Comparison of Gait Kinematic and Temporospatial Measures Between Lock and Pin and a New Vacuum Suspension System in Transtibial Amputees

Vanessa A. Kellems

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The Graduate School

COMPARISON OF GAIT KINEMATIC AND TEMPOROSPATIAL
MEASURES BETWEEN LOCK AND PIN AND A NEW
VACUUM SUSPENSION SYSTEM
IN TRANSTIBIAL AMPUTEES

A Thesis Submitted in Partial Fulfillment
of the Requirements for the Degree of
Master of Science

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College of Natural and Health Sciences
School of Sport and Exercise Science
Biomechanics Emphasis

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Entitled: *Comparison of Gait Kinematics and Spatiotemporal Measures Between Lock and Pin and a New Vacuum Suspension System in Transtibial Amputees*

has been approved as meeting the requirement for the Degree of Master of Science in College of Natural and Health Sciences in School of Sports and Exercise Science, Program of Exercise Science

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ABSTRACT


Walking in healthy populations has been characterized by symmetrical gait patterns and used to determine variability within pathological populations. Gait patterns of individuals with transtibial amputation are commonly asymmetrical due to the inability to completely replicate the physiological ankle with prosthetics; thus, contributing to inter-limb asymmetries. Currently, only a few studies have reported the effects of suspension systems on gait mechanics. In this study, functional differences between lock and pin (PIN) and a newly developed internal vacuum suspension system (Smart PUCK™) (PUCK) during overground and treadmill walking were evaluated in five transtibial amputees. Statistically, no differences were found between groups in the distance walked during the six-minute walk test. However, clinically, the PUCK group tended to walk further compared to the PIN suspension system as suggested by the moderate effect size (0.42). Temporospatial asymmetries, stance and swing time, appeared to increase in the PUCK system compared to the PIN system but did not reach significance. The stance time symmetry index (SI) in the PUCK was more asymmetrical, thus approaching a significant difference compared to the PIN suspension system. However, stride length was remarkably similar between limbs in both systems. Kinematic asymmetries persisted in both suspension systems, but knee flexion was significantly greater in the PIN
amputated limb during swing compared to the PUCK non-amputated limb. The type of suspension system does not appear to influence the knee angular velocity and hip angles and angular velocity. Differences in hip angular velocities were observed between the amputated and non-amputated limbs. These differences were attributed to the inability to replicate the biological limb, not the type of suspension system. Differences seen between suspension system may be attributed to the significantly greater mass in the PUCK and the use of a neoprene sleeve only used in the PUCK. Based on these outcomes, the type of suspension system may influence these individual’s functional performance during treadmill and overground walking. Further work should be conducted to focus on clinical performance in both suspension systems to contrast the outcomes with quality of life measures to better inform clinicians and patients of potential advantages of suspension systems.
Table of Contents

CHAPTERS

I. GENERAL INTRODUCTION .................................................................1
   Introduction .................................................................................1
II. REVIEW OF LITERATURE .................................................................10
   General Introduction .................................................................10
   Suspension System .......................................................................13
   Lock and Pin Suspension System .................................................16
   Vacuum Suspension System ........................................................17
   SmartPuck™ ................................................................................21
   Non-Amputee Gait .........................................................................22
   Inter-Limb Symmetry .....................................................................23
   Walking Gait Pattern of Individuals with Transtibial Amputees ........24
   Temporospatial Parameter ............................................................25
   Joint Kinematics ...........................................................................26
      Ankle .......................................................................................26
      Knee .......................................................................................27
      Hip .........................................................................................28
   Treadmill vs. Overground Walking .................................................29
      Temporospatial .........................................................................30
      Kinematic .................................................................................30
      Comparison of Suspension System ............................................31
      Temporospatial .........................................................................31
      Knee .......................................................................................32
      Hip .........................................................................................33
   Functional Gait (Six-Minute Walk Test) .............................................34
   Summary .......................................................................................35
III. METHODOLOGY .............................................................................37
   Introduction ..................................................................................37
   Participants ...................................................................................37
   Materials and Data Collection .......................................................38
      Six-Minute Walk Test ................................................................38
      Motion Capture ..........................................................................39
   Data Analysis ................................................................................42
   Statistical Analysis ........................................................................43
IV. RESULTS .......................................................................................44
   Results ..........................................................................................44
   Six-Minute Walk Test .....................................................................44
   Comparison between Suspension Systems and Limbs ....................45
Temporospatial ................................................................. 45
Kinematic .......................................................................... 46
Discussion ......................................................................... 50
V. CONCLUSION .................................................................. 63
   Conclusion ....................................................................... 63
REFERENCES .................................................................... 66
APPENDIX A ..................................................................... 84
   INSTITUTIONAL REVIEW BOARD DOCUMENTS .......... 84
APPENDIX B ..................................................................... 87
   INDIVIDUAL DATA ....................................................... 87
List of Figures

2.1 Example of the Lock and Pin Suspension System................................. 14
2.2 Example of a Suction Suspension System........................................... 14
2.3 Example of a Vacuum Suspension System........................................... 15
2.4 The SmartPuck™ Suspension System.................................................. 21
3.1 Illustration of the cluster and individual markers used......................... 40
3.2 Treadmill was embedded in the ground............................................... 41
List of Tables

4.1 Functional test: 6-minute walking test ......................................................... 44
4.2 Temporospacial data for both socket condition while walking ................... 48
4.3 Symmetry index of temporospatial and kinematic data from each socket ..... 48
4.4 Mean, Standard Deviation, p-value, and effect size for both socket and limb 49
4.4 Continued ........................................................................................................ 50
1A Temporospatal data for PIN suspension system for each subject while .... 88
2A Temporospatal data between PIN and PUCK suspension system for each .... 88
3A Kinematic data in the PIN suspension system for each subject while ....... 89
4A Functional test: 6-minute walking test ............................................................ 90
5A Symmetry Index data in PIN and PUCK suspension system for each......... 90
6A Temporospatal data for PIN suspension system for each subject while .... 91
7A Kinematic data in the PIN suspension system for each subject while ....... 92
CHAPTER I

GENERAL INTRODUCTION

Introduction of the Study

Human locomotion is the movement of the lower extremities through space and include gaits such as walking and running. Walking in healthy populations has been characterized by symmetrical gait patterns and used to determine variability within pathological populations. Symmetrical gait patterns refer to inter-limb symmetries including temporospatial, kinematic, and kinetic variables (Claeys, 1983; Hamill, Bates, & Knutzen, 1984; Hannah, Morrison, & Chapman, 1984; Sadeghi, 2003; Seeley, Umberger, & Shapiro, 2008; Polk, Stumpf, & Rosengren, 2017).

Walking can be measured on either a treadmill or overground depending upon the goal of the study. In theory, if the treadmill belt speed is constant, an individual should show limited differences between overground and treadmill walking conditions. The literature indicates evidence of minor differences between treadmill and overground walking suggesting both will result in similar outcomes. Averaging participant’s overground walking velocity can be used to calculate the appropriate walking velocity for the treadmill (Alton, Baldey, Caplan, & Morrissey, 1998; Parvataneni, Ploeg, Olney, & Brouwer, 2009; Riley, Paolini, Della Croce, Paylo, & Kerrigan, 2007; Watt et al., 2010). Matching walking velocities during treadmill and overground walking has shown similar temporospatial distances (Lee & Hidler, 2008; Parvataneni et al., 2009; Riley et al., 2007; Watt et al., 2010).
It is commonly assumed the left and right limb, have symmetrical kinetic, kinematic, and temporospatial profiles thus creating an inter-limb symmetrical gait pattern (Sadeghi, 2003) during walking. Inter-limb symmetry is defined as the comparison of temporospatial, kinematic, and kinetic characteristics between the left and right limb (or: injured vs uninjured, amputated vs non-amputated) equaling 0% on the symmetry index (SI). Therefore, anything varying from zero is considered an asymmetrical gait pattern, but may still be within a normal range (see below) (Eq. 1).

$$SI = \frac{(X_A - X_N)}{(X_A + X_N)/2} \times 100$$ (1)

$X_A$ and $X_N$ represent the variable of interest measured for the injured limb ($X_A$) and non-injured limb ($X_N$) (Herzog, Nigg, Read, & Olsson, 1989). SI values of zero indicate perfect symmetry, anything varying from this is consider asymmetrical. Studies have shown non-amputee’s normal gait cycles do not reach 0% SI. A SI from 2 to 4% is within a normal symmetrical range for non-amputees (Forczek & Staszkiewicz, 2012).

Knowledge of healthy individuals is an important standard to establish which allows researchers to identify abnormalities and pathologies in individuals with injuries, surgical interventions, pathological conditions, and amputations.

Over 3 million Americans are predicted to live with limb loss by 2050 suggesting one in ninety five American’s will have an amputation due to dysvascular disease, trauma, or cancer; doubling from the estimated 1.6 million in 2005 (Ziegler-Graham, MacKenzie, Ephraim, Travison, & Brookmeyer, 2008). Roughly 25% of these individuals have undergone a transtibial amputation (TTA) equaling about 30,000 individuals (Skinner & Effeney, 1985).
After an amputation, individuals are fit for a prosthesis in which the residual limb is covered with a liner to protect the distal portion of the limb as it sits within a rigid socket that connects to the prosthetic foot. Although these individuals are able to ambulate with these devices, they do not completely replicate the structure of the physiological limb leading to inter-limb asymmetries within the gait cycle compared to a non-amputee population (Sanderson & Martin, 1997; Seliktar & Mizrahi, 1986).

Individuals with TTA exhibit temporospatial, kinematic, and kinetic walking asymmetries compared to non-amputee populations (Sadeghi, Allard, & Duhaime, 2001; Sanderson & Martin, 1997). People with TTA tend to have slower walking speeds (1.12 m·s\(^{-1}\) to 1.18 m·s\(^{-1}\)) (Hsu, Nielsen, Lin-Chan, & Shurr, 2006) compared to non-amputees (1.27 m·s\(^{-1}\) to 1.46 m·s\(^{-1}\)) (Bohannon, 1997). Commonly seen temporospatial asymmetries include: increased stance time and stride length on the non-amputated limb compared to the amputated limb, and increased stride width (Board, Street, & Caspers, 2001; Breakey, 1975; Globe, Marino, & Potvin, 2003; Grumillier, Martinet, Paysant, Andre, & Beyaert, 2008; Miller, 1987; Royer & Wasilewski, 2006; Sanderson & Martin, 1997; Xu, Greenland, Bloswick, Zhao, & Merryweather, 2017). Asymmetrical gait from a clinical perspective, can negatively contribute to secondary conditions including: increase risk of osteoarthritis (OA), redness, blistering, swelling, milking, which can impact daily lives (Hurley, McKenney, Robinson, Zadravec, & Pierrynowski, 1990; Lemaire & Fisher, 1994; Royer & Wasilewski, 2006).

Kinematic asymmetries have been suggested to be due to the missing ankle musculature of the ankle and the limited motion of the prosthetic foot (Aruin, Nicholas, & Latash, 1997; Nolan & Lees, 2000; Winter, 1983). Throughout the gait cycle the
amputated limb has demonstrated reduced ankle angle range of motion (ROM) and moments in the amputated limb compared to the intact ankle in people with TTA (Kepple, Siegel, & Stanhope, 1997; Sanderson & Martin, 1997). Furthermore, the design of the prosthetic foot has improved over the years (Donn, Porter, & Roberts, 1989; Smith & Martin, 2007); however, most of these prosthetic feet are passive devices unable to replicate the non-amputated ankle, therefore contributing to an asymmetrical gait cycle (Nolan & Lees, 2000).

At the knee, kinematic asymmetries consist of: decreased knee flexion and extensor moments during early stance and decreased knee extension and knee flexor moment during terminal stance in the amputated limb compared to the non-amputated limb (Isakov, Burger, Krajnik, Gregoric, & Marineck, 1996; Nolan & Lees, 2000; Powers, Rao, & Perry, 1998; Sanderson & Martin, 1997; Winter & Sienko, 1988).

Additionally, the non-amputated limb has shown increased joint power at the knee to compensate for the amputated limb’s significantly reduced knee power generation at push off (Nolan & Lees, 2000; Powers et al., 1998; Sadeghi et al., 2001; Siegel, Kepple, & Stanhope, 2004). Lastly, at the hip of the amputated limb, individuals with TTA have demonstrated an increased hip extension through stance (Eng & Winter, 1995; Isakov et al., 1996; Perry, 1992; Sanderson & Martin, 1997) resulting decreased hip moments and increased power generation compared to the non-amputated limb (Winter & Sienko, 1988; Gitter, Czerniecki, & DeGroot, 1991).

The socket suspension system and the socket have been considered the most important components of the prosthesis since both interface with the residual limb (Gholizadeh, Abu Osman, Eshraghi, & Ali, 2014b). Over the decades, socket suspension
systems have failed to continuously preserve residual limb volume throughout the day in consequence negatively impacting patient quality of life (Gailey, Allen, Castles, Kucharik, & Roeder, 2008). A poorly fitting prosthesis can result in ineffective transfer of forces from the socket to the residual limb in such a way that places the residual limb at risk for secondary conditions (blustering, sores, redness, etc.) (Xiaohong, Ming, & Lee, 2003).

Currently, there are three main modes of socket suspension systems: lock and pin (PIN), suction, and vacuum (Beil & Street, 2004; Smith, 2003). PIN suspension systems are characterized by a metal pin attached to the distal end of the liner which then connects to a mechanical lock attached to the proximal end of the socket (Beil & Street, 2004). This suspension system has been known for the ease of donning and doffing the prosthesis (Gholizadeh et al., 2012) where the individual presses a button to release the pin from the socket. However, this suspension system contributes to fluctuating residual limb volume which can lead to short and long-term skin changes due to improper loads on the limb (Beil, Street, & Covey, 2002; Beil & Street, 2004; Board et al., 2001; Eshraghi et al., 2014; Gerschutz, Denune, Colvin, & Schober, 2010; Sanders, Harrison, Allyn, & Myers, 2009).

Furthermore, individuals with TTA wearing a PIN displayed a greater step length on the amputated limb (0.62 m) compared to the non-amputated limb (0.54 m) as a result of a shorter stance time on the amputated limb (Gholizadeh et al., 2014b). These asymmetrical step lengths have been attributed to the asymmetrical stance and swing times seen in the PIN (66.7% vs. 61.7% of gait cycle) (Gholizadeh et al., 2014b).
These temporospatial asymmetries are interrelated with kinematic asymmetries seen at the knee and hip in the PIN suspension system. Individuals with TTA exhibit in the PIN an increased knee ROM in the amputated limb (61.5°) compared to the non-amputated limb (52.5°) (Gholizadeh et al., 2014b). While in the PIN suspension system individuals with TTA have shown a slightly greater knee flexion in the amputated limb (66.9°) compared to the non-amputated limb (52.5°) during swing (Gholizadeh et al., 2014b).

Individuals with TTA while wearing a PIN have an increased hip extension during terminal stance in the non-amputated limb compared to the amputated limb in the PIN (-2.4° vs 2.6°) (Gholizadeh et al., 2014b). Overall, individuals with TTA experienced a greater hip extension in terminal stance in the PIN due to the decreased plantarflexors in the amputated ankle and decreased proprioception hence compensating with a greater angle to allow the limb time to enter the swing phase (Gholizadeh et al., 2014b). Furthermore, the non-amputated limb in the PIN suspension system demonstrated a greater ROM in the PIN compared to the amputated limb (37.2° vs 36.1°) (Gholizadeh et al., 2014b).

Elevated vacuum systems are characterized by an external pump to remove the air molecules between the socket and residual limb allowing the entire residual limb to experience a secure fit (Ferraro, 2011). This suspension system has shown to improve overall residual limb health by increasing tissue perfusion and improving circulation therefore decreasing the amount of limb volume loss throughout the day (Rink et al., 2016). Board et al. (2001) found a net gain of 3.7% while individuals with TTA walked in a vacuum suspension system. Furthermore, maintaining this secure connection within
the socket suspension system (due to consistent negative pressure), to improves gait symmetry in people with TTA (Board et al., 2001; Gerschutz et al., 2010; Street, 2006). This is suggested to reduce skin complications (Beil et al., 2002; Beil & Street, 2004), increase spatial awareness (Street, 2006), and decrease pistoning in this system (Eshraghi et al., 2012).

