Biomechanical Adaptations to Lower Limb Amputation during Functional Mobility Tasks

by

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Dedication

To my parents, for their undying support.

Your sacrifices to provide a better life for your children continue to motivate me to achieve my goals and to be the best version of myself.

Acknowledgements

There are so many people to thank for helping me get to this point in my career. None of my this would have been possible without the tremendous support and encouragement I received from the amazing mentors, friends, and family throughout my journey.

Preface

Chapters 2-5 have been written as separate manuscripts for publication. As such there may be some repetition of content, particularly the methods section, between the chapters.

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Abstract

People with a unilateral transtibial amputation (TTA) complete functional tasks asymmetrically, using compensatory strategies to accommodate for the lost ankle muscle function. These strategies may contribute the greater intact limb joint pain, low-back pain, and greater risk of falling commonly reported in this population. Prosthetists attempt to reduce asymmetries during the prosthetic alignment process. However, this process, which focuses on straight-line walking, may not capture the effect of prosthetic alignment on other functional tasks. The purpose of this dissertation was to determine how people with TTA maintain dynamic balance during turning and seat transfers and to quantify the effects of prosthetic alignment during seat transfers.

The first aim was to determine how balance regulation during turning is affected by the side the prosthesis is on and quantify how people with TTA maintain dynamic balance during a 90-degree turn. Participants with TTA had greater range of whole-body angular momentum when turning with the prosthesis on the inside compared to outside of the turn. There were altered head/trunk and legs interactions between turns and groups. The observed differences when turning with the prosthesis on the inside of a turn may suggest people with TTA have a greater risk of balance loss during turning.

The second aim was to quantify the effect of prosthetic alignment on dynamic balance during functional tasks**.** We compared the range and number of adjustments of whole-body angular momentum during walking, sit-to-stand, stand-to-sit, sit-to-walk, and walk-to-sit between different alignments. Sit-to-stand was the only task where alignment significantly affected angular

momentum, although differences in magnitudes were small. Participants with TTA had less balance control compared to non-amputees, across alignments. These results suggest that acute changes in prosthetic alignment likely do not affect balance control during seat transfers.

The third aim was to determine the effects of anterior-posterior alignment shifts on movement strategies during sit-to-stand. We compared 3D ground reaction force impulses, sagittal-plane knee moments, anterior/posterior center of pressure position, and 3D trunk range of motion between alignments. The posterior alignment reduced braking impulse asymmetry and axial trunk range of motion compared to other alignments. These results suggest that prosthetic alignment may affect the movement strategies used during sit-to-stand which may have implications for asymmetric and altered movement patterns found in people with TTA.

The fourth aim was to determine the effect of prosthetic alignment on hip and low-back joint contact forces during sit-to-stand in people with a unilateral transtibial amputation. Using a musculoskeletal simulation framework, there were no differences in hip and L4-L5 joint contact forces between alignments. Participants with TTA had a greater peak hip joint contact force on the intact side hip compared to the amputated side across all alignments. This result may have important implications as greater cumulative intact hip loading throughout daily life may increase the risk of hip joint pain and degeneration in people with TTA.

Together, these studies support the idea that even highly functional individuals with a lower limb amputation have decreased balance control and altered joint loading across a range of functional tasks. Results from these studies also suggest that people with TTA develop compensatory strategies in response to acute changes in prosthetic alignment do not affect balance or joint loading during seat transfers. Future work should explore whether these findings extend to long-term changes in alignment or to lower functioning individuals.

