis.

Shank segmental kinematics and kinetics in prosthetic gait have, to the author's knowledge, not been investigated and reported on in this way before. These results had interesting implications for the limitations of current prosthetic systems and revealed possible explanations for the asymmetric nature of prosthetic gait that have gone unexamined until now with the focus having primarily been on the ankle-foot mechanics. As prosthetic users typically have lower levels of function than individuals without amputation it is important to improve the design of prostheses to enable users to reach their highest levels of function. There are good implications for future research of the theories suggested from these results, such as to confirm that the asymmetric spatiotemporal parameters are truly due to the avoidance of the lowering and downward acceleration of the proximal shank. There is a true possibility of design reform for new prosthetic systems which take the kinematics and kinetics of the

proximal shank into account that has opened the door for the ultimate improved function and quality of life of prosthetic users.

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# Appendix A

# PERMISSION FOR USING GRAPHS FROM PREVIOUS WORK

# TRAVIS POLLEN

1531 N. Lawrence Street, Philadelphia, PA 19122 | (215) 439-4025 | trp59@drexel.edu

April 4, 2018

# To Whom It May Concern:

I, Travis Pollen, hereby give Rosa Kolbeinsdottir permission to use graphs published in my thesis.

Sincerely,

fron tol

Travis Pollen

# Appendix B

INFORMED CONSENT DOCUMENT

IRB Approved From: 04/07/2016 to: 04/12/2017

## INFORMED CONSENT TO PARTICIPATE IN RESEARCH

Title of Project: Human Movement Analysis Database

Principal Investigator(s): Stanhope, Steven J.

You are being invited to participate in a research study. This consent form tells you about the study including its purpose, what you will be asked to do if you decide to take part, and the risks and benefits of being in the study. Please read the information below and ask us any questions you may have before you decide whether or not you want to participate.

#### WHAT IS THE PURPOSE OF THIS STUDY?

The purpose of this study is to collect information on the different ways people use rehabilitation devices, such as artificial legs or ankle braces to move when they walk, jog, or run. Scientists and doctors often compare information obtained from diverse groups of people to patient information in order to better understand the effects of disease and treatment on patient problems. You will be one of approximately 300 participants in this study.

#### WHY ARE YOU BEING ASKED TO PARTICIPATE?

You are being asked to participate in this study because we expect that you use normal patterns to move and we wish to see how your pattern of moving changes when you wear different types of artificial legs or braces.

Subjects with an unsafe, unsteady, or highly variable movement pattern upon visual observation will be excluded. Subjects who are unable to repeatedly execute the movement pattern in the desired manner will be excluded from participation

## WHAT WILL YOU BE ASKED TO DO?

Before participating in this study, all of the movement tasks you will be asked to carry out will be explained by Dr. Stanhope or another member of the research team. The type of movement task (walk, jog, or run) you will be asked to perform will be determined in advance by the research team based on your prosthetic/orthotic prescription. You may wish to not perform a specific task and not to take part in the study. If you wish to continue, your participation in this study will involve one or potentially more visits to the University of Delaware for a maximum of 2 hour per visit. A visual walking test will be performed by a member of the research team to determine how your joints move, how strong you are, and your comfortable walking speed. These procedures should not cause any discomfort.

Prior to your instrumented movement test, your ankle muscle strength may be measured while you are seated in a device, called a Kincom, which controls your ankle motion and measures how much force you can produce. The test will measure the strength of your ankle muscles during repetitive trials in which you will be provided with approximately 1-minute rest between each trial. Strength will be measured during a

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static test. Additionally, you may be asked to perform a heel rise test where you will stand on one leg with your foot on an incline. You will be allowed a light touch on the wall in front of you to help you maintain your balance while performing the test. A thread will be attached to your heel. You will be asked to perform as many and as high heel-rises as you can until fatigued. You will hear a clicking sound that will guide you to the appropriate frequency of the heel-rises. If you wear a prosthesis, you will not be asked to perform either of these tests on your prosthetic side.