While wearing a vacuum suspension system, people with TTA to have a greater step length in the non-amputated limb (0.797 m) compared to the amputated limb (0.723 m) due to the increased limb speed on the non-amputated limb thusly decreasing stance time on the amputated limb (Xu et al., 2017). This decreased stance time on the amputated limb with the vacuum suspension system trends towards a increased inter-limb symmetry, however, asymmetries were still seen between limb stance (64.8% of gait cycle) and swing time (62.7% of gait cycle) (Xu et al., 2017).

While wearing a vacuum suspension system, individuals with TTA experience increased knee angles throughout the gait cycle attributing to an increased knee ROM on the amputated limb (72.2°) compared to the non-amputated limb (65.9°) (Xu et al., 2017). During swing, the amputated limb showed a slightly greater knee flexion (66.3°) than the non-amputated limb (63.4°) but did not reach significance (Xu et al., 2017) resulting in a fairly symmetrical gait cycle.

At the hip, wearing a vacuum suspension system with a vacuum level of 20 Hg demonstrated an increases hip extension during terminal stance in the amputated limb (9.0°) compared to 0 Hg vacuum levels (8.1°) (Xu et al., 2017). Furthermore, when wearing a vacuum suspension system, individuals with TTA demonstrated a very similar inter-limb symmetry in hip ROM (52.8° vs. 53.3°) (Xu et al., 2017). The greater ROM
seen in the vacuum suspension system may be contributed to the greater step length seen in this system as well as the increased limb speed to maintain walking speeds (Xu et al., 2017).

Although vacuum suspension systems are able to maintain pressure throughout the gait cycle, improving in temporospatial and kinematic symmetry, this system still experience some volume loss due to air leakage at the hose connecting the external pump resulting in some trends toward asymmetry (Komolafe, Wood, Caldwell, Hansen, & Fatone, 2013) and volume changes in the residual limb.

The SmartPuck (PUCK) is a newly designed, adaptable vacuum suspension system that houses the SmartPuck™ within the socket preventing air leakage, which is common in other vacuum suspension systems. Furthermore, the SmartPuck™ acts similarly to other vacuum suspension systems by drawing blood, lymph, and nutrients into the residual limb therefore increasing tissue perfusion and maintain a secure fit within the system. The design of the PUCK may be similar to other vacuum suspension systems, but the internal vacuum design may improve functional mobility and improve temporospatial, kinematic, and kinetic symmetries in individuals with TTA by creating a secure fit between the residual limb and the socket throughout the day.

Currently there is no research on the performance of this system compared to PIN suspension systems. Thereupon, we evaluated functional outcomes in the PIN and the PUCK suspension systems in individuals with TTA. Therefore, the purpose of this study was to determine if socket suspension systems can improve functional ability in individuals with TTA while walking. Since maintaining a sufficient fit between the
suspension system and the residual limb has shown to improve gait symmetry it is hypothesized that:

   H1 While wearing the PUCK, participants will show increased walking speeds and increased walking distances during overground walking.

   H2 While wearing the PUCK, participants will show increased inter-limb temporospatial symmetry during treadmill walking.

   H3 While wearing the PUCK, participants will show increased inter-limb joint kinematic symmetry during treadmill walking.

Pathological populations, TTA, are commonly asymmetrical due to the inability to completely replicate the physiological ankle with prosthetics. The relationship between suspension system and the residual limb is suspected to influence individuals with TTA gait patterns. However, at present, only a few studies have reported the effects of suspension systems on gait mechanics and functional outcomes. Therefore, this study will determine if maintaining a sufficient fit between the socket suspension system and residual limb can improve functional outcomes thus gait symmetry in individuals with TTA during overground and treadmill walking.
CHAPTER II

REVIEW OF LITERATURE

General Introduction

It was estimated 1.6 million individuals were living with lower-limb loss in 2005 and by 2050 this number is expected to double to 3.2 million (Ziegler-Graham et al., 2008). Lower limb loss can result from dysvascular disease (54%), trauma (45%), malignancy (2%), and congenital abnormalities (0.8%) (Dillingham, Pezzin, & MacKenzie, 2002; Ziegler-Graham et al., 2008). Lower limb loss can occur at any level on the limb, however, TTA, account for ~30,000 of lower-limb amputations (Skinner & Effeney, 1985).

Individuals with amputation face adverse effects such as: decreased physical activity (Amtmann, Morgan, Kim, & Hafner, 2015; Desveaux et al., 2016; Paxton, Murray, Stevens-Lapsley, Sherk, & Christiansen, 2016), dissatisfaction with prosthetic system used for ambulation (Dillingham, Pezzin, MacKenzie, & Burgess, 2001; Safari & Meier, 2015), and comorbidities such as osteoarthritis and low back pain (Amtmann et al., 2015; Batten, Kuys, McPhail, Varghese, & Nitz, 2015; Gailey et al., 2008). The development of comorbidities are suggested to result from adapting to lower-limb loss and long-term prosthetic use (Gailey et al., 2008). The long-term use of a prosthetic device may contribute to inter-limb loading asymmetries.

Individuals with unilateral, TTA have demonstrated inter-limb asymmetries during walking including: temporospatial, kinematic, and kinetic measurements and
overall slower walking speed (1.12 m·s\(^{-1}\) to 1.18 m·s\(^{-1}\)) (Hsu et al., 2006) compared to non-amputees (1.27 m·s\(^{-1}\) to 1.46 m·s\(^{-1}\)) (Bohannon, 1997). Specifically, temporospatial asymmetries include: increased stance and stride length on the non-amputated limb compared to the amputated limb, and increased stride width (Board et al., 2001; Breakey, 1975; Globe et al., 2003; Grumillier et al., 2008; Miller, 1987; Royer & Wasilewski, 2006; Sanderson & Martin, 1997; Xu et al., 2017).

The missing ankle joint contributes to this asymmetrical gait pattern due to the limited motion of the prosthetic foot and the missing musculature about the ankle (Aruin et al., 1997; Nolan & Lees, 2000; Winter, 1983). People with TTA have reduced ankle angles and moments on the amputated limb throughout the gait cycle compared to the non-amputated ankle (Kepple et al., 1997; Sanderson & Martin, 1997). Although, modern prosthetic foot design has improved over the years (Donn et al., 1989; Smith & Martin, 2007), most prosthetic feet are passive devices that are unable to reproduce equivalent power to the non-amputated ankle, hence contributing to an asymmetrical gait cycle (Nolan & Lees, 2000).

People with TTA asymmetries in knee joint moments compared to non-amputated population (Nolan & Lees, 2000; Powers et al., 1998). These differences include: decreased knee flexion angle and decreased extensor moments during heel strike and increased knee extension angle and increased knee flexor moment during terminal stance in the amputated limb compared to the non-amputated limb (Isakov et al., 1996; Nolan & Lees, 2000; Powers et al., 1998; Sanderson & Martin, 1997; Winter & Sienko, 1988). These reduced knee extension angles and moments in the amputated limb result in the non-amputated limb producing more power to compensate for the amputated limb
significantly reduced knee power generation (Nolan & Lees, 2000; Powers et al., 1998; Sadeghi et al., 2001; Siegel et al., 2004).

People with TTA maintain a greater extended hip angle in the amputated limb through stance compared to the non-amputated limb (Eng & Winter, 1995; Isakov et al., 1996; Perry, 1992; Sanderson & Martin, 1997). As a result, during early through mid-stance, the amputated hip generates more positive muscular power during this phase and pulls the body forward after heel strike. This mechanism becomes more pronounced with increased walking speeds (Winter & Sienko, 1988; Gitter et al., 1991).

Although individuals with TTA experience asymmetrical gait patterns, these differences may be attributed to other factors including choice of prosthetic foot, prosthetic alignment, and socket suspension system (Pitkin, 1997). Few studies have investigated the influence of socket suspension systems on the gait of individuals with TTA (Board et al., 2001; Pitkin, 1997; Sanders, Harrison, Myers, & Allyn, 2011; Xu et al., 2017). Research suggests improving the connection of the socket to the residual limb may reduce inter-limb asymmetries in people with TTA (Board et al., 2001; Pitkin, 1997; Sanders et al., 2011; Xu et al., 2017). Individuals with TTA while wearing a vacuum suspension system at higher vacuum levels have a more symmetrical gait cycle compared to a non-vacuum condition, suggesting maintaining a constant pressure can improve this population’s gait pattern (Board et al., 2001; Sanders et al., 2011; Xu et al., 2017). However, little research has been conducted on socket suspension systems during different functional tasks such as walking and stair ascent.
Suspension Systems

Following an amputation, individuals are generally prescribed a prosthesis to assist with ambulation. For individuals with TTA, the components of a prosthesis typically include: a socket, liner, prosthetic foot, and suspension system. The prosthetic foot is designed to mimic the function of a real foot by providing degrees of freedom similar to a physiologic ankle and to assist with ambulation. The prosthetic socket is typically a rigid cast of the residual limb that connects the residual limb to the prosthesis and is designed by indenting around regions where individuals can bear more loads, such as the patellar tendon region. The liner acts as a protective barrier between the socket and residual limb to prevent skin abrasions and provides cushioning to the limb. The socket suspension system refers to how the prosthesis is physically attached to the residual limb.

It has been suggested the socket suspension system and socket are the most important components of the prosthesis because both are directly in contact with the residual limb (Gholizadeh, Abu Osman, Eshraghi, & Abd Razak, 2014a). Currently, there are three common modes of suspension systems: PIN, suction, and vacuum (Beil & Street, 2004; Smith, 2003).
PIN suspension systems attach the residual limb to the socket through a metal pin extending from the distal end of the liner that locks into a mechanical lock at the distal end of the socket (Beil & Street, 2004).

Figure 2.1. Example of the Lock and Pin Suspension System. Adapted from Scheck & Siress Prosthetics, Orthotics, 2019 (on the left) and Amputee Supply Inc., 2019.

Figure 2.2. Example of a Suction Suspension System. Adapted from Prosthetic & Orthopedic Care, 2019 (on the left) and OPC, 2017 (on the right).
A suction suspension system uses a liner with a gasket located on the exterior of the liner to create a seal between the liner and the socket. A one-way valve at the distal end of the socket allows air to escape and creates a vacuum around the distal portion of the residual limb allowing the liner to be anchored to the socket (Beil & Street, 2004). Although, the suction suspension system is similar to the vacuum suspension system, only the distal end of the residual limb is under vacuum hence this design can subject different areas of the residual limb to volume fluctuations.

The vacuum suspension system utilizes a gel liner without a pin or internal gasket. Additionally, a neoprene sleeve extends over the proximal end of the socket to the distal end of the thigh (in TTA) to create a tight seal. The vacuum is created within the socket through an active or passive pump attached to the distal end of the socket (Ferraro, 2011; Street, 2006). The suction suspension system was not investigated in this study due to similarities in applying pressure between suction and vacuum suspension systems.
Lock and Pin Suspension System

There are multiple considerations when choosing a socket suspension system. The ability to don and doff the prosthesis easily is seen as an important factor in prosthetic use (Gagnon-Gautheir, Grise, & Potvin, 1999). An appealing characteristic of the PIN suspension system is the ease of donning and doffing the prosthesis (Gholizadeh et al., 2012) is attributed to the metal pin extending from the distal end of the liner mechanical locking at the distal end of the socket (Beil & Street, 2004). Users simply need to press a button to release the pin from the socket.

The relationship between the socket and suspension system is highly important due to fluctuations in the residual limb volume throughout the day. Limb volume changes can be an obstacle for prosthetists when trying to achieve proper socket fit (Beil et al., 2002; Board et al., 2001; Gerschutz et al., 2010; Goswami, Lynn, Street, & Harlander, 2003; Zachariah, Saxena, Fergason, & Sanders, 2004). Individuals with TTA have shown on average to fall more while wearing a PIN suspension system compared to a vacuum suspension system suggesting users had an insecure fit within the socket caused by limb volume fluctuation leading to their falls (Ferraro, 2011).

A technique used by TTA to combat volume loss while wearing a PIN system, is to add ply (similar to socks) over the residual limb (Beil et al., 2002; Board et al., 2001; Gerschutz et al., 2010; Sanders et al., 2009). The addition of more ply throughout the day can be a hindrance to amputees because of the need to don and doff the prosthesis multiple times per day to add ply (Klute et al., 2011). This can be especially challenging while wearing long pants. Furthermore, an increase in pistoning (vertical movement of
the residual limb within the socket) has been associated with PIN suspension systems more than vacuum suspension systems (Klute et al., 2011).

A proper-fitting prosthesis results in the effective transfer of forces from the socket to the residual limb (Xiaohong et al., 2003). The effective transfer of force is necessary to minimize residual limb tissue damage and allow amputees to perform daily tasks without pain (Xiaohong et al., 2003). Pain and discomfort have been reported as a result of users wearing the PIN suspension system which can result in short and long-term skin changes due to improper loads on the limb (Beil & Street, 2004; Eshraghi et al., 2014). For example, during the swing phase of gait, a tension is applied distally while compression is applied proximally causing a short-term redness on the distal end of the stump, a phenomenon referred to as “milking” (the stretching of the skin at the distal end of the limb) (Beil & Street, 2004). If short-term effects are not addressed, this phenomenon can further develop into permanent long-term problems such as hyperplasia, distal bulbous shape, thickening of the distal skin, (Beil & Street, 2004; Gholizadeh et al., 2014a). Irregular pressures created during gait in the PIN could be a contributing factor to skin conditions such as edema (limb swelling) and skin ulcers (pressure sores) (Beil & Street, 2004; Levy, 1995; Salawu, Middleton, Gilbertson, Kodavali, & Neumann, 2016).

**Vacuum Suspension System**

A suspension system that can maintain pressure within the socket and liner might improve overall residual limb health by increasing tissue perfusion and improving circulation thereby reducing the amount of limb volume loss throughout the day (Rink et al., 2016). Vacuum suspension systems to provide a better connection between the prosthesis and residual limb (Board et al., 2001; Goswami et al., 2003; Klute et al., 2011;
Street, 2006). Studies have used different techniques to measure the limb volume of individuals with TTA wearing a vacuum suspension system. These techniques include: 1) anecdotal evidence of the participant’s experience with the system (Street, 2006), 2) forming a cast of the residual limb before and after walking with a vacuum suspension system (Board et al., 2001), 3) using an alginate-water mixture then filled with water and measured at three walking distances using a weight to volume conversion to determine volume change (Goswami et al., 2003).

These studies found that while wearing a vacuum suspension system, individuals with TTA, experienced less volume loss in an under-sized socket and a slight volume increase in over-sized sockets (Goswami et al., 2003). Board et al. (2001) found a net gain of 3.7% while individuals with TTA walked in a vacuum suspension system. Furthermore, these individuals with TTA experienced a 4-6% volume increase within the under sized vacuum suspension system (Board et al., 2001). Individuals who have switched to a vacuum suspension have anecdotally expressed the prosthesis feels like it is a part of the residual limb resulting in greater spatial awareness (Street, 2006) and decrease in pistoning in this system (Eshraghi et al., 2012).

This feeling of greater connection is due to the vacuum suspension system’s ability to maintain a consistent negative pressure creating a uniform contact over the entire residual limb within the socket through the entire gait cycle (Board et al., 2001; Gerschutz et al., 2010; Street, 2006). In contrast, positive pressure experienced during stance can force interstitial fluid out of the limb (back into the blood stream and lymphatic vessels) disrupting this uniform contact over the entire residual limb (Beil et al., 2002; Beil & Street, 2004; Levy, 1995; Musgrave, Zechman, & Main, 1969), causing a
decrease in volume resulting in a poor fit (Board et al., 2001). Consequently, large forces are exerted onto the distal end of the limb resulting in skin complications such as edema and venous stasis (Thirsk, Kamm, & Shapiro, 1980; Zicot, Parker, & Caro, 1977).