CHAPTER 1. Introduction

Lower limb loss is a debilitating condition that affects functional mobility and independent living which may lead to a decreased quality of life (Deans et al., 2008; Gallagher and Maclachlan, 2004; Harness and Pinzur, 2001; Sinha and Van Den Heuvel, 2011). Below-knee limb loss, or transtibial amputation (TTA), is the most common major lower limb amputation affecting 70% of the population with lower limb loss in the U.S (Ziegler-Graham et al., 2008). In addition, the number of people with TTA is estimated to increase to 1 million by 2050, doubling the number of people with TTA in 2005 (Ziegler-Graham et al., 2008). Due to the loss of an ankle joint and its associated musculature, people with TTA compensate by using proximal segments and relying on the intact limb to perform activities of daily living. These compensations alter movement strategies which may lead to high rates of secondary health conditions such as osteoarthritis (Morgenroth et al., 2012; Norvell et al., 2005) and low back pain (Ehde et al., 2001; Highsmith et al., 2019; Kulkarni et al., 2005) and an increased risk of falls (Miller et al., 2001b; Yu et al., 2010). As a consequence of these conditions, people with TTA reduce their activity levels (Bussmann et al., 2008), leading to immobility and institutionalization due to loss of independence (Remes et al., 2009; Stineman et al., 2009).

Specific compensations can vary across people with TTA and can be affected by decreased balance confidence. While people with TTA typically exhibit an asymmetrical gait pattern likely required as a consequence of the inability to control the ankle, lack of confidence in the prosthesis may also contribute to this asymmetry. For example, it is common for people with TTA to walk with a shorter step length on the intact limb (Isakov et al., 2000; Nolan et al., 2003). This asymmetric pattern is often attributed to a lack of confidence in the prosthesis to be able to provide body support when full body weight is over the amputated limb. Similarly, people with TTA spend more time in double limb support compared to non-amputees (Wilken and Marin, 2009) likely to improve stability. In addition, people with TTA increase trunk movement (Krajbich, 2016; Yoder et al., 2015) to avoid fully loading the amputated limb which may have important implications for the increased rates of low back pain in people with TTA (Kulkarni et al., 2005). Furthermore, the lack of confidence in loading their prosthesis results in the increased reliance on the intact limb which may increase the risk of developing knee osteoarthritis (Morgenroth et al., 2011; Norvell et al., 2005).

Asymmetric gait can also be attributed to the properties of the prosthesis. As prosthesis discomfort can also be a reason for people with TTA to avoid loading their amputated limb, prosthetists attempt to choose the prosthesis components (e.g., socket, pylon, foot, suspension mechanism; Figure 1.1) with appropriate mechanical properties (e.g., stiffness) that maximize comfort while minimizing asymmetry (Krajbich, 2016). In addition, the prosthetist can change the orientation of the prosthetic foot with respect to the socket, or 'alignment' (Figure 1.2). Proper orientation ensures that forces from the ground are directed appropriately for load transfer and comfort between the socket and residual limb. The alignment process is typically iterative and involves active patient feedback (Boone et al., 2012; Mizrahi et al., 1986; Zahedi, 1986). As such, its success depends on the experience of the prosthetist and the ability of the patient to detect when the prosthesis is not being loaded properly (Boone et al., 2012; Mizrahi et al., 1986; Zahedi, 1986). In addition, the alignment process only focuses on standing and straight-line walking. However, it is currently unclear whether an ‘optimal’ alignment for standing, or walking is also appropriate for other common tasks such as seat transfers and changing direction.

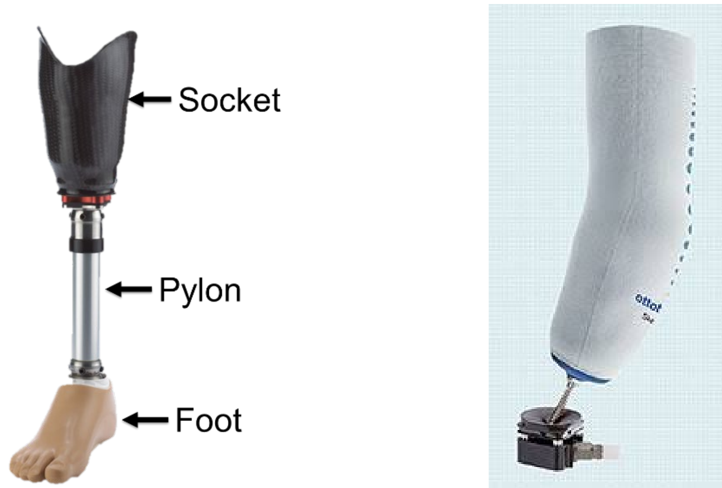

Figure 1.1. A) Prosthesis with socket, pylon, and foot. **B)** Pin locking liner and mechanism, an example of a socket suspension mechanism (https://opedge.com/Articles/ViewArticle/NEWS_2013-12-01_17)