During your instrumented movement test, you will be requested to wear a t-shirt and shorts. You may be asked to walk, jog, or run overground and/or on a treadmill. When walking on the treadmill, a body weight support system may be used during data collection. The support system is designed to safely provide constant body weight support up to 100% body weight, for individuals up to 150kg. This system includes an overhead harness system that will be used as subjects walk over a split-belt treadmill. It is designed to catch subjects in the event of a fall and the harness can be quickly removed. Subjects will be fitted with an overhead harness.

Small plastic reflective balls will be attached to your body. To do this, your arms and legs will be wrapped with a soft, rubber-like material. A piece of firm material called a shell may then be attached to the rubber sleeves with Velcro or a self-adherent bandage. The small round balls may also be attached to your skin using an adhesive. After the reflective balls have been attached, the harness will then be secured to the body weight support system. Additionally, we may also want to test your muscles using electromyography (EMG). To do this, we will attach small metal electrodes to the surface of your skin using an adhesive. EMG is a measurement tool that is used to assess muscle function. Lastly, we may also ask you to breathe through an oxygen valve during the movement task to obtain a measurement of your metabolic energy expenditure. You should not feel any discomfort with these tests.

Once the above items are in place, you will be asked to perform a task several times while scientific cameras record the positions of the reflective balls. The cameras do not take pictures of your face or body parts. Each instrumented movement test will require a maximum of 2 hours to complete. You may rest at any time. Following the instrumented movement test, we will ask you to complete a questionnaire evaluating the performance of any artificial leg or brace you may wear.

If you are wearing an artificial leg or an ankle brace, you may be asked to repeat the protocol multiple times (in the same visit or different visits) with different types or settings of artificial legs or braces. However, you may decline our request and ask to stop participating at any time. If the protocol is repeated within the same visit, you will be given ample time to get acclimated to moving with the different artificial leg or brace (a minimum of 5 minutes), until you feel stable, comfortable, and until you feel that your movement pattern is reproducible.

# WHAT ARE THE POSSIBLE RISKS AND DISCOMFORTS?

Possible risks of participating in this research study are minimal; no more than those incurred during normal walking, jogging, or running and customary training and supervised use of a rehabilitation device. There is

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a slight chance of a mild skin irritation from the attachment of adhesive circles to the skin during the gait analysis portion of the study. The soft, rubber-like material may feel tight, but if it is uncomfortable or interferes with your movements, tell one of the investigators and it will be readjusted. This material may cause a skin irritation, but the material is worn only for a short period of time and skin reactions are rare. There is also a slight chance of skin irritation due to wearing an artificial leg or brace or the harness of the body weight support system; however, adjustments will be made so that you will remain as comfortable as possible. Your safety will be continuously monitored while you are walking, jogging, or running with the artificial legs or braces.

## WHAT IF YOU ARE INJURED DURING YOUR PARTICIPATION IN THE STUDY?

If you are injured during research procedures, you will be offered first aid at no cost to you. If you need additional medical treatment, the cost of this treatment will be your responsibility or that of your third-party payer (for example, your health insurance). By signing this document, you are not waiving any rights that you may have if injury was the result of negligence of the university or its investigators.

#### WHAT ARE THE POTENTIAL BENEFITS?

You will not benefit directly from taking part in this research.

#### NEW INFORMATION THAT COULD AFFECT YOUR PARTICIPATION:

During the course of this study, we may learn new information that could be important to you. This may include information that could cause you to change your mind about participating in the study. We will notify you as soon as possible if any new information becomes available.

# HOW WILL CONFIDENTIALITY BE MAINTAINED? WHO MAY KNOW THAT YOU PARTICIPATED IN THIS RESEARCH?

Each subject will be assigned a unique numerical subject identifier that will be used to label and track all data. Documents containing patient identifiers and the keys for breaking subject identification codes will be kept separately in a secured location with access limited to the PI. When results of a University research study are reported in medical journals or at scientific meetings, the people who take part are not named and identified. In most cases, the University will not release any information about your research involvement without your written permission. However, if you sign a release of information form, for example, for an insurance company, the University will give the insurance company information from your instrumented movement analysis record. This information might affect (either favorably or unfavorably) the willingness of the insurance company to sell you insurance.

The confidentiality of your records will be protected to the extent permitted by law. Your research records may be viewed by the University of Delaware Institutional Review Board, which is a committee formally designated to approve, monitor, and review biomedical and behavioral research involving humans. Records relating to this research will be kept for at least three years after the research study has been completed.