People with TTA exhibit a significantly greater positive pressure during stance (6.7 kPa) while in the PIN suspension system compared to suction suspension system (1.1 kPa) (Beil & Street, 2004). Maintaining negative pressure (-7 to -69 kPa) within a vacuum suspension during swing causes fluid to be drawn out of the blood stream into the limb tissues, therefore, creating a more secure fit and decreasing any extra movement within the socket (decreasing risk of skin complications) (Beil et al., 2002; Beil & Street, 2004; Chino, Pearson, & Cockroll, 1975).

At higher vacuum pressure (-69 kPa), individuals with TTA to have an average of 27% increase in negative pressure during swing and a decrease of positive pressure (~7%) during stance compared to normal total-surface weight bearing sockets (Beil et al., 2002). Board et al. (2001) reported a 3.7% increase in residual limb volume while using a vacuum suspension system (-78 kPa), suggesting that while wearing a vacuum suspension system, individuals with TTA experienced an increase limb volume through redistribution of fluid in the residual limb by drawing more fluid into the limb.

Although vacuum suspension systems are able to maintain pressure throughout the gait cycle better than PIN systems, they still experience some volume loss due to air leakage at the hose connecting the externally housed pump resulting in loss of pressure (Komolafe et al., 2013). Therefore, although better at maintaining limb volume than PIN systems, they are still imperfect and allow for some volume changes throughout the day (Beil et al., 2002; Board et al., 2001).
In addition to mitigating volume loss, vacuum suspension systems have also been shown to reduce skin complications of the residual limb due to negative pressure within the socket interface (Beil et al., 2002; Beil & Street, 2004). A case study of a transtibial diabetic amputee reported after two months of using a vacuum suspension, wounds on the distal end of the residual limb had completely closed and by three months, skin color had returned to normal (Gerschutz et al., 2010). The tightly sealed vacuum suspension created around the residual limb can result in a secure fit, thereupon, decreasing the risk of secondary injuries to their skin (Board et al., 2001; Ferraro, 2011; Gerschutz et al., 2010; Goswami et al., 2003; Klute et al., 2011; Street, 2006).

Although vacuum suspension systems have been known to successfully maintain the user’s residual limb volume, users have anecdotally expressed dislike with this system due to the time constraints of donning (Klute et al., 2011). Klute et al. (2011) reported five participants had difficulty ambulating with the vacuum suspension system, upon that leading to decreased activity levels compared to ambulating in the PIN suspension system. However, this finding is inconsistent within the literature (Board et al., 2001). Most of the literature has shown individuals with TTA have greater asymmetries while ambulating in PIN suspension system compared to vacuum systems (Ferraro, 2011; Gholizadeh et al., 2014a). Previous studies have found step length and stance time symmetry improve under a higher (67.72 kPa) vacuum level than a lower vacuum (16.93 kPa) (Board et al., 2001; Xu et al., 2017). In addition, other complaints with the vacuum suspension system include: increased sweating, skin irritation, and pain (Ali et al., 2014). Therefore, maintaining limb volume within the socket may improve fit (which can allow
for greater load onto the residual limb resulting in more symmetrical kinematic and kinetic measures) but may create other challenges not encountered with other systems. 

**SmartPuck.**

![SmartPuck](image)

*Figure 2.4. The SmartPuck™ Suspension System. Adapted from 5280 Prosthetics, 2017.*

The SmartPuck (PUCK) is a newly designed, adaptable vacuum suspension system that allows users to communicate with the system (via smart phone) to adjust the vacuum level of the system to respond to the activity level of the individual (sitting, walking, sport). The PUCK is designed to sit internally to the socket to prevent any air leakage that occurs in traditional vacuum suspension systems where the pump is housed externally to the socket. By housing the SmartPuck™ within the socket, and eliminating points for air leakage, the volume of the limb is maintained more consistently than other vacuum systems. Similar to other vacuum suspension systems, the PUCK draws blood, lymph, and nutrients into the residual limb by creating negative pressure within the socket resulting in increased tissue perfusion.

Another appealing feature of the PUCK is it gives users the ability to make easy adjustments to the vacuum level throughout the day based on their activity level.
Moreover, the developers are currently working to improve the PUCK to eventually obtain information such as: number of steps taken per day, walking speed, distance traveled, and moisture level. The information the PUCK provides can give users and prosthetists more insight into how an amputee utilizes the prosthesis and assist with prosthetic alignment.

Although the PUCK demonstrates similar characteristics as other vacuum suspension systems, the internal design may improve functional mobility and improve temporospatial, kinematic, and kinetic symmetries. However, given the suggested benefits of the PUCK, no research has been performed to compare this system to other vacuum or PIN systems.

We will investigate functional outcomes in individuals with TTA using the PIN and PUCK suspension systems. Most studies evaluate the interaction of the socket and the residual limb; however, few studies have focused on the functional mobility differences while using different suspension systems. This is the first study that will focus on the influence of structural differences between PIN and PUCK as discussed above kinematics and kinetic gait parameters.

**Non-Amputee Gait**

Human locomotion is the movement of the lower extremities through space and can include gaits such as walking and running. Walking in healthy populations has been characterized by symmetrical gait patterns. The symmetrical gait patterns generally refer to inter-limb symmetries of measures such as temporospatial measures, joint angles, moments, powers, work, and vertical ground reaction forces (GRF) (Claeys, 1983; Hamill et al., 1984; Hannah et al., 1984; Sadeghi, 2003; Seeley et al., 2008; Polk et al., 2017).
Walking parameters of healthy individuals are often used to determine variability within pathological populations.

The walking gait cycle is broken down into two general phases: stance and swing. The stance phase is characterized by one limb making contact with the ground and propelling the body forward by creating vertical impulses to counteract the force of gravity and anterior impulses to propel the body forward (Eng & Winter, 1995; Perry, 1992). This phase is measured as the total time the foot is in contact with the ground. Swing phase begins at the end of stance phase as the foot leaves the ground. The swing phase is characterized by the leg moving forward in relation to the other until it makes ground contact again. This phase is measured as the total time the foot is aerial.

**Inter-Limb Symmetry**

Frequently, it is assumed the left and right limb have symmetrical kinetic, kinematic and temporospatial measurement as a resulting in inter-limb symmetrical gait pattern (Sadeghi, 2003) during walking. Inter-limb symmetry is defined as the comparison of kinetic, kinematic, or temporospatial characteristics between the left and right limb (or: injured vs uninjured, amputated vs non-amputated) equaling 0% on the symmetry index (SI); anything varying from zero is considered an asymmetrical gait pattern, but may still be within a normal range (see below) (Eq. 1).

\[
SI = \frac{(X_A - X_N)}{(X_A + X_N)/2} \times 100
\]  

(1)

Where \(X_A\) and \(X_N\) represents the variable of interest measured for the injured limb \((X_A)\) and non-injured limb \((X_N)\) (Herzog et al., 1989). SI values of zero indicate perfect symmetry, anything varying from this is consider asymmetrical.
Walking Gait Patterns of Individuals with Transtibial Amputees

Knowledge of healthy individuals is an important standard to establish which allows researchers to identify abnormalities in individuals with injuries, surgical interventions, pathological conditions, and amputations. The purpose of this literature review is to investigate differences in gait between non-amputees and those with TTA. These individuals tend to show large inter-limb asymmetries while walking.

Walking asymmetries in individuals with TTA are attributed to multiple factors including: the loss of the ankle musculature, structural design of the prosthesis, decreased proprioception, and fluctuating limb volume (Engsberg, Lee, Patterson, & Harder, 1991; Engsberg, Lee, Tedford, & Harder, 1993; Hurley et al., 1990; Nolan & Lees, 2000; Royer & Wasilewski, 2006; Sanderson & Martin, 1997; Seliktar & Mizrahi, 1986; Suzuki, 1972; Winter & Sienko, 1988), which may contribute to this population’s inter-limb gait asymmetries.

Non-amputee normal gait cycles do not reach 0% on the symmetry index therefore suggesting an asymmetrical gait cycle. Forczek and Staszkiewicz (2012) found non-amputees SI ranged from 2 to 4% suggesting inter-limb differences. However, these noted differences were considered within a normal symmetrical range, anything less than 5% on the SI indicated symmetry. However, studies have suggested gait asymmetries shouldn’t be considered a pathological phenomenon in which Sadeghi, Allard, & Duhaime (1997) reported functional asymmetries in non-amputees suggesting inter-limb asymmetries are a reflection of the subject’s limb dominance. According to Sadeghi et al. (1997), the left limb demonstrated a significant increase in power compared to the right limb at the hip during heel strike in the frontal plane while the right limb generated more
energy at the knee during terminal stance. Although, significant differences were observed in this study, these asymmetries were not calculated using a symmetry index equation as exact numerical result was not determined for these results. Even though, differences are seen between limbs as asymmetrical gait pattern these small differences are not large enough to contribute to negative impact caused by asymmetries.

**Temporospatial Parameters**

People with TTA have demonstrated inter-limb temporospatial asymmetries including: decreased stance time, increased swing time, shorter single limb support, decreased stride length, and stride width in the amputated limb (Board et al., 2001; Breakey, 1975; Globe et al., 2003; Grumillier et al., 2008; Miller, 1987; Royer & Wasilewski, 2006; Sanderson & Martin, 1997; Xu et al., 2017).

Asymmetries seen in individuals with TTA can be attributed to the population’s slower walking speeds ranging from 1.12 m·s⁻¹ to 1.18 m·s⁻¹ (Genin, Bastien, Franck, Detrembleur, & Williams, 2008; Hsu et al., 2006). This decrease in walking speed may be a result of decreased lateral stability (Lin, Winston, Mitchell, Girlinghouse, & Crochet, 2014) created by a poor fit within the socket of the suspension system. To overcome this decrease in lateral stability, individuals with TTA tend to increase their stride width to increase their bases of support (Lin et al., 2014). Lin et al. (2014) reported individuals with TTA walking at comfortable speeds (1.27 m·s⁻¹ ± 2.29 m·s⁻¹) demonstrated a step width of 15.29 ± 5.79 cm (Lin et al., 2014) compared to non-amputees (1.27 m·s⁻¹ to 1.46 m·s⁻¹) whose preferred step width is ~ 12 cm (Bohannon, 1997; Donelan, Kram, & Kuo, 2001).
At self-selected walking speeds, non-amputated populations spend about 60% of their gait cycle in stance and the other 40% in swing (Sadeghi et al., 1997; Siegel et al., 2004). Studies have found individuals with TTA spend more time in stance in the non-amputated limb (~ 65% of gait) allowing for the amputated limb to have a longer swing time (~ 38% of gait) (Board et al., 2001; Breakey, 1975; Royer & Wasilewski, 2006; Sanderson & Martin, 1997; Siegel et al., 2004; Xu et al., 2017).

These asymmetrical stance and swing times are attributed to a 4 cm decrease in step length of the amputated limb (Xu et al., 2017). Furthermore, studies have found non-amputated controls have a slightly greater stride length compared to individuals with TTA (Houdijk, Pollmann, Groenwold, Wiggerts, & Polomski, 2009; Powers et al., 1998). Walking at a control speed of 1.31 m·s\(^{-1}\) non-amputates have a slightly longer step length (0.77m) compared to individuals with TTA step length (0.75 m) (Houdijk et al., 2009). Results may vary from previous research due to the new technology suggesting different results between studies.

**Joint Kinematics**

**Ankle.** The ankle joint of people with TTA is replaced with a prosthetic ankle that functions differently than non-amputated ankles due to the limited ROM of the prosthetic ankle, the mechanics of the prosthetic ankle itself, and the missing musculature about the ankle (Aruin et al., 1997; Nolan & Lees, 2000; Winter, 1983). In non-amputees, the ankle plantarflexors are the primary contributors for generating energy to propel the body forward and assist with controlling the forward rotation of the shank over the foot during stance (Chen, Kuo, & Andriacchi, 1997; Winter, 1983). Despite these inter-limb differences, those with TTA, show similar walking patterns to non-amputees.
However, individuals with TTA walk at slower speeds accompanied by a significant decrease in ankle plantarflexion angle during push off (Browne & Franz, 2017). People with TTA tend to have a decreased ROM in the amputated ankle due to the rigid structure of the prosthesis which limits the individual’s ability to replicate the ankle’s natural motion (Nolan & Lees, 2000). More specifically, an energy storing and releasing foot cannot actively plantarflex during push-off like a non-amputated ankle. The decreased ROM seen in the amputated ankle can result in a significantly lower angular velocity in this limb compared to the non-amputated limb (Sanderson & Martin, 1997). Studies have shown people with TTA increase ROM of the non-amputated ankle (~6°) (Nolan & Lees, 2000) to compensate for the limited ROM on the amputated ankle (Aruin et al., 1997; Miller, 1987; Winter & Sienko, 1988).

**Knee.** The missing ankle joint in people with TTA alters knee kinematics to compensate for this missing joint compared to non-amputated individuals (Nolan & Lees, 2000; Powers et al., 1998; Sanderson & Martin, 1997). These differences include: decreased initial knee flexion compared to non-amputated controls (Powers et al., 1998), decreased knee flexion and increased knee extension in the amputated limb during stance compared to the non-amputated limb (Isakov et al., 1996; Nolan & Lees, 2000; Powers et al., 1998; Sanderson & Martin, 1997; Winter & Sienko, 1988). Although these asymmetrical inter-limb knee angles continue through the entire gait cycle, studies have found minimal change in the angular velocity at the knee suggesting differences between the amputated and non-amputated are small (Bateni & Olney, 2002; Isakov et al., 1996; Powers et al., 1998; Sanderson & Martin, 1997).
People with TTA tend to have a reduced knee flexion angle on their amputated limb during the loading may suggest this is a potential protective mechanism to allow less weight to be loaded onto the amputated limb (Isakov et al., 1996; Nolan & Lees, 2000). Nolan and Lees (2000) found during stance, the amputated limb had a significantly reduced peak knee flexion (6°) compared to the non-amputated limb (30°) and non-amputated controls (24°). This decreased knee flexion is an adopted movement technique suggested to prevent large forces being placed on the soft tissue of the distal end of the residual limb (Isakov et al., 1996). Furthermore, during terminal stance, the amputated limb has increased knee extension (Powers et al., 1998; Sanderson & Martin, 1997). These asymmetries can lead to asymmetrical loads between limbs which can may increase the prevalence of OA in the non-amputated limb (Isakov et al., 1996).

Hip. People with TTA greater extended hip position through the majority of the stance phase than the non-amputated limb (Eng & Winter, 1995; Isakov et al., 1996; Perry, 1992; Sanderson & Martin, 1997). Isakov et al. (1996), found while walking at freely chosen speeds, the amputated limb had a greater maximum extension in stance compared to the non-amputated limb (7.31° vs. 6.57°), however, this difference was not significant. This trend continued with the faster walking speeds as the amputated hip was at 8.11° compared to the non-amputated hip 7.95° during extension in stance. At toe-off, the non-amputated limb had a greater extension angle (3.98°) compared to the amputated limb (3.32°) (Isakov et al., 1996). However, at faster walking speeds the amputated limb demonstrated a greater hip extension angle at toe off (4.21°) compared to the non-amputated limb (3.10°) (Isakov et al., 1996). Studies have suggested this extended position on the amputated limb is a compensation mechanism to allow the hip more time
to develop more power to compensate for the decreased power contribution from the ankle of this limb (Eng & Winter, 1995; Perry, 1992).

Although, individuals with TTA have shown small differences between the amputated and non-amputated limb throughout the gait cycle, studies have shown these differences contribute to small angular velocity suggesting little change is occurring at these points (Bateni & Olney, 2002; Isakov et al., 1996; Powers et al., 1998; Sanderson & Martin, 1997).

**Treadmill vs. Overground Walking**

Walking can be measured on either a treadmill or overground surface depending upon the goal of the study. In theory, if the treadmill belt speed is constant an individual should show limited differences between overground and treadmill walking conditions. However, some researchers have reported significant differences in some biomechanical variables between overground and treadmill conditions. Researchers have debated if treadmill walking is representative of overground walking and if temporospatial, kinematic, and kinetic measures collected on a treadmill are appropriate to use with clinical populations.