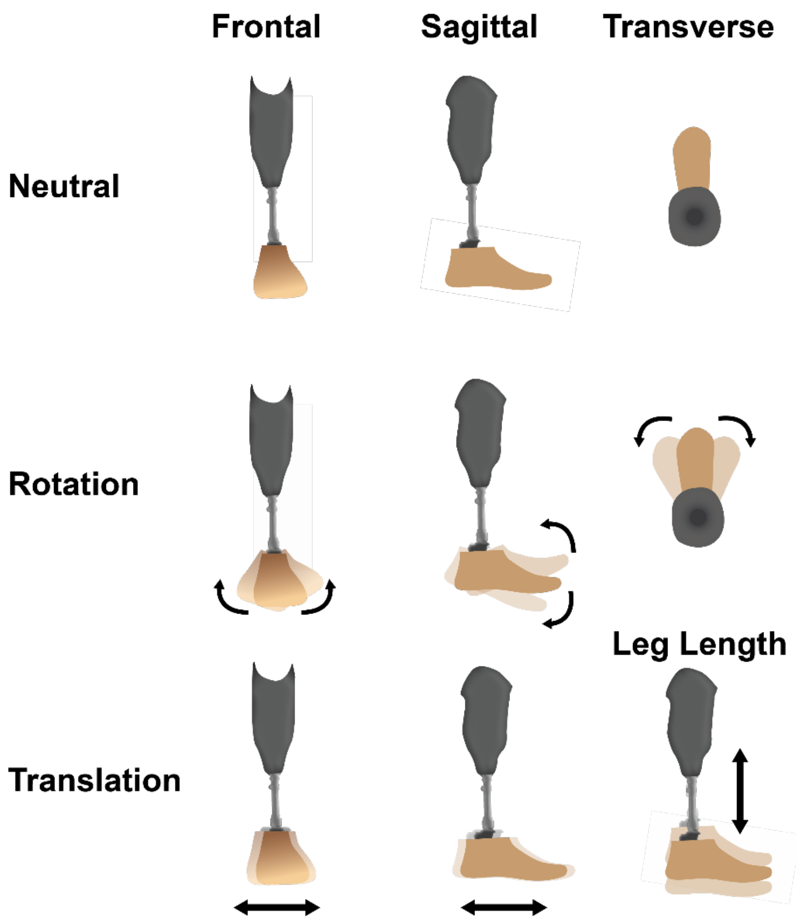

Figure 1.2. Prosthetic alignment description for transtibial prostheses

Functional mobility throughout daily life generally depends on the ability to change or maintain body position, carrying, moving and handling objects, and moving around within the environment (Radhakrishnan et al., 2017; WHO, 2001). However, prosthetists and a majority of the scientific literature primarily focus on people with TTA walking in a straight line. It is important to understand how people with TTA perform other common tasks that are important for functional mobility. Seat transfers (Bussmann et al., 2004; Bussmann et al., 2008) and turning (Glaister et al., 2007) are tasks that are common during daily life and are necessary to move within and between environments and activities of daily living. Performing turns and seat transfers may be more challenging for people with TTA as they require changing body position with greater momentum control (Nolasco et al., 2019; Pai and Rogers, 1991) to avoid losing balance, compared to walking in a straight line.

The remainder of this chapter will cover the effect of amputation on the functional mobility tasks of walking, turning, and seat transfers, balance in people with TTA, musculoskeletal modeling and simulations applied to functional tasks in people with TTA, the effects of alignment on functional tasks, and the purpose and aims of this dissertation.

1.1 Amputation Effects on Functional Mobility Tasks

The biological ankle muscles provide up to 80% of the mechanical power used during walking (Soo and Donelan, 2010; Winter, 1983). Ankle muscle forces are important for body support, forward progression, and swing initiation (Liu et al., 2008; Neptune et al., 2001). While energy storing and returning prostheses can sufficiently replace the contribution of plantarflexors to vertical body support (Silverman and Neptune, 2012; Zmitrewicz et al., 2007), the loss of a functional ankle and the lack of control over a prosthesis is still associated with compensatory strategies to be able to walk. These compensatory strategies are characterized by between-limb asymmetries in muscle activity, temporal-spatial parameters, kinematics, and kinetics.