#### WILL THERE BE ANY COSTS TO YOU FOR PARTICIPATING IN THIS RESEARCH?

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IRB Approved From: 04/07/2016 to: 04/12/2017

There are no costs for participating in this study.

### WILL YOU RECEIVE ANY COMPENSATION FOR PARTICIPATION?

You will not receive compensation for participating in this study.

## DO YOU HAVE TO TAKE PART IN THIS STUDY?

Taking part in this research study is entirely voluntary. You do not have to participate in this research. If you choose to take part, you have the right to stop at any time. If you decide not to participate or if you decide to stop taking part in the research at a later date, there will be no penalty or loss of benefits to which you are otherwise entitled.

Your decision to stop participation, or not to participate, will not influence current or future relationships with the University of Delaware.

As a student, if you decide not to take part in this research, your choice will have no effect on your academic status or your grade in the class.

You may be withdrawn from the study for one of the following reasons:

-failure to follow instructions
-the investigator decides that continuation could be harmful to you
-you need treatment not allowed in the study
-the study is canceled
-other administrative reason (e.g., necessary documentation is not in place at the time of the study)

If, at any time, you decide to end your participation on this research study please inform our research team by telling the investigator(s).

#### WHO SHOULD YOU CALL IF YOU HAVE QUESTIONS OR CONCERNS?

If you have any questions about this study, please contact the Principal Investigator, Principal Investigator, Steven J. Stanhope, Ph.D.; 540 South College Ave, Telephone: (302) 831-3496 or <a href="https://www.stanhope.org/action.org">stanhope.org</a> (302) 831-3496 or <a href="https://www.stanhope.org">stanhope.org</a> (302) 831-3496 or <a href="https://www.stanhope.org">https://www.stanhope.org</a> (302) 831-3496 or <a href="https://www.stanhope.org"/https://www.stanhope.org"/https://www.stanhope.org"/https:

If you have any questions or concerns about your rights as a research participant, you may contact the University of Delaware Institutional Review Board at <u>hsrb-research@udel.edu</u> or (302) 831-2137.

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Your signature on this form means that: 1) you are at least 18 years old; 2) you have read and understand the information given in this form; 3) you have asked any questions you have about the research and those questions have been answered to your satisfaction; 4) you accept the terms in the form and volunteer to participate in the study. You will be given a copy of this form to keep.

Printed Name of Participant	Signature of Participant	Date
Person Obtaining Consent	Person Obtaining Consent	Date
(PRINTED NAME)	(SIGNATURE)	

## OPTIONAL CONSENT FOR ADDITIONAL USES OF VIDEO RECORDINGS/PHOTOGRAPHS

I voluntarily give my permission for the researchers in this study to use videos and photographs of me collected as part of this research study to be used in publications, presentations, and/or for educational purposes. I understand that no identifying information beyond that contained in the video recording and/or photographs will be provided to educational/scientific audiences; however my facial features (and/or those of child) may be seen.

(Signature of Participant)

(Date)

(Printed Name of Participant)

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# OPTIONAL CONSENT TO REVEAL SUBJECT IDENTITY:

The data collected in this protocol may be useful to clinicians and healthcare providers to facilitate objective clinical decision-making on my behalf. Therefore, I hereby consent to allow my identity and associated data obtained in this protocol to be revealed to the following individual(s) or organization(s):

(Name of individual/organization)

(Signature of Participant)

(Date)

## OPTIONAL CONSENT TO BE CONTACTED FOR FUTURE STUDIES:

Do we have your permission to contact you regarding participation in future studies? Please write your initials next to your preferred choice.

\_\_\_\_YES \_\_\_\_NO

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# Appendix C





Figure 13 Segmental power in stance for one subject, when the signal is positive power is flowing into the segment; Note that for the pelvis segment, intact and prosthetic side refer to the stance side.



Figure 14 Segmental power in stance for one subject, when the signal is positive power is flowing into the segment; Note that for the pelvis segment, intact and prosthetic side refer to the stance side.



Figure 15 Segmental power in stance for one subject, when the signal is positive power is flowing into the segment; Note that for the pelvis segment, intact and prosthetic side refer to the stance side.