The literature indicates there are minor differences in individual’s gait patterns between treadmill and overground walking suggesting either surface is sufficient and is dependable on the focus of the study. Furthermore, studies have been averaging the participant’s overground walking velocity to calculate a walking velocity appropriate for treadmill walking (Alton et al., 1998; Parvataneni et al., 2009; Riley et al., 2007; Watt et al., 2010) suggesting any variability that might occur between surfaces due to walking
velocity is controlled for and has no influence on other measurements. These differences will be discussed below.

**Temporospatial.** Studies have demonstrated no differences in non-amputee’s temporospatial gait pattern compared to treadmill and overground walking (Lee & Hidler, 2008; Parvataneni et al., 2009; Riley et al., 2007; Watt et al., 2010). Parvataneni et al. (2009) found in non-amputated individuals matching speeds during treadmill and overground walking resulted in similar temporal-distance parameters. Additionally, Watt et al. (2010) reported similar walking velocities averaging 1.25 m·s⁻¹ on treadmill and 1.27 m·s⁻¹ on overground in older non-amputated adults resulted in generally similar kinematic and kinetic measurements between the two surfaces. Watt et al. (2010), reported in elderly non-amputated participants a decrease in stride length of 0.09m and 0.06s in stride time during treadmill walking. However, these older non-amputated individuals demonstrated significantly shorter stride time and length during treadmill walking corresponding to an increased cadence. Studies have shown non-amputated individuals walking at controlled speeds (1.25 m·s⁻¹ treadmill and 1.27 m·s⁻¹ overground) and self-selected speeds had a significantly higher cadence while walking on a treadmill (Alton et al., 1998).

**Kinematics.** Studies examining kinematics during treadmill and overground walking in non-amputated and amputated individuals have reported significant differences between groups at the ankle, knee, and hip (Alton et al., 1998; Button, Moyle, & Davids, 2010; Lee & Hidler, 2008; Riley et al., 2007; Watt et al., 2010). While walking at controlled speeds (1.25 m·s⁻¹ treadmill and 1.27 m·s⁻¹ overground) and self-selected
speeds those with TTA increased flexion and ROM at the hip on the treadmill (Alton et al., 1998).

Contrary to Alton et al. (1998), Watt et al. (2010) reported elderly non-amputated participants had a decreased hip extension and flexion during stance, increased knee flexion during stance, decreased ankle dorsiflexion, and plantarflexion angles through gait resulting in a decreased ROM at the knee on a treadmill compared to overground. Furthermore, other studies have reported similar findings to Watt et al. (2010) supporting a decrease in knee ROM during treadmill walking (Lee & Hidler, 2008; Riley et al., 2007). Even though these studies have found differences between treadmill and overground walking, studies have suggested these differences are minor and are typically less than 2° or 3° in difference (Button et al., 2010; Parvataneni et al., 2009; Riley et al., 2007; Watt et al., 2010) suggesting walking on these surfaces will do not constitute clinically meaningful differences.

There is very little literature examining direct differences in individuals with TTA between treadmill and overground walking. One study examined three participants with TTA gait kinematics in treadmill and overground walking found mean differences to be less than 3° at the knee and hip between treadmill and overground walking (Button et al., 2010). However, these results should be interpreted with caution as having only three participants is difficult to justify generalizing these results to a TTA population.

**Comparison of Suspension Systems**

**Temporospatial.** People with TTA wearing a vaccum suspension system to have a longer step length on the amputated limb (0.723 m) (Xu et al., 2017) compared to these individuals step length while wearing a PIN (0.62 m) (Gholizadeh et al., 2014b). While
wearing a PIN, individuals with TTA have a greater step length on the amputated limb (0.62 m) compared to the non-amputated limb (0.54 m) as a result of a longer swing time on the amputated limb (Gholizadeh et al., 2014b). However, this trend was different when individuals with TTA wore a vacuum suspension system in which the users had a greater step length on their non-amputated limb (0.797 m) compared to the amputated limb (0.723 m) due to the increased limb speed on the non-amputated limb thus decreasing stance time on the amputated limb (Xu et al., 2017). Furthermore, these individuals demonstrated a greater symmetrical step length while wearing the vacuum compared to the PIN.

However, in the PIN (66.7 % vs. 61.7% of gait cycle) and vacuum (64.8% vs. 62.7% of gait cycle) individuals continue to show a greater stance time on the non-amputated limb compared to the amputated (Gholizadeh et al., 2014b; Xu et al., 2017) resulting in an increased swing time on the amputated limb (Gholizadeh et al., 2014b; Xu et al., 2017).

Structural differences between the suspension systems may have contributed to the differences seen within these studies as the individuals with TTA walked at faster speeds in the vacuum suspension system (1.2 m·s⁻¹ to 1.4 m·s⁻¹) (Xu et al., 2017) compared to in a PIN (0.93 m·s⁻¹) (Gholizadeh et al., 2014b). Individuals with TTA faster walking speeds in the vacuum suspension system may be due to a secure fit resulting in greater confidence with the prosthesis (Gholizadeh et al., 2014b; Pitkin, 1997; Xu et al., 2017).

**Knee.** People with TTA have demonstrated greater knee angles throughout the gait cycle while wearing a vacuum suspension system correlating to an increased ROM
on the amputated limb (72.2°) (Xu et al., 2017) compared to a PIN (61.5°) (Gholizadeh et al., 2014b). The amputated limb a greater ROM compared to the non-amputated limb in both the vacuum suspension system (72.2° vs. 65.9°) (Xu et al., 2017) and a PIN (61.5° vs. 52.5°) (Gholizadeh et al., 2014b).

Individuals with TTA demonstrated a slightly greater knee flexion during the swing phase in the amputated limb while in the PIN (66.9°) (Gholizadeh et al., 2014b) compared to the vacuum (66.3°). While wearing a vacuum suspension system, the amputated limb showed a slightly greater knee flexion (66.3°) than the non-amputated limb (63.4°) but did not reach significance (Xu et al., 2017). This trend was similar while wearing a PIN where the amputated limb (66.9°) had a greater knee flexion compared to the non-amputated limb (52.5°) (Gholizadeh et al., 2014b). Although, the PIN demonstrated a greater knee flexion during swing, these individuals with TTA had a greater asymmetrical gait pattern compared to the vacuum system (Gholizadeh, et al., 2014b).

**Hip.** While walking in a vacuum suspension system, the amputated limb had greater hip extension (9.0°) during the stance phase (Xu et al., 2017) than while wearing a PIN (2.6°) (Gholizadeh et al., 2014b). Individuals with TTA had a greater hip extension angle during terminal stance in the non-amputated limb compared to the amputated limb in the PIN (-2.4° vs 2.6°) (Gholizadeh et al., 2014b). However, in the vacuum suspension system, the amputated limb demonstrated a greater hip extension in terminal stance compared to the non-amputated limb (9.0° vs 12.2°) (Xu et al., 2017). Overall, individuals with TTA experienced a greater hip extension in terminal stance in the PIN compared to the vacuum due to the decreased plantarflexors in the amputated ankle and
decreased proprioception therefore compensating with a greater angle to allow the limb time to enter the swing phase (Gholizadeh et al., 2014b; Xu et al., 2017). Individuals with TTA showed to have the closer inter-limb symmetry at the hip during swing in the vacuum compared to the PIN suspension system (Gholizadeh et al., 2014b; Xu et al., 2017).

People with TTA have a larger difference in the amputated hip ROM between the vacuum and PIN suspension system (53.3° vs. 36.1°). Additionally, in the sagittal plane the non-amputated and amputated limb showed a very similar ROM in a vacuum (52.8° vs. 53.3°) (Xu et al., 2017) and PIN (37.2° vs. 36.1°) (Gholizadeh et al., 2014b). The greater ROM seen in the vacuum suspension system may be contributed to the greater step length seen in this system as well as the increased limb speed to maintain walking speeds (Xu et al., 2017).

**Functional Gait (Six-Minute Walk Test)**

The six-minute walk test (6MWT) is commonly used by clinicians to measure an individual’s maximal distance covered in a six-minute time period (Lin et al., 2014) reflecting onto a population’s ability to ambulate in day to day activities in the community (Lin et al., 2014; Lin & Bose, 2008; Linberg et al., 2013). The 6MWT has been reported as a reliable and valid test - retest for individuals with TTA (Gailey et al., 2002; Lin & Bose, 2008; Lin et al., 2014; Lin & Bose, 2008).

The 6MWT was originally derived from a 12-minute running test used to measure healthy adult’s exercise capacity (McGavin, Gupta, & McHardy, 1976) but then later modified into the two-minute walk test (2MWT) and 6MWT (Butland, Pang, Gross, Woodcook, & Geddess, 1982). Although, this test was designed for individuals with
cardiac or respiratory diseases (American Thoracic Society, 2002; Sadaria & Bohannon, 2001; Solway, Brooks, Lacasse, & Thomas, 2001) it has been used for individuals of different populations including: children (Li et al., 2005), elderly (Harada, Chiu, & Stewart, 1999), cerebral palsy (Anderson, Asztalos, & Mattson, 2006), and amputation (Gailey et al., 2002).

People with TTA walk an average of 556.2 m over three trials during the 6MWT (Lin & Bose, 2008). Another study conducted by Lin et al. (2014) found individuals with TTA walked 513.29 m ± 137.05 during the 6MWT. However, studies using the 6MWT to measure structural differences of prosthesis including 4 different types of feet have found no significant differences between walking distance among these prosthetic feet (Gailey et al., 2008; Linberg et al., 2013; Wunderman, Schmid, Myers, Jacobsen, & Stergiou, 2017). Although, these studies found no significant differences, Linberg et al. (2013) suggested the use of the 6MWT provides an understanding into the functional ability of people with TTA.

**Summary**

Individuals living with lower-limb loss is a growing population expected to reach 3.2 million by 2050 (Ziegler-Graham et al., 2008) in which ~30,000 are TTA. These individuals tend to show large inter-limb asymmetries including: temporospatial, kinematic, and kinetic measurements during overground and treadmill walking (Bohannon, 1997; Hsu et al., 2006). Currently there is little literature investigating differences in gait in suspension systems in TTA. Research suggest asymmetries seen in TTA may be contributed to the design of the suspension system causing these
abnormalities. Therefore, determining if socket suspension systems can improve functional ability in individuals with TTA while walking would enhance the literature.
CHAPTER III

METHODOLOGY

Introduction

Participants

Participants with unilateral, TTA were recruited from the Northern Colorado region (n = 5 males; 97.4 ± 22.04 kg, 1.67 ± 0.12 m; 56.8 ± 11.2 years. Inclusion criteria included: 18 to 65 years of age, TTA resulting from: trauma, bone cancer, or birth defect, no neurological, cardiac, or vascular problems that could limit their function, no diagnoses of health condition that affects muscle function, a healthy residual limb (no pressure sores or ulcers), currently wearing a lock and pin or vacuum suspension system, and have at least 6 months of experience in their current prosthesis. Additionally, participants must have had a body mass index less than 35 kg·(m²), rated with level K3 and K4 amputations, and were able to walk for 10 minutes continuously without assistance. People classified as a K3 are capable of walking at comparable walking cadences and are able to ambulate across most environment obstacles compared with non-amputees. A classification of K4 includes the capability exceeding individual’s basic locomotion measured as the individual exhibiting greater levels of “stress, impact, or energy” (HCFA, 2001). Prosthetists assessed each participant’s physical activity prior to the start of the study to determine if the participant met the inclusion criteria. Exclusion criteria: Participants were excluded from this study if they were unable to walk without
assistance for less than 10 minutes or developed any external injury to the residual limb (sores, ulcers, etc.).

The study was designed to evaluate the influence two different type of suspension systems (PIN and PUCK) on gait symmetry over two separate visits. Participants were randomly assigned to either their original suspension system or to the alternative suspension (PIN or PUCK) system and were fitted by certified prosthetists. Participants were allowed a minimum of one week to acclimate to each suspension system before data collection.

The data collection for this study was conducted at the University of Northern Colorado Biomechanics Lab in Gunter Hall. This study obtained approval by the Institutional Review Board at the University of Northern Colorado. Upon arrival to the Biomechanics Lab, participants provided their written and verbal consent before starting the data collection.

Participants were asked to change into tight-fitting clothes to facilitate motion capture. Anthropometrics (height, mass, mass of the prosthetic limb) and general history were obtained from each participant for the use of the labs records and inputs to create three-dimensional models of the participants for further analysis.

**Materials and Data Collection**

**Six-Minute walk test.** Upon arrival to the lab, participants completed the six-minute walk test (6MWT) which measures the distance an individual can walk in six minutes. The 6MWT is a functional test that a clinically assesses a participant’s functional mobility (Gailey et al., 2002) as an indication of their ability to ambulate in their daily life (Lin & Bose, 2008). This functional test has been shown to be a reliable
and valid test used to identify health, populations and individuals with TTA performance of mobility, physical function, and aerobic capacity (Gailey et al., 2002; Lin & Bose, 2008; Linberg et al., 2013).

A section of the Gunter hall corridor was marked off with cones 100 ft apart to indicate a course for the participant to follow. Participants were provided the following instructions for the 6MWT: “The objective of this test is to walk as far as possible in 6 minutes. You can go around the cones however you like.” Researchers did not speak with the participants to prevent distracting or encouraging the participants as they walked. During the last 15 seconds of the trial researchers trailed behind the participant to obtain the ending location (distance) at the end of the 6 minutes.

**Motion capture.** Retroreflective markers were placed over anatomical landmarks on the upper and lower body along with six lower extremity clusters and four upper extremity clusters (e.g. thigh, forearm) using Coban™ and hypoallergenic tape (Fig. 3.1). Sixteen individual markers were removed after calibration as these markers were only used to identify joint axis orientation. A 10-camera motion capture system (VICON, Oxford, UK) was used to collect motion data at 100 Hz.
Figure 3.1. Illustration of the cluster and individual markers used on participants for data collection. The striped markers indicate markers used for calibration only and the solid color are the tracking markers that remained on the participant during the entire collection.

Kinematic data (100 Hz) collected on a tandem belt treadmill (AMTI, Watertown, MA) (Fig. 3.2). Participants were asked to walk for six minutes on the treadmill. The last 30 seconds were analyzed to ensure the participant had acclimatized to the treadmill. Participants were instructed to stand toward the front of the treadmill and to hold the safety bar as the treadmill started. They were allowed to hold the bar until they felt confident enough to let go and walk unaided on the treadmill. Additionally, the
participants were informed to use the emergency button attached to the top of the safety bar any time they needed to immediately stop the treadmill. Furthermore, two sets of color wraps were attached to the bar to indicate the width of the treadmill (still within safe range) (Fig. 3.2).

Figure 3.2. Treadmill was embedded in the ground. An emergency stop button and cord were mounted on the rail in front of the participant in case of emergency.

The treadmill started at a slow velocity, gradually working towards the participant’s self-selected velocity. Once participant’s velocity was determined, participants were instructed to walk for six minutes to allow for an acclimation to the treadmill. Studies have suggested non-amputated adults require only 2 to 3 minutes to familiarize themselves with the treadmill while others have suggested 4 to 6 minutes is an appropriate length of time to allow for the participant to acclimate; correspondingly, 6
minutes in the current study is reasonable and does not run the risk of causing fatigue in the participants (Alton et al., 1998; Matsas, Taylor, & McBurney, 2000; Parvataneni et al., 2009).

As the participants felt they had reached a comfortable velocity they could maintain for the 6-minute walk test, participants were instructed to move and maintain their body position over the middle of the treadmill. Over the duration of the test, the participants were periodically asked about their comfort on the treadmill and walking velocity.