Throughout the gait cycle, people with TTA make adjustments that are asymmetric between limbs likely to avoid loading their amputated limb and compensate for the lack of ankle muscle function. For example, people with TTA increase double support time (Wilken and Marin, 2009), decrease intact limb step length (Isakov et al., 2000; Nolan et al., 2003), and spend less time on the amputated limb (Isakov et al., 2000; Nolan et al., 2003). The increased reliance on the intact limb causes greater vertical ground reaction forces (GRFs) compared to the amputated limb (Arya et al., 1995; Nolan et al., 2003; Sanderson and Martin, 1997). At the knee on the amputated side, people with TTA walk with lower knee abduction moment compared to the intact limb (Molina-Rueda et al., 2014) and non-amputees (Beyaert et al., 2008; Sanderson and Martin, 1997; Silverman et al., 2008). This compensation may be a due to reduced confidence in motion control of the knee (Sanderson and Martin, 1997) which results in the need for knee stability. As knee motion primarily occurs in the sagittal plane, knee stability during single limb stance likely comes from the greater knee extensor and flexor muscle activity (Fey et al., 2010; Seyedali et al., 2012).

Altered trunk strategies are also common in people with TTA as they have less confidence in loading their body weight on the amputated limb. This lack of confidence or reluctance to load the amputated limb and/or need to control frontal plane forces lead to greater trunk motion toward the amputated side during the amputated limb single stance (Hendershot and Wolf, 2014; Molina-Rueda et al., 2014; Molina Rueda et al., 2013; Yoder et al., 2015). Concurrent with greater trunk lean toward the amputated limb, people with TTA have greater antagonistic oblique muscle forces on the intact side (Yoder et al., 2015) likely for trunk control. In non-amputees, the hip abductors counteract the effect of gravity by rotating the body towards the stance limb (Neptune and McGowan, 2016). People with TTA have reduced frontal plane hip abduction moment in the amputated limb compared to the intact limb (Molina Rueda et al., 2013; Royer and Wasilewski, 2006) and non-amputees (Molina Rueda et al., 2013). Thus, in order to rotate the body towards the amputated limb people with TTA increase their trunk motion likely to compensate for the reduced hip abductor moment. Moreover, people with TTA have greater eccentric hip abductor activity to rotate the trunk from the amputated side to the intact side (Sadeghi et al., 2001).

While energy storing and returning prostheses contribute to vertical body support, they do not generate the energy provided by the plantarflexors for forward progression and swing initiation (Silverman and Neptune, 2012; Zmitrewicz et al., 2007). As such, the amputated limb has lower vertical and anterior-posterior GRFs compared to the intact limb and to non-amputees (Nolan et al., 2003; Sanderson and Martin, 1997). In addition, the prosthesis decelerates the trunk in the horizontal direction compared to plantarflexor muscles (Zmitrewicz et al., 2007), thus requiring other muscles to generate forward propulsion (Silverman and Neptune, 2012; Zmitrewicz et al., 2007). In particular, amputated limb hip extensors deliver energy to the trunk during early stance while hip flexors redistribute energy to the trunk during late stance and pre-swing for forward progression (Silverman and Neptune, 2012; Zmitrewicz et al., 2007). Furthermore, people with TTA also have altered muscle activity in the intact limb. Specifically, the soleus, rectus femoris, and gluteus maximus deliver more energy to the trunk during the first half of stance (Silverman and Neptune, 2012; Zmitrewicz et al., 2007) likely to compensate for the reduced forward propulsion from the amputated limb. These muscle compensations result in kinetic compensations at the lower limb joints. For example, the greater amputated limb hip extensors generate greater hip extensor moment (Bateni, 2002; Gitter et al., 1991; Winter and Sienko, 1988), work (Silverman et al., 2008), and power (Bateni, 2002; Gitter et al., 1991; Winter and Sienko, 1988) during early stance. On the intact limb, the greater hip extensor activity also results in greater hip joint work compared to non-amputees at faster walking speeds (Silverman et al., 2008). At the knee, the intact limb has greater positive joint work compared to the amputated limb and to non-amputees walking at a self-selected speed (Beyaert et al., 2008). However, this compensation is speed dependent as people with TTA increase positive intact limb knee work compared to the amputated limb and non-amputees (Sanderson and Martin, 1997; Silverman et al., 2008).