**Data Analysis**

The last 30 seconds of the 6-minute walking trial on the treadmill were captured to allow for at least three successful gait cycles. A successful gait cycle was defined as heel strike to heel strike of the same foot. Kinematic data were filtered with a low-pass fourth-order, zero-lag, recursive Butterworth digital filter (F<sub>c</sub> = 6 Hz). Models were created in Visual3D (C-motion, Germantown, MD) using the subject’s anthropometric data and inertial measures from de Leva (1996) and Ferris, Smith, Heise, Hinrichs & Martin (2017). Outcome variables included: temporospatial measures (stance and swing time, step length, stride length stride) and kinematic measures (joint angles, angular velocity, timing) for each lower extremity joint and were used to calculate a symmetry index (SI) between the two limbs (non-amputated and amputated):

\[
SI = \frac{(X_A - X_N)}{(X_A + X_N)/2} \times 100
\]  

Where \(X_A\) represents the gait variables measured for the amputated limb and \(X_N\) represents the gait variables measured for the non-amputated (Board et al., 2001).

Temporospatial measures were calculated in Visual 3D were filtered at 60 Hz to capture
GRF allowing for initial and final foot contact (Germantown, MD). SI values of zero indicate perfect symmetry, anything varying from this is considered asymmetrical.

**Statistical Analysis**

A MANOVA a priori contrast technique was used to identify significant differences of walking symmetry between the limbs and suspension systems of the PIN and PUCK ($\alpha = .05$). An alpha level of 0.05 was used to determine any significant effects of the suspension system on gait symmetry. Furthermore, due to the small sample size, effect sizes were calculated to determine meaningfulness of any differences. Effect sizes (ES) were computed based on Cohen’s d (Thomas, Silverman, & Nelson, 2015).

$$ ES = \frac{M_1 - M_2}{S_p} $$

(2)

Where $M_1$ is the mean for the PIN suspension system and $M_2$ is the mean for the PUCK suspension system. Effect sizes greater than or equal to 0.8 were considered large, effect sizes of 0.5 were considered moderate, and effect sizes equal to or less than 0.2 were considered small. Pooled standard deviation ($S_p$) was computed as:

$$ S_p = \frac{\sqrt{S_1^2 + S_2^2}}{2} $$

(3)

where $S_1$ and $S_2$ are the standard deviations of the PIN and PUCK suspension system.
CHAPTER IV

RESULTS

Results

All participants reported in this study successfully completed both sessions. Participants demographics were discussed above in the previous chapter. Total limb mass (socket, pylon, foot, shoe) for the PIN (1.99 ± 0.21 kg) was significantly lighter ($p = .003$) than the PUCK (2.31 ± 0.24 kg). Walking velocities ranged from 0.5 - 1.6 m·s$^{-1}$, average 1.05 ± 0.44 m·s$^{-1}$.

Six-Minute Walk Test

Participants walked similar distances while wearing the PUCK (1473 ± 124.88 ft) and PIN (1405 ± 196.16 ft) suspension systems ($F(1,18) = 0.342$, $p = 0.575$) during the 6MWT. Although not significantly different the effect size shows the suspension system has a moderate effect (0.41) on the individual ability to walk for a long period of time. Anicdotally, all participants, except for one, walked farther during in the PUCK than the PIN. Individual subject data for be seen in Table 6A.

Table 4.1

<table>
<thead>
<tr>
<th>Functional test: six-minute walking test</th>
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</thead>
<tbody>
<tr>
<td>6MWT (ft) PIN</td>
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<tr>
<td>-----------------------------------------</td>
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<tr>
<td>1405 ± 196.16</td>
</tr>
</tbody>
</table>

Note: $p^{a}$ represents significance between PIN and PUCK suspension systems
Comparison between Suspension Systems and Limbs

Temporospatial

No significant differences were seen between the PIN and PUCK amputated limb. A few dependent variables in the PIN and PUCK amputated limb; stance time (as percent of gait), swing time, swing time (as percent of gait), were found to be with moderate effect sizes (Table 4.4). No significant differences were seen between the PIN and PUCK non-amputated limb. Swing time between the PIN and PUCK non-amputated limbs had a moderate found effect size (Table 4.4).

Both suspension systems demonstrated commonly reported temporospatial asymmetries in individuals with TTA. Although not significant, some trends were noted between systems and limbs. More specifically, the PUCK amputated limb demonstrated a trend toward decreased stance time (as percent of gait cycle) compared to the PUCK non-amputated limb ($F(1,16) = 3.61$, $p = 0.076$) and PIN non-amputated limb ($F(1,16) = 4.01$, $p = 0.062$) (Table 4.4). Although these trends did not reach significance, effect size shows this dependent variable was largely influenced by the suspension and limb. Furthermore, the PUCK amputated limb demonstrated a trend towards increased swing time (as percent of gait cycle) compared to the PIN non-amputated ($F(1,16) = 3.89$, $p = 0.066$) (Table 4.4). Although not significant, the effect size of this relationship shows the suspension system and limb may have a large influence of the individuals swing time (as a percent of gait).

Double limb support time (DLST) is the time when there are two feet on the ground during the gait cycle. Non-amputated limb DLST is measured as heel strike on the non-amputated limb to toe-off in the amputated limb while amputated limb DLST is
measured as the amputated limb at heel-strike to toe-off in the non-amputated limb. Both systems showed similar DLST between the amputated and non-amputated limbs (Table 4.2). There were no significant differences between the systems although the both PUCK limbs tended to be slightly shorter than the PIN limbs. Stride width was also similar between suspension systems (Table 4.2).

**Kinematic**

No significant differences were found at the knee and hip angles between the amputated limbs in each of the suspension systems. Although not significant, the knee flexion during swing of the PUCK amputated limb was \( \sim 8^\circ \) smaller than the PIN amputated limb \((F(1,16) = 2.69, p = 0.104)\) resulting in a large effect size of 1.09 (Table 4.4). Furthermore, no significant differences were found at the knee and hip angular velocity between the amputated limbs in each of the suspension systems (Table 4.4).

No significant differences were found at the knee and hip angles between the amputated limbs in each of the suspension systems. Although not significant, the knee flexion during swing this difference between suspension systems lead to a large effect size (1.09) (Table 4.4). Furthermore, no significant differences were found at the knee and hip angular velocity between the amputated limbs in each of the suspension systems (Table 4.4).

No significant differences were found in the knee joint angle between the limbs and suspension system during any point during stance. The PIN amputated limb had significantly increased knee flexion through swing compared to the PUCK non-amputated limb \((F(1,16) = 4.41, p = 0.033)\) with a large effect size (1.14) (Table 4.6). No
significant differences were found between the suspension system and limbs in knee angular velocity.

No significant differences were found between suspension system and limbs in the hip joint angle. However, significant differences were seen in hip angular velocity between the limbs and suspension system during early stance. Specifically, the PUCK amputated limb hip flexion angular velocity during early stance was significantly (~27 deg·s⁻¹) larger than the PUCK non-amputated limb ($F(1,16) = 4.41, p = 0.019$) (Table 4.6) with a large effect size (1.3). Similarly, the PIN amputated limb hip flexion angular velocity during stance was significantly (~21 deg·s⁻¹) larger than the PIN non-amputated limb ($F(1,16) = 4.41, p = 0.021$) (Table 4.6) with a large effect size (1.46). Furthermore, the PUCK non-amputated was significantly (~171.31 deg·s⁻¹) larger than the PIN amputated limb ($F(1,16) = 4.40, p = 0.029$) (Table 4.6). Additionally, the the PUCK amputated limb was significantly (~24 deg·s⁻¹) larger than the PIN non-amputated limb ($F(1,16) = 4.39, p = 0.014$) (Table 4.6) with a moderate effect size (0.54).
**Table 4.2**

*Temporospatial data for both socket conditions while walking.*
*Data are shown as mean ± standard deviation.*

<table>
<thead>
<tr>
<th></th>
<th>PIN</th>
<th>PUCK</th>
<th>p</th>
<th>EFFECT SIZE</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stride Width (m)</td>
<td>0.16 ± 0.03</td>
<td>0.15 ± 0.05</td>
<td>0.767</td>
<td>0.25</td>
</tr>
<tr>
<td>Non-amputated DLST (s)</td>
<td>0.19 ± 0.08</td>
<td>0.16 ± 0.04</td>
<td>0.548</td>
<td>0.47</td>
</tr>
<tr>
<td>Amputated DLST (s)</td>
<td>0.19 ± 0.04</td>
<td>0.17 ± 0.04</td>
<td>0.816</td>
<td>0.5</td>
</tr>
</tbody>
</table>

Note: Non-amputated DLST is measured as heel-strike on the non-amputated limb to toe-off on the amputated limb. Amputated DLST is measured as heel-strike on the amputated limb to toe-off on the non-amputated limb. p* represents significant difference between the PIN and PUCK suspension system.

**Table 4.3**

*Symmetry Index of temporospatial and kinematic data for each socket condition.*
*Data are shown as mean ± standard deviation.*

<table>
<thead>
<tr>
<th></th>
<th>PIN</th>
<th>PUCK</th>
<th>p</th>
<th>EFFECT SIZE</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cycle Time (%)</td>
<td>-1.12 ± 1.03</td>
<td>1.30 ± 1.28</td>
<td>0.649</td>
<td>0.29</td>
</tr>
<tr>
<td>Stance Time (%)</td>
<td>-3.88 ± 9.54</td>
<td>-6.06 ± 4.94</td>
<td>0.986</td>
<td>0.40</td>
</tr>
<tr>
<td>Swing Time (%)</td>
<td>7.77 ± 6.22</td>
<td>10.83 ± 8.93</td>
<td>0.180</td>
<td>0.42</td>
</tr>
<tr>
<td>Stride Length (%)</td>
<td>2.58 ± 9.05</td>
<td>-0.13 ± 1.54</td>
<td>0.147</td>
<td>0.16</td>
</tr>
<tr>
<td>DLST (%)</td>
<td>-4.61 ± 30.96</td>
<td>-8.13 ± 5.89</td>
<td>0.833</td>
<td>0.41</td>
</tr>
<tr>
<td>Knee ROM (%)</td>
<td>-11.63 ± 33.66</td>
<td>0.44 ± 4.06</td>
<td>0.497</td>
<td>0.50</td>
</tr>
<tr>
<td>Hip ROM (%)</td>
<td>-0.71 ± 26.34</td>
<td>12.84 ± 15.38</td>
<td>0.400</td>
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</tbody>
</table>

Note: Symmetry Index (SI) is measuring the symmetry between the amputated and the non-amputated limb. SI values of zero indicate perfect symmetry, anything varying from this is consider asymmetrical. p* represents significant difference between the PIN and PUCK suspension system.
<table>
<thead>
<tr>
<th>Temporo-spatial</th>
<th>PIN</th>
<th>PUCK</th>
<th>NON-AMPUTATED</th>
<th>AMPUTATED</th>
<th>NON-AMPUTATED</th>
<th>AMPUTATED</th>
<th>p^b</th>
<th>p^c</th>
<th>p^d</th>
<th>p^e</th>
<th>p^f</th>
<th>ES^a</th>
<th>ES^b</th>
<th>ES^c</th>
<th>ES^d</th>
<th>ES^e</th>
<th>ES^f</th>
<th>ES^g</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cycle Time (s)</td>
<td>1.08 ± 0.11</td>
<td>1.07 ± 0.10</td>
<td>1.05 ± 0.10</td>
<td>1.07 ± 0.11</td>
<td>0.957 0.650 0.789 0.851 0.809 0.831 0 0.29 0.2 0.05 0.09 0.19</td>
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<tr>
<td>Stance Time (s)</td>
<td>0.74 ± 0.11</td>
<td>0.70 ± 0.09</td>
<td>0.71 ± 0.09</td>
<td>0.68 ± 0.08</td>
<td>0.716 0.670 0.829 0.522 0.321 0.563 0.03 0.30 0.01 0.06 0.62 0.35</td>
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<tr>
<td>Stance Time (%)</td>
<td>67.62 ± 2.71</td>
<td>65.04 ± 2.98</td>
<td>67.40 ± 3.22</td>
<td>63.48 ± 4.00</td>
<td>0.459 0.919 0.271 0.231 0.062 0.076 0.44 0.07 0.8 0.91 1.21 1.08</td>
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<tr>
<td>Swing Time (s)</td>
<td>0.35 ± 0.01</td>
<td>0.37 ± 0.03</td>
<td>0.39 ± 0.06</td>
<td>0.34 ± 0.03</td>
<td>0.534 0.814 0.240 0.341 0.125 0.082 0.42 0.45 1 0.89 0.93 1.03</td>
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<tr>
<td>Swing Time (%)</td>
<td>32.40 ± 2.70</td>
<td>35.02 ± 2.98</td>
<td>36.45 ± 3.89</td>
<td>32.68 ± 3.30</td>
<td>0.496 0.892 0.273 0.221 0.066 0.086 0.41 0.09 0.74 1.03 1.21 1.05</td>
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<tr>
<td>Stride Length (m)</td>
<td>1.10 ± 0.33</td>
<td>1.12 ± 0.30</td>
<td>1.15 ± 0.31</td>
<td>1.15 ± 0.31</td>
<td>0.898 0.643 0.883 0.543 0.630 0.985 0.32 0.16 0.32 0.30 0.16 0</td>
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<tr>
<td>Knee Peak 0 - 5%</td>
<td>7.98 ± 5.32</td>
<td>5.02 ± 7.98</td>
<td>5.89 ± 7.24</td>
<td>8.58 ± 4.38</td>
<td>0.832 0.832 0.391 0.457 0.613 0.515 0.11 0.12 0.55 0.44 0.31 0.45</td>
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<tr>
<td>Knee Peak 10 - 20%</td>
<td>16.12 ± 6.30</td>
<td>13.36 ± 9.46</td>
<td>14.45 ± 8.43</td>
<td>17.91 ± 3.96</td>
<td>0.817 0.817 0.342 0.561 0.724 0.467 0.12 0.34 0.63 0.34 0.22 0.52</td>
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<tr>
<td>Knee Peak 30-50%</td>
<td>8.01 ± 7.54</td>
<td>9.92 ±11.89</td>
<td>10.04 ± 10.46</td>
<td>5.75 ± 6.93</td>
<td>0.984 0.752 0.495 0.752 0.737 0.482 0.01 0.31 0.43 0.19 0.22 0.48</td>
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<tr>
<td>Knee Peak 60-80%</td>
<td>58.74 ± 7.59</td>
<td>63.91 ± 8.09</td>
<td>55.54 ± 7.27</td>
<td>52.53 ± 7.80</td>
<td>0.104 0.43 0.033* 0.66 0.104 0.81 1.09 0.81 1.14 0.66 0.43 0.40</td>
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<tr>
<td>Knee ROM 0-100%</td>
<td>55.33 ± 5.23</td>
<td>52.00 ± 19.43</td>
<td>54.42 ± 6.87</td>
<td>54.33 ± 8.11</td>
<td>0.109 0.149 0.134 0.253 0.614 0.906 0.17 0.15 0.16 0.23 0.15 0.01</td>
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<tr>
<td>Hip Peak 5 – 15%</td>
<td>24.99 ± 7.89</td>
<td>25.82 ± 10.91</td>
<td>26.94 ± 10.98</td>
<td>26.86 ± 4.38</td>
<td>0.857 0.203 0.866 0.895 0.755 0.990 0.10 0.30 0.13 0.09 0.52 0.01</td>
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<tr>
<td>Hip Peak 45 – 65%</td>
<td>-7.10 ± 12.60</td>
<td>-6.45 ± 15.50</td>
<td>-3.90 ± 18.74</td>
<td>-4.27 ± 15.80</td>
<td>0.803 0.200 0.831 0.949 0.753 0.971 0.15 0.20 0.14 0.05 0.20 0.02</td>
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<tr>
<td>Hip Peak 70-90%</td>
<td>27.49 ± 7.69</td>
<td>30.82 ± 10.36</td>
<td>30.96 ± 10.42</td>
<td>30.23 ± 13.11</td>
<td>0.984 0.379 0.930 0.625 0.914 0.611 0.01 0.23 0.22 0.37 0.38 0.06</td>
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<tr>
<td>Hip ROM 0-100%</td>
<td>35.65 ± 8.06</td>
<td>36.01 ± 10.20</td>
<td>36.73 ± 7.04</td>
<td>33.07 ± 9.89</td>
<td>0.667 0.143 0.628 0.534 0.846 0.956 0.08 0.25 0.29 0.04 0.14 0.43</td>
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### Table 4.4 Continued

<table>
<thead>
<tr>
<th>Angular Velocity (°s⁻¹)</th>
<th>Knee Peak 0 - 5%</th>
<th>Knee Peak 10 - 20%</th>
<th>Knee Peak 30-50%</th>
<th>Knee Peak 60-80%</th>
<th>Knee Peak 80-90%</th>
<th>Hip Peak 5 –15%</th>
<th>Hip Peak 45 – 65%</th>
<th>Hip Peak 60-80%</th>
</tr>
</thead>
<tbody>
<tr>
<td>Knee Peak 0 - 5%</td>
<td>7.35 ± 41.70</td>
<td>-23.20 ± 52.48</td>
<td>30.94 ± 51.53</td>
<td>-3.66 ± 112.79</td>
<td>0.222</td>
<td>1.04</td>
<td>0.645</td>
<td>0.394</td>
</tr>
<tr>
<td>Knee Peak 10 - 20%</td>
<td>71.09 ± 71.72</td>
<td>30.94 ± 51.53</td>
<td>86.42 ± 45.17</td>
<td>108.44 ± 46.65</td>
<td>0.686</td>
<td>0.331</td>
<td>0.413</td>
<td>0.674</td>
</tr>
<tr>
<td>Knee Peak 30-50%</td>
<td>83.42 ± 165.70</td>
<td>102.40 ± 67.43</td>
<td>176.75 ± 174.14</td>
<td>139.39 ± 119.85</td>
<td>0.326</td>
<td>0.552</td>
<td>0.869</td>
<td>0.411</td>
</tr>
<tr>
<td>Knee Peak 60-80%</td>
<td>270.26 ± 164.00</td>
<td>98.91 ± 112.78</td>
<td>310.15 ± 92.19</td>
<td>286.84 ± 96.34</td>
<td>0.637</td>
<td>0.710</td>
<td>0.403</td>
<td>0.782</td>
</tr>
<tr>
<td>Knee Peak 80-90%</td>
<td>-168.65 ± 211.53</td>
<td>238.83 ± 155.61</td>
<td>-276.30 ± 81.85</td>
<td>-280.24 ± 186.03</td>
<td>0.381</td>
<td>0.364</td>
<td>0.780</td>
<td>0.546</td>
</tr>
<tr>
<td>Hip Peak 5 –15%</td>
<td>-28.95 ± 11.98</td>
<td>-202.53 ± 238.38</td>
<td>-31.22 ± 36.64</td>
<td>4.36 ± 11.47</td>
<td>0.872</td>
<td>0.872</td>
<td>0.029</td>
<td>0.019</td>
</tr>
<tr>
<td>Hip Peak 45 – 65%</td>
<td>95.70 ± 102.94</td>
<td>87.18 ± 90.51</td>
<td>85.28 ± 105.66</td>
<td>85.87 ± 113.67</td>
<td>0.891</td>
<td>0.891</td>
<td>0.014</td>
<td>0.021</td>
</tr>
<tr>
<td>Hip Peak 60-80%</td>
<td>146.55 ± 72.86</td>
<td>75.66 ± 85.42</td>
<td>139.57 ± 28.47</td>
<td>149.72 ± 72.19</td>
<td>0.880</td>
<td>0.945</td>
<td>0.517</td>
<td>0.826</td>
</tr>
</tbody>
</table>

Note: p denotes p – value; ES – effect size; b comparison between PIN and PUCK amputated; c comparison between PIN and PUCK non-amputated; d comparison between PIN amputated and PUCK non-amputated; e comparison between PIN non-amputated and PIN amputated; f comparison between PUCK amputated and PIN non-amputated; g comparison between PUCK amputated and PUCK non-amputated; * represents significant difference (p < 0.05).
Discussion

The purpose of this study was to identify differences in performance during functional tasks and treadmill walking in two different suspension systems (PIN and PUCK) in individuals with TTA. We used several methods to evaluate differences between suspension systems via a clinical measure (6MWT) and biomechanical outcomes.

Results from the 6MWT found while individuals on average in the PUCK suspension system walked (1473 ft) compared to the PIN suspension system (1405 ft). Individuals in the PUCK suspension system, on average, walked nearly 70 feet farther than while wearing the PIN suspension system. Although these differences were not statistically different, a moderate effect size was found (0.41), suggesting that with an increased sample size, we may find significant differences between groups in distance traveled. This greater walking distance is perceived as greater physical activity levels in these individuals (Deans, McFadyen, & Rowe, 2008) therefore a perceived better quality of life in the PUCK compared to the PIN suspension system. Even though, this difference was not significant, previous studies have reported in heart disease populations the minimal clinically important differences in the 6MWT ranges from ~78 to 148 ft. (du Bois et al., 2011; Gremeaux et al., 2011; Mathai, Puhan, Lam, & Wise, 2012). Based on these results, with an increased sample size this walking distance may reach clinical significance as suggested by the moderate effect size.

The current study results of overground 6MWT walking distance in the PIN (1405 ft) and PUCK (1473 ft) are shorter than previously reported distances in the literature (Lin & Bose, 2008; Lin et al., 2014). These studies relied on conducting the 6MWT using
words of encouragement over three different trials (~ 1786.7 ± 211.6 ft; 1817.6 ± 234.3 ft; 1870.1 ± 262.8 ft) (Lin & Bose, 2008). Furthermore, the data reported in the current study protocol only included individuals with traumatic amputation to complete one trial per session in each suspension system with no words of encouragement which may have resulted in decreased walking distances compared to previous studies (Lin & Bose, 2008). In contrast, this previous study did not record the suspension system or type of prosthesis used. Lin & Bose (2008) primary focus was to measure physiological variables with clinical tests and included individuals with dysvascular amputations. In general, these results partial did support our first hypothesis that while wearing the PUCK in which a few dependent variables in the PUCK, participants would show an increase in walking distance during overground walking as indicated by the moderate effect size. A greater sample size may show significant differences between suspension systems during overground walking as indicated by the moderate effect size.

Traditionally, the fit of the socket changes throughout the day, due to the loss of fluid in the residual limb, causing gait asymmetries in individuals with TTA (Beil et al., 2002; Board et al., 2001; Gerschutz et al., 2010; Goswami et al., 2003; Sanderson & Martin, 1997; Zachariah et al., 2004). The second and third hypotheses were aimed at answering whether the choice of socket suspension system would reduce these asymmetries during walking. Walking velocities (1.05 ± 0.44 m·s⁻¹) were controlled between conditions and testing days. No significant temporospatial asymmetries were found in the current study. However, a few measures approached significance resulting in moderate and large effect sizes, which will be discussed below. These measures may not have approached significance due to the individuals walking velocities (1.05 ± 0.44 m·s⁻¹
Studies have shown temporospatial inter-limb asymmetries continue to persist during walking in individuals walking at 1.12 m·s\(^{-1}\) to 1.18 m·s\(^{-1}\) (Genin et al., 2008; Hsu et al., 2006) compared to non-amputees walking at 1.27 m·s\(^{-1}\) to 1.46 m·s\(^{-1}\) (Bohannon, 1997). Individuals within previous study (Bohannon, 1997) that walked at greater velocities near non-amputee speed demonstrated greater inter-limb symmetry compared to individuals walking at slower velocities (Sanderson & Martin, 1997).

Although differences were seen between the amputated limbs of the suspension systems these dependent variables were not significant but showed a moderate and large effect size. Specifically, these asymmetrical angles during swing were noted between the amputated limbs of the suspension systems resulting in a \(~8^\circ\) knee flexion difference resulting in a large effect size (1.09). This large effect size suggest struetural differences including the neoprene sleeves maintaining negative pressure over the entire limb or the increased mass in the PUCK that influenced these individuals knee flexion in the amputated limb. For instance, the negative pressure over the entire limb creating a tight seal using the neoprene sleeve allowing for the individual to feel secure in the suspension system. However, this secure feeling with the neoprene sleeve could restrict the amputated limb knee flexion compared to the PIN system that does not use a neoprene sleeve.

Furthermore, this decreased knee flexion angle may also be attributed to mass differences between the systems. The mass of the PUCK suspension system (2.31 kg) was significantly heavier compared to the PIN suspension system (1.99 kg) suggesting the greater inertia of the PUCK system may have led to an increase in resistance to movement compared to the PIN.
Studies have shown manipulating the inertial properties can result in increased knee kinematic asymmetries (Hillery, Wallace, McIlhaggerm, & Watson, 1997; Smith & Martin, 2013). Specifically, these inertial masses applied to the prosthesis was shown to increase knee flexion in swing in the amputated limb compared to the non-amputated limb (Hillery et al., 1997). PUCK suspension system appeared to reduce knee kinematic asymmetries through swing instead of increasing contrary to previous studies have found (Hillery et al., 1997; Smith & Martin, 2013). These decreased kinematic asymmetries in the PUCK may contribute more to the neoprene sleeve than the increased mass which led to reduced kinematic asymmetries. However, this increased mass in the PUCK may contribute to the increased temporospatial asymmetries seen in the PUCK compared to the PIN.

The current study demonstrated slightly smaller (non-significant) knee flexion during swing in the PIN amputated limb (63.91°) compared to the results of Gholizadeh et al. (2014b) (66.9°). These differences between the studies may be attributed to some variability seen between studies including the type of amputation and individual’s activity levels. Gholizadeh et al. (2014b) included individuals with traumatic and dysvascular TTA along with decreased mobility (K2 and K3) contributing to slower walking velocities (0.93 m·s⁻¹) compared to the current study protocol that only included traumatic amputation and higher activity levels (K3 and K4) resulting in faster average walking velocities (1.05 m·s⁻¹).

Furthermore, in the PUCK amputated limb, knee flexion during swing was slightly smaller (55.54°) compared to Xu et al. 2017 (63.4°) in an elevated vacuum suspension system. Differences seen between studies may be attributed to the following:
a variety of causes of amputation (trauma, vascular, and other causes), controlled walking speed between 1.20 to 1.40 m·s$^{-1}$, and design of the elevated vacuum suspension system used in Xu et al. (2017) compared to the current study. Elevated vacuum suspension systems are known to experience leakage via the external pump causing a spike in positive pressure resulting in fluctuation (Komolafe et al., 2013) compared to the internally house pump in the PUCK allowing for the residual limb to maintain continuous pressure.

Although no significant differences were seen between the PIN and PUCK amputated limb temporospatial dependent variables including stance time (as percent of gait), swing time, swing time (as percent of gait), were found to be with moderate effect sizes. Differences in limb mass may have contributed to these differences seen in stance time as the PUCK suspension system (2.31 kg) is significantly heavier than the PIN (1.99 kg) suspension system. Previous studies have shown increased mass on one limb led to significantly greater stance time on the unloaded limb (Skinner & Barrack, 1990; Smith & Martin, 2007). Skinner and Barrack (1990) found non-amputees with a loaded limb (1.82 kg) showed an increase of about 20 ms in stance time for the unloaded limb. Smith and Martin (2007), suggest stance and swing time asymmetries appear immediately in non-amputees as an increase of mass (1.95 kg) was attached to one limb. However, these asymmetries appeared immediately these asymmetries were complete within five minutes of exposure to the additional load, suggesting individuals in the PUCK would have been acclimated to the heavier mass before walking on the treadmill. The current study protocol has individuals in each suspension system for at least a week prior to data collection and 6 minutes on the treadmill.
Consistent with the literature, the non-amputated limb demonstrated greater temporospatial asymmetries compared to the amputated limb (Board et al., 2001; Gerschutz et al., 2010; Goswami et al., 2003; Sanderson & Martin, 1997). Inter-limb asymmetries appeared greater in the PUCK suspension system compared to the PIN suspension system. Specifically, PUCK amputated limb showed a shorter stance time (63.5%) compared to the PUCK non-amputated limb (67.40%) \((p = 0.076)\) with a large effect size (1.08) and PIN non-amputated limb (67.62%) \((p = 0.062)\) which approached significance with a large effect size (1.21). Increasing our sample size, we may be able to detect significant differences.

The current study demonstrated slightly greater (non-significant) stance and swing time (as a percentage of the gait cycle) in the PIN amputated limb (65.04%; 35.02%) compared to the results of Gholizadeh et al. (2014b) (61.7%; 33.2%). These differences between the studies may be attributed to some variability seen discussed above. The PUCK amputated limb stance time (as a percentage of the gait cycle) in the was greater (63.5%) compared to stance time reported while using other elevated vacuum suspension systems (62.7%) (Xu et al., 2017). Differences seen between studies may be attributed to the differences discussed above.

Stance time SI was approaching a significant difference between the PUCK (-6.06%) and PIN (-3.88%) suspension system \((p = 0.097)\) with a moderate effect size (0.40). These negative SI values indicate the individuals relied more on the non-amputated limb which is a commonly seen compensation method in individuals with TTA (Board et al., 2001; Gerschutz et al., 2010; Goswami et al., 2003; Sanderson & Martin, 1997). Based on these results, compensatory strategies persisted in both of the
suspension systems and do not appear to disappear with the internal vacuum in the
PUCK. Although the PUCK was trending towards significantly greater asymmetries,
previous studies suggest good symmetry in temporospatial variables in this population are
defined as ± 10% symmetry (Dingwell, Davis, & Frazier, 1996; Robinson, Herzog, &
Nigg, 1987). This definition of good symmetry within the literature would suggest that
time spent in stance, as a percent of the gait cycle, for the PUCK and PIN are comparable
to non-amputee gait.

Even though, swing time SI did not reach a significant difference between the
PUCK (10.83%) and PIN (7.77%) suspension system ($p = 0.180$), there was a moderate
effect size (0.42) between the suspension system in this variable. These positive SI
values indicate the individuals relied more on the amputated limb which is a commonly
seen compensation method in individuals with TTA (Gerschutz et al., 2010; Goswami et
al., 2003; Sanderson & Martin, 1997). Furthermore, these asymmetries appeared to
persist in both suspension systems regardless of structural design suggesting that other
factors may be influencing these asymmetries additionally. Even though these variables
are not perfectly symmetrical by definition these results are considered good symmetry
within the literature comparable to non-amputee gait (Dingwell et al., 1996; Robinson et
al., 1987).

A notable finding in the PUCK suspension system was that individuals appeared
to have a remarkably symmetrical stride length (SI: -0.13%). This may be due to the
reduced ROM at the knee during swing caused by the presence of the neoprene sleeve for
the PUCK condition. The neoprene sleeve encases the entire residual limb to allow this
area to experience an air tight seal within the suspension system, however, this may be
preventing the individual to flex their knee completely through swing. Further investigation may lead to statistical differences and further insight into functional performance with a greater sample size. These results partially support our hypothesis that while wearing the PUCK in which a few dependent variables in the PUCK, will show an increased inter-limb temporospatial symmetry during treadmill walking.

The PUCK knee ROM showed remarkable inter-limb symmetry (-1.38%) compared to the PIN (10.52%). Although this SI value is not 0%, indicated as a perfect symmetry, Forczek and Staszkiewicz (2012) found non-amputee normal range of kinematic gait symmetry is within 2 to 4% SI. Based on these results, the amputated limb during swing in the PUCK appeared to behave more like the PUCK non-amputated limb than in the limbs of the PIN suspension system, suggesting an increase in symmetry in the PUCK. The internal vacuum may have provided a secure fit within the socket giving the individual more spatial awareness to determine appropriate angles throughout the gait cycle to maintain inter-limb symmetry in the PUCK. In summary, our results did not statistically support the third hypothesis in which while wearing the PUCK, participants will show increased inter-limb knee and hip angle symmetry during treadmill walking, however the moderate effect size suggest these trends may reach significant difference with a larger sample size.

Knee kinematics are altered due to the individual compensating for the missing ankle joint compared to non-amputated individuals (Nolan & Lees, 2000; Powers et al., 1998; Sanderson & Martin, 1997). Altered knee kinematics consist of the following: decreased knee flexion and increased knee extension in the amputated limb during stance compared to the non-amputated limb (Isakov et al., 1996; Nolan & Lees, 2000; Powers et
al., 1998; Sanderson & Martin, 1997; Winter & Sienko, 1988). These trends were consistent with the findings of the PIN and PUCK suspension system in this study.

Decreased knee flexion angles were seen in the PUCK suspension system (amputated: 52.53° vs. non-amputated: 55.54°) compared to the PIN suspension system (amputated: 63.91° vs. non-amputated: 58.74°). Although no significant differences in knee flexion were found between limbs in each of the suspension systems, moderate effect sizes were found in the PUCK suspension system (0.43) and the PIN suspension system (0.66), suggesting asymmetries continue regardless of system, but appears to decrease in the PUCK suspension system.

Additionally, differences in knee flexion were found between suspension systems and limbs. Although not significant, a moderate effect (0.43) in knee flexion was seen between the PUCK amputated and the PIN non-amputated limb, suggesting more may be influencing these asymmetries beside the suspension system. Furthermore, significant differences in knee flexion were found between the PIN amputated and the PUCK non-amputated limb ($p = 0.033$) and a large effect size (1.14). Although no differences were found between the non-amputated limbs, decreased knee flexion in the PUCK non-amputated limb compared to the PIN non-amputated limb suggest the non-amputated limbs behaves differently in each suspension system and individual requires a great knee flexibility in the PIN due to a structural difference.

Knee flexion during swing in the PIN suspension system of the current study are consistent with previous findings measuring functional performance in PIN systems (Gholizadeh et al., 2014b). Gholizadeh et al. 2014b found the amputated limb (66.9°) demonstrated greater knee flexion compared to the non-amputated limb (52.5°) compared
to the current study PIN (amputated: 63.91° vs non-amputated: 58.74°). These differences between the studies may be attributed to some variability seen between studies as discussed above including the type of amputation and individual’s activity levels.

Knee flexion during swing in the PUCK suspension system were smaller (amputated: 52.53° vs. non-amputated: 55.54°) than those reported in elevated vacuum suspension system (amputated: 66.3° vs. non-amputated: 63.4°) (Xu et al., 2017). These decreased knee angles are attributed to the slower walking speed in trauma related amputation individuals averaging 1.05 m·s⁻¹ compared to Xu et al. (2017) at 1.30 m·s⁻¹ with dysvascular amputees and other causes of amputation. Differences exist between dysvascular-and trauma-related amputations in regard to capability, use of prosthesis, and activity levels due to nature of the vascular amputation (Amtmann et al., 2015; Desveaux et al., 2016; Tudor-Lock et al., 2011). Dysvascular amputee capabilities are decreased compared to trauma related amputations in which these individuals are classified as K1 and K2 with limited activity levels therefore resulting in less time spent using their prosthesis (11.3 ± 4.4 hours) (Amtmann et al., 2015). These differences may attribute to differences discussed above.

Previous studies have found no significant differences between limbs at the knee and hip in angular velocity in amputees (Sanderson & Martin, 1997). However, significant differences were seen at the hip angular velocity during early stance. Specifically, significant differences were seen between the amputated and non-amputated limbs of each suspension system, where the amputated limb had significantly greater hip flexion angular velocities than the non-amputated limbs during early stance (PUCK: $p = 0.019$; PIN: $p = 0.021$). Furthermore, these significant differences are supported by the
large effect sizes (PUCK: 1.3; PIN: 1.46). Differences were seen between limbs and suspension systems including the PIN amputated limb demonstrated a greater hip angular velocity compared to the PUCK non-amputated limb ($p=0.029$) with a large effect size (2.84). Additionally, the PUCK amputated limb showed a greater hip angular velocity compared to the PIN non-amputated limb ($p=0.014$) with a moderate effect size (0.54). Overall these differences in angular velocity are likely due to the nature of the prosthetic foot. Kinematic asymmetries have been suggested to be due to the missing ankle musculature and the limited motion of the prosthetic foot (Aruin et al., 1997; Nolan & Lees, 2000; Winter, 1983) requiring the hip to increase its angular velocity to allow for the amputated limb pass through swing safely.

Although the design of the prosthetic foot has improved over the years (Donn et al., 1989; Smith & Martin, 2007) to mimic the function of a biological foot by providing degrees of freedom similar to a physiologic ankle and to assist with ambulation. Most of these prosthetic feet are passive devices that are unable to replicate the non-amputated ankle rollover motion in consequence contributing to an asymmetrical gait cycle (Nolan & Lees, 2000). As a result, the amputated limb may experience a greater angular velocity due to the prosthetic design.

In conclusion, socket fit can change throughout the day due to the fluctuations in limb volume leading to improper fit between the residual limb and socket causing asymmetries. Although individuals with TTA experience asymmetrical gait patterns, these differences may be attributed to other factors including choice of prosthetic foot and prosthetic alignment (Pitkin, 1997). Although few studies have investigated the influence of socket suspension systems on the gait of individuals with TTA (Board et al.,
2001; Pitkin, 1997; Sanders et al., 2011; Xu et al., 2017) the literature suggest improving the connection of the socket to the residual limb may reduce inter-limb asymmetries in people with TTA (Board et al., 2001; Pitkin, 1997; Sanders et al., 2011; Xu et al., 2017). Based on the results of the current study, the choice of socket suspension system has little impact on individuals with TTA walking ability. Gait asymmetries continued to persist in both suspension systems and individuals reached good temporospatial and kinematic symmetry (similar to previous studies), suggesting the type of suspension system may not influence gait symmetry.

Further, most of the variables did not reach statistical significance some were trending towards significance in which a greater sample size may allow for more statistical power to resolve values that are trending towards significance. Some limitations of this study may have prevented these variable reach significance differences. This study was conducted on a sample size and this may have impact on the statically relevance of the results as many variables were close to reaching statistical significance. Only comparing the PIN to the PUCK suspension system has limited our knowledge of how this new system compares to older vacuum suspension systems. Overall the type of the suspension system may have some influence on individuals gait symmetry however, there are other factors that may be still influencing this population gait.
CHAPTER V

CONCLUSION

Conclusion

The hypothesis that while wearing the PUCK, participants would show an increase in walking distance during overground walking was partially supported. Indicated by the moderate effect size a larger sample size may show significant differences between the PIN and PUCK suspension system during overground walking. The initial proposition that participants while wearing the PUCK would result in increased inter-limb temporospatial and joint angles symmetry during treadmill walking was partially supported. The PUCK suspension system demonstrated an increase in symmetry in stance time, cycle time, and stride length. Although these variables did not reach perfect symmetry define as 0%, these variables are considered within good symmetry range in non-amputated populations. However, increasing asymmetry were seen in other temporospatial dependent variables in the PUCK suspension system, these variables are still in range of good symmetry as seen in non-amputated populations (Dingwell et al., 1996).

Furthermore, kinematic symmetry appeared to increase in the PUCK suspension system specifically at the knee during swing in which the limbs showed a close resemblance. Although, these variables were not perfectly symmetrical these variables reached good symmetry as indicated by ± 10% (Dingwell et al., 1996). Individual subjects showed increase inter-limb symmetry in the PUCK compared to the PIN.
suspension system. Even though the sample size was small, subjects walking an average above 1.05 m·s\(^{-1}\) indicated in increase symmetry in the PUCK compared to the PIN suspension system. Individual data in each subject can be seen in the appendix for further detail. One subject had a difficult time walking on the treadmill in which could have skewed the data averages causing this individual to become an outlier. Furthermore, this individual had the greatest length of experience as a TTA indicated by his time of amputation. Therefore, his experience should have allowed for consistent data, however this was his first time during this study walking on a treadmill. Therefore, his walking velocities were the slowest among the sample group and did not reflect his ability to walk overground in either suspension system.

Although the socket fit can change throughout the day due to fluctuations in the residual limb causing a poor fit between the residual limb and socket, the type of socket suspension system has little impact on individuals with TTA walking abilities. In conclusion, gait asymmetries continued to persist in both suspension systems and individuals reached good temporospatial and kinematic symmetry (similar to previous studies), suggesting the type of suspension system may not influence gait symmetry. However, the type of the suspension system may have some influence on individuals gait symmetry along with other factors that may be still influencing this population’s gait.

Future research should focus on evaluating physiological and kinetic measures between the suspension systems to corroborate and understand the differences observed in the temporospatial and joint angle data. Although the type of suspension system may still influence these individuals’ gait, there is very little literature available on this topic. The effects the suspension system has on individuals with amputation should be studied
more to enhance this research in this area, it would be beneficial to compare the PUCK to other suspension systems in transtibial and transfemoral amputees.
REFERENCES


https://doi.org/10.1016/j.apmr.2007.11.005
APPENDIX A

INSTITUTIONAL REVIEW BOARD DOCUMENT
Institutional Review Board

DATE: February 18, 2019

TO: Abbie Ferris, PhD
FROM: University of Northern Colorado (UNCO) IRB

PROJECT TITLE: [1014426-3] Comparison of Prosthetic Suspension Systems on Function and Satisfaction

SUBMISSION TYPE: Continuing Review/Progress Report

ACTION: APPROVED
APPROVAL DATE: February 18, 2019
EXPIRATION DATE: February 8, 2020
REVIEW TYPE: Expedited Review

Thank you for your submission of Continuing Review/Progress Report materials for this project. The University of Northern Colorado (UNCO) IRB has APPROVED your submission. All research must be conducted in accordance with this approved submission.

This submission has received Expedited Review based on applicable federal regulations.

Please remember that informed consent is a process beginning with a description of the project and insurance of participant understanding. Informed consent must continue throughout the project via a dialogue between the researcher and research participant. Federal regulations require that each participant receives a copy of the consent document.

Please note that any revision to previously approved materials must be approved by this committee prior to initiation. Please use the appropriate revision forms for this procedure.

All UNANTICIPATED PROBLEMS involving risks to subjects or others and SERIOUS and UNEXPECTED adverse events must be reported promptly to this office.

All NON-COMPLIANCE issues or COMPLAINTS regarding this project must be reported promptly to this office.
Based on the risks, this project requires continuing review by this committee on an annual basis.
Please use the appropriate forms for this procedure. Your documentation for continuing review must be received with sufficient time for review and continued approval before the expiration date of February 8, 2020.
Please note that all research records must be retained for a minimum of three years after the completion of the project.

If you have any questions, please contact Nicole Morse at 970-351-1910 or nicole.morse@unco.edu. Please include your project title and reference number in all correspondence with this committee.

This letter has been electronically signed in accordance with all applicable regulations, and a copy is retained within University of Northern Colorado (UNCO) IRB’s records.
APPENDIX B

INDIVIDUAL DATA FOR PIN AND PUCK SUSPENSION SYSTEM WHILE WALKING
Table 1A
Temporospatial data for PIN suspension system for each subject while walking. Data are shown as mean ± standard deviation.

<table>
<thead>
<tr>
<th>Temporospatial System</th>
<th>PIN_01</th>
<th>PIN_02</th>
<th>PIN_03</th>
<th>PIN_04</th>
<th>PIN_05</th>
<th>PIN_01</th>
<th>PIN_02</th>
<th>PIN_03</th>
<th>PIN_04</th>
<th>PIN_05</th>
</tr>
</thead>
<tbody>
<tr>
<td>Temporospatial Cycle</td>
<td>1.06 ± 0.01</td>
<td>0.99 ± 0.99</td>
<td>1.36 ± 0.10</td>
<td>1.00 ± 0.00</td>
<td>1.24 ± 0.04</td>
<td>1.08 ± 0.01</td>
<td>1 ± 0.02</td>
<td>1.42 ± 0.08</td>
<td>1.03 ± 0.00</td>
<td>1.28 ± 0.11</td>
</tr>
<tr>
<td>Temporospatial Stance</td>
<td>0.70 ± 0.02</td>
<td>0.60 ± 0.98</td>
<td>1.00 ± 0.09</td>
<td>0.65 ± 0.01</td>
<td>0.82 ± 0.03</td>
<td>22.31 ± 0.67</td>
<td>0.64 ± 0.02</td>
<td>0.90 ± 0.14</td>
<td>0.68 ± 0.01</td>
<td>0.91 ± 0.09</td>
</tr>
<tr>
<td>Temporospatial Swing</td>
<td>0.67 ± 0.01</td>
<td>0.59 ± 0.00</td>
<td>0.64 ± 0.01</td>
<td>0.65 ± 0.00</td>
<td>0.66 ± 0.00</td>
<td>0.67 ± 0.01</td>
<td>0.63 ± 0.01</td>
<td>0.74 ± 0.02</td>
<td>0.67 ± 0.01</td>
<td>0.71 ± 0.02</td>
</tr>
<tr>
<td>Temporospatial Swing</td>
<td>0.35 ± 0.01</td>
<td>0.40 ± 0.00</td>
<td>0.35 ± 0.12</td>
<td>0.35 ± 0.01</td>
<td>0.42 ± 0.03</td>
<td>0.35 ± 0.02</td>
<td>0.36 ± 0.0</td>
<td>0.51 ± 0.13</td>
<td>0.32 ± 0.02</td>
<td>0.36 ± 0.02</td>
</tr>
<tr>
<td>Temporospatial Swing</td>
<td>0.22 ± 0.02</td>
<td>0.40 ± 0.00</td>
<td>0.25 ± 0.06</td>
<td>0.35 ± 0.00</td>
<td>0.33 ± 0.05</td>
<td>0.33 ± 0.01</td>
<td>0.36 ± 0.00</td>
<td>0.38 ± 0.11</td>
<td>0.33 ± 0.01</td>
<td>0.29 ± 0.05</td>
</tr>
<tr>
<td>Temporospatial Stance</td>
<td>1.18 ± 0.03</td>
<td>1.6 ± 1.6</td>
<td>0.68 ± 0.08</td>
<td>1.12 ± 0.00</td>
<td>0.92 ± 0.81</td>
<td>1.17 ± 0.03</td>
<td>1.6 ± 0.03</td>
<td>0.71 ± 0.07</td>
<td>1.12 ± 0.01</td>
<td>0.97 ± 0.11</td>
</tr>
</tbody>
</table>

Table 2A
Temporospatial data between PIN and PUCK suspension system for each subject while walking. Data are shown as mean ± standard deviation.

<table>
<thead>
<tr>
<th>Temporospatial Comparsion</th>
<th>PIN_01</th>
<th>PIN_02</th>
<th>PIN_03</th>
<th>PIN_04</th>
<th>PIN_05</th>
<th>PUCK_01</th>
<th>PUCK_02</th>
<th>PUCK_03</th>
<th>PUCK_04</th>
<th>PUCK_05</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stride Width (m)</td>
<td>0.16 ± 0.00</td>
<td>0.12 ± 0.00</td>
<td>0.17 ± 0.17</td>
<td>0.13 ± 0.02</td>
<td>0.21 ± 0.05</td>
<td>0.13 ± 0.04</td>
<td>0.12 ± 0.01</td>
<td>0.17 ± 0.01</td>
<td>0.11 ± 0.12</td>
<td>0.21 ± 0.00</td>
</tr>
<tr>
<td>NA DLST (s)</td>
<td>0.19 ± 0.02</td>
<td>0.13 ± 0.02</td>
<td>0.19 ± 0.13</td>
<td>0.17 ± 0.01</td>
<td>0.25 ± 0.06</td>
<td>0.18 ± 0.02</td>
<td>0.11 ± 0.01</td>
<td>0.21 ± 0.03</td>
<td>0.16 ± 0.01</td>
<td>0.2 ± 0.00</td>
</tr>
<tr>
<td>AMP DLST (s)</td>
<td>0.16 ± 0.02</td>
<td>0.10 ± 0.01</td>
<td>0.31 ± 0.07</td>
<td>0.16 ± 0.00</td>
<td>0.18 ± 0.03</td>
<td>0.16 ± 0.01</td>
<td>0.10 ± 0.01</td>
<td>0.31 ± 0.02</td>
<td>0.14 ± 0.01</td>
<td>0.18 ± 0.02</td>
</tr>
</tbody>
</table>
Table 3A

Kinematic data in the PIN suspension system for each subject while walking. Data are shown as mean ± standard deviation.

<table>
<thead>
<tr>
<th>Angle (deg)</th>
<th>PIN_01</th>
<th>PIN_02</th>
<th>PIN_03</th>
<th>PIN_04</th>
<th>PIN_05</th>
<th>PIN_01</th>
<th>PIN_02</th>
<th>PIN_03</th>
<th>PIN_04</th>
<th>PIN_05</th>
</tr>
</thead>
<tbody>
<tr>
<td>Knee Peak</td>
<td>2.27 ± 1.2</td>
<td>-0.77 ± 0.5</td>
<td>19.03 ± 2.8</td>
<td>3.44 ± 0.6</td>
<td>1.11 ± 2.8</td>
<td>13.63 ± 0.9</td>
<td>1.28 ± 1.7</td>
<td>10.30 ± 2.5</td>
<td>3.69 ± 0.8</td>
<td>11.23 ± 1.9</td>
</tr>
<tr>
<td>Knee Peak</td>
<td>18.8 ± 0.7</td>
<td>9.45 ± 0.9</td>
<td>26.78 ± 0.9</td>
<td>9.50 ± 0.7</td>
<td>2.25 ± 2.8</td>
<td>25.28 ± 0.3</td>
<td>18.37 ± 0.3</td>
<td>14.59 ± 0.5</td>
<td>17.14 ± 0.9</td>
<td>12.37 ± 1.3</td>
</tr>
<tr>
<td>Knee Peak</td>
<td>11.6 ± 1.1</td>
<td>1.94 ± 0.4</td>
<td>29.27 ± 0.8</td>
<td>6.59 ± 1.9</td>
<td>-0.58 ± 3.6</td>
<td>16.57 ± 0.5</td>
<td>-1.87 ± 0.4</td>
<td>13.30 ± 0.9</td>
<td>-0.79 ± 0.6</td>
<td>4.61 ± 0.6</td>
</tr>
<tr>
<td>Knee Peak</td>
<td>67.9 ± 0.4</td>
<td>70.10 ± 0.3</td>
<td>65.94 ± 1.0</td>
<td>72.16 ± 0.5</td>
<td>52.92 ± 1.9</td>
<td>69.24 ± 0.0</td>
<td>58.74 ± 0.3</td>
<td>56.67 ± 2.0</td>
<td>58.96 ± 1.0</td>
<td>54.00 ± 1.6</td>
</tr>
<tr>
<td>Knee ROM</td>
<td>27.17 ± 8.5</td>
<td>71.44 ± 24.5</td>
<td>37.59 ± 11.9</td>
<td>69.37 ± 23.0</td>
<td>54.43 ± 18.6</td>
<td>56.53 ± 17.4</td>
<td>60.64 ± 19.3</td>
<td>50.32 ± 17.4</td>
<td>59.75 ± 19.1</td>
<td>49.40 ± 15.7</td>
</tr>
<tr>
<td>Hip Peak</td>
<td>21.4 ± 0.1</td>
<td>29.36 ± 1.1</td>
<td>42.17 ± 2.3</td>
<td>17.88 ± 0.4</td>
<td>14.62 ± 4.0</td>
<td>31.75 ± 0.3</td>
<td>27.30 ± 0.5</td>
<td>31.66 ± 0.5</td>
<td>18.75 ± 0.3</td>
<td>15.17 ± 0.7</td>
</tr>
<tr>
<td>Hip Peak</td>
<td>-8.82 ± 0.9</td>
<td>-7.14 ± 1.3</td>
<td>20.08 ± 0.6</td>
<td>-19.43 ± 0.1</td>
<td>-15.23 ± 2.5</td>
<td>-5.97 ± 0.3</td>
<td>-15.28 ± 0.6</td>
<td>13.28 ± 0.6</td>
<td>-17.47 ± 0.5</td>
<td>-6.71 ± 0.7</td>
</tr>
<tr>
<td>Hip Peak</td>
<td>28.5 ± 0.7</td>
<td>34.46 ± 0.2</td>
<td>46.40 ± 0.8</td>
<td>25.91 ± 0.5</td>
<td>18.82 ± 0.9</td>
<td>31.60 ± 0.3</td>
<td>28.69 ± 0.6</td>
<td>35.84 ± 0.5</td>
<td>22.34 ± 1.0</td>
<td>15.41 ± 1.3</td>
</tr>
<tr>
<td>Hip ROM</td>
<td>37.34 ± 12.4</td>
<td>43.29 ± 14.7</td>
<td>19.50 ± 5.7</td>
<td>45.34 ± 14.9</td>
<td>34.59 ± 12.0</td>
<td>39.28 ± 13.7</td>
<td>44.20 ± 16.0</td>
<td>30.78 ± 9.9</td>
<td>39.84 ± 13.4</td>
<td>24.16 ± 8.5</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Angular Velocity (deg/s)</th>
<th>PIN_01</th>
<th>PIN_02</th>
<th>PIN_03</th>
<th>PIN_04</th>
<th>PIN_05</th>
<th>PIN_01</th>
<th>PIN_02</th>
<th>PIN_03</th>
<th>PIN_04</th>
<th>PIN_05</th>
</tr>
</thead>
<tbody>
<tr>
<td>Knee Peak</td>
<td>-65.39 ± 19.1</td>
<td>63.54 ± 17.8</td>
<td>-35.14 ± 33.6</td>
<td>-62.27 ± 12.6</td>
<td>-16.75 ± 29.8</td>
<td>51.95 ± 20.3</td>
<td>99.02 ± 17.0</td>
<td>-31.29 ± 57.1</td>
<td>43.42 ± 24.9</td>
<td>-8.09 ± 19.6</td>
</tr>
<tr>
<td>Knee Peak</td>
<td>152.55 ± 7.5</td>
<td>91.49 ± 3.8</td>
<td>114.85 ± 17.9</td>
<td>-24.12 ± 16.0</td>
<td>20.67 ± 11.8</td>
<td>122.61 ± 15.5</td>
<td>159.06 ± 19.34</td>
<td>52.69 ± 32.2</td>
<td>165.38 ± 12.6</td>
<td>12.24 ± 21.1</td>
</tr>
<tr>
<td>Knee Peak</td>
<td>45.90 ± 27.7</td>
<td>371.44 ± 21.5</td>
<td>18.93 ± 34.3</td>
<td>-54.34 ± 27.2</td>
<td>35.16 ± 18.2</td>
<td>42.76 ± 28.8</td>
<td>70.76 ± 17.0</td>
<td>3.62 ± 2.5</td>
<td>261.82 ± 33.5</td>
<td>15.59 ± 12.7</td>
</tr>
<tr>
<td>Knee Peak</td>
<td>395.82 ± 17.0</td>
<td>467.10 ± 3.8</td>
<td>72.41 ± 69.7</td>
<td>152.41 ± 17.9</td>
<td>263.53 ± 47.6</td>
<td>329.44 ± 15.3</td>
<td>375.53 ± 7.2</td>
<td>51.45 ± 14.3</td>
<td>349.37 ± 19.2</td>
<td>88.38 ± 43.5</td>
</tr>
<tr>
<td>Knee Peak</td>
<td>-368.34 ± 17.0</td>
<td>-414.16 ± 4.0</td>
<td>10.96 ± 50.5</td>
<td>39.31 ± 16.4</td>
<td>-111.04 ± 70.8</td>
<td>-260.87 ± 189</td>
<td>-434.22 ± 18.1</td>
<td>55.88 ± 14.0</td>
<td>-413.62 ± 5.7</td>
<td>40.15 ± 51.8</td>
</tr>
<tr>
<td>Hip Peak</td>
<td>-30.13 ± 6.1</td>
<td>-37.55 ± 2.3</td>
<td>-8.05 ± 25.9</td>
<td>-34.50 ± 4.3</td>
<td>-34.54 ± 30.0</td>
<td>7.82 ± 8.6</td>
<td>-0.59 ± 18.1</td>
<td>8.77 ± 10.9</td>
<td>33.55 ± 11.2</td>
<td>-13.92 ± 12.5</td>
</tr>
<tr>
<td>Hip Peak</td>
<td>128.81 ± 51.5</td>
<td>133.48 ± 81.4</td>
<td>-13.05 ± 10.1</td>
<td>231.51 ± 5.9</td>
<td>-2.26 ± 15.9</td>
<td>75.60 ± 20.1</td>
<td>166.48 ± 7.6</td>
<td>-12.25 ± 1.5</td>
<td>155.50 ± 14.3</td>
<td>-7.02 ± 3.1</td>
</tr>
<tr>
<td>Hip Peak</td>
<td>195.92 ± 6.1</td>
<td>150.62 ± 19.2</td>
<td>-40.89 ± 72.2</td>
<td>230.64 ± 6.5</td>
<td>117.68 ± 19.5</td>
<td>172.65 ± 3.2</td>
<td>215.27 ± 11.4</td>
<td>-6.99 ± 14.3</td>
<td>144.05 ± 6.1</td>
<td>22.03 ± 30.7</td>
</tr>
</tbody>
</table>
Table 4A

Functional: Six-Minute Walk Test

<table>
<thead>
<tr>
<th>Overground Walking</th>
<th>PIN_01</th>
<th>PIN_02</th>
<th>PIN_03</th>
<th>PIN_04</th>
<th>PIN_05</th>
<th>PUCK_01</th>
<th>PUCK_02</th>
<th>PUCK_03</th>
<th>PUCK_04</th>
<th>PUCK_05</th>
</tr>
</thead>
<tbody>
<tr>
<td>6MWT (ft)</td>
<td>1150</td>
<td>1460</td>
<td>1200</td>
<td>1595</td>
<td>1620</td>
<td>1390</td>
<td>1365</td>
<td>1360</td>
<td>1610</td>
<td>1640</td>
</tr>
</tbody>
</table>

Table 5A

Symmetry Index data in PIN and PUCK suspension system for each subject while walking. Data are shown as mean ± standard deviation.

<table>
<thead>
<tr>
<th>SYMMETRY INDEX OF ALL SUBJECTS</th>
<th>PIN_01</th>
<th>PIN_02</th>
<th>PIN_03</th>
<th>PIN_04</th>
<th>PIN_05</th>
<th>PUCK_01</th>
<th>PUCK_02</th>
<th>PUCK_03</th>
<th>PUCK_04</th>
<th>PUCK_05</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cycle time (s)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>-0.14 ± 1.1</td>
<td>-0.25 ± 1.0</td>
<td>1.13 ± 7.8</td>
<td>-0.34 ± 0.0</td>
<td>0.08 ± 5.8</td>
<td>-0.11 ± 1.8</td>
<td>0.08 ± 0.0</td>
<td>-0.56 ± 2.0</td>
<td>0 ± 1.0</td>
<td>-0.14 ± 1.3</td>
<td></td>
</tr>
<tr>
<td>Stance time (s)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>-0.25 ± 3.6</td>
<td>-1.9 ± 1.7</td>
<td>-2.61 ± 3.3</td>
<td>-1.02 ± 1.7</td>
<td>0.43 ± 7.9</td>
<td>0.74 ± 2.1</td>
<td>2.23 ± 1.7</td>
<td>0 ± 2.4</td>
<td>0.28 ± 1.8</td>
<td>1.23 ± 2.8</td>
<td></td>
</tr>
<tr>
<td>Swing time (s)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>0 ± 4.9</td>
<td>2.44 ± 1.5</td>
<td>9.44 ± 18.9</td>
<td>1.91 ± 3.7</td>
<td>12.59 ± 2.6</td>
<td>-1.76 ± 9.6</td>
<td>-3.25 ± 1.3</td>
<td>-1.97 ± 13.0</td>
<td>-0.53 ± 1.8</td>
<td>-12.16 ± 3.4</td>
<td></td>
</tr>
<tr>
<td>DLST (s)</td>
<td>4.29 ± 10.2</td>
<td>6.36 ± 13.1</td>
<td>-12.62 ± 16.0</td>
<td>1.50 ± 5.9</td>
<td>5.85 ± 23.9</td>
<td>3.35 ± 11.7</td>
<td>0.95 ± 0.6</td>
<td>0 ± 45.7</td>
<td>2.81 ± 3.5</td>
<td>3.05 ± 9.3</td>
</tr>
</tbody>
</table>
Table 6A

*Temporospatial data for PIN suspension system for each subject while walking. Data are shown as mean ± standard deviation.*

<table>
<thead>
<tr>
<th>Temporospatial</th>
<th>AMPUTATED</th>
<th>NON-AMPUTATED</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cycle Time (s)</td>
<td>PUCK_01</td>
<td>PUCK_02</td>
</tr>
<tr>
<td></td>
<td>1.08 ± 0.03</td>
<td>0.99 ± 0.01</td>
</tr>
<tr>
<td>Time (s) Stance</td>
<td>0.70 ± 0.03</td>
<td>0.62 ± 0.01</td>
</tr>
<tr>
<td>Time (%) Stance</td>
<td>0.64 ± 0.13</td>
<td>0.58 ± 0.00</td>
</tr>
<tr>
<td>Swing Time (s)</td>
<td>0.40 ± 0.02</td>
<td>0.37 ± 0.05</td>
</tr>
<tr>
<td>Swing Time (%)</td>
<td>0.35 ± 0.04</td>
<td>0.41 ± 0.01</td>
</tr>
<tr>
<td>Stride Length (m)</td>
<td>1.18 ± 0.04</td>
<td>1.6 ± 0.02</td>
</tr>
</tbody>
</table>
Table 7A

*Kinematic data in the PIN suspension system for each subject while walking. Data are shown as mean ± standard deviation.*

**PUCK SUSPENSION SYSTEM KINEMATIC**

<table>
<thead>
<tr>
<th>Angle (deg)</th>
<th>AMPUTATED</th>
<th>NON-AMPUTATED</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>PUCK_01</td>
<td>PUCK_02</td>
</tr>
<tr>
<td>Knee Peak</td>
<td>0 - 5%</td>
<td>6.46 ± 0.1</td>
</tr>
<tr>
<td>Knee Peak</td>
<td>10 - 20%</td>
<td>22.31 ± 0.7</td>
</tr>
<tr>
<td>Knee Peak</td>
<td>30-50%</td>
<td>16.69 ± 0.3</td>
</tr>
<tr>
<td>Knee Peak</td>
<td>60-80%</td>
<td>59.18 ± 2.0</td>
</tr>
<tr>
<td>Knee ROM</td>
<td>0-100%</td>
<td>52.88 ± 15.6</td>
</tr>
<tr>
<td>Hip Peak</td>
<td>5 - 15%</td>
<td>20.84 ± 1.2</td>
</tr>
<tr>
<td>Hip Peak</td>
<td>45 - 65%</td>
<td>-9.93 ± 1.5</td>
</tr>
<tr>
<td>Hip Peak</td>
<td>70-90%</td>
<td>25.56 ± 2.3</td>
</tr>
<tr>
<td>Hip ROM</td>
<td>0-100%</td>
<td>36.85 ± 12.3</td>
</tr>
<tr>
<td>Angular Velocity (deg·s⁻¹)</td>
<td>AMPUTATED</td>
<td>NON-AMPUTATED</td>
</tr>
<tr>
<td>Knee Peak</td>
<td>0 - 5%</td>
<td>2.57 ± 27.6</td>
</tr>
<tr>
<td>Knee Peak</td>
<td>10 - 20%</td>
<td>163.91 ± 9.6</td>
</tr>
<tr>
<td>Knee Peak</td>
<td>30-50%</td>
<td>41.43 ± 13.5</td>
</tr>
<tr>
<td>Knee Peak</td>
<td>60-80%</td>
<td>283.84 ± 9.8</td>
</tr>
<tr>
<td>Knee Peak</td>
<td>80-90%</td>
<td>-294.43 ± 17.8</td>
</tr>
<tr>
<td>Hip Peak</td>
<td>5 - 15%</td>
<td>-30.55 ± 5.6</td>
</tr>
<tr>
<td>Hip Peak</td>
<td>45 - 65%</td>
<td>-22.81 ± 9.2</td>
</tr>
<tr>
<td>Hip Peak</td>
<td>60-80%</td>
<td>141.75 ± 11.0</td>
</tr>
</tbody>
</table>