1.1.2 Biomechanics of turning

Compared to straight-line walking, turning is inherently asymmetric as it requires different mechanics between the inside and outside legs in order to quickly reorient the head, trunk, and pelvis (Courtine and Schieppati, 2003a). Asymmetric temporal-spatial parameters normally found during turning include shorter inside leg strides (Courtine and Schieppati, 2003b; Orendurff et al., 2006), and greater stance duration (Courtine and Schieppati, 2003b; Ventura et al., 2015) on the inside leg compared to the outside leg. Kinetically, turning requires greater medial GRFs and impulses on the outside limb to redirect the body COM towards the inside leg (Glaister et al., 2008; Orendurff et al., 2006). Furthermore, while medial impulses accelerate the body toward the contralateral limb during straight-line walking, impulses from both legs accelerate the body in the direction of the turn (Orendurff et al., 2006). This difference between turning and walking in a straight-line indicates that the neuromuscular strategy has to be altered to change direction. In particular, the difference in function between the inside leg during turning compared to straight-line walking results in greater differences in muscle contributions compared to the outside leg vs. straight-line comparison (Ventura et al., 2015). Specifically, the inside leg hip abductors and adductors, gastrocnemius, and tibialis posterior have altered contribution to medial COM acceleration impulse compared to straight-line walking (Ventura et al., 2015). As the inside leg has to accelerate the COM in the opposite direction during turning compared to straight-line walking, inside leg muscle modulation may be more important than outer leg muscle modulation

(Ventura et al., 2015). In addition, the soleus and gastrocnemius on the inside leg have greater contribution to medial COM acceleration compared to the outside leg (Ventura et al., 2015), indicating the importance of ankle muscle force modulation on the inside leg during turning.

As people with TTA lack ankle muscle modulation and have asymmetric gait, the asymmetric characteristics of turning result in altered kinematics, kinetics, and muscle activity for various degrees of turning compared to non-amputees. For example, the stride length asymmetry during circular turning is exacerbated in people with TTA as the inside leg stride length is even smaller compared to non-amputees (Segal et al., 2011). In addition, people with TTA spend more time on the intact limb during 180 degree turns regardless if it is on the inside or outside of the turn (Clemens et al., 2018b). The greater time spent on the intact limb is likely associated with the greater intact limb kinetics on either side of a circular turn compared to the amputated limb (Segal et al., 2011; Ventura et al., 2011). For example, people with TTA have greater intact limb internal rotation knee moment compared to the amputated limb (Segal et al., 2011). In addition, the intact limb external hip rotation moment (Segal et al., 2011) and frontal hip work (Ventura et al., 2011) are also greater when the intact limb is on the inside of the turn. The greater external hip rotation moment of the intact limb when it is on the inside of a turn likely compensates for the reduced hip, knee, and ankle internal rotation moments of the amputated limb on the outside of a turn (Segal et al., 2011). Moreover, the reduced amputated limb rotation moments on the outside of the turn may be used to avoid excessive rotation between the residual limb and socket. Despite these differences between intact and amputated limb when compared on the same side of a turn, people with TTA have similar medial-lateral GRF patterns and GRF impulses between limbs on both sides of the turn (Segal et al., 2011; Ventura et al., 2011). While the outside leg GRF is similar to non-amputees, the inside leg GRF pushes the body toward the turn throughout most of the stance phase and then pushes the body away from the turn at the end of the stance phase (Segal et al., 2011; Ventura et al., 2011). This GRF pattern on the inside leg occurs for both the amputated and intact limbs suggesting a compensatory strategy to maintain the body COM within the base of support. In particular, it is suggested that this change in direction of GRF may be caused by trunk lean away from the turn to ensure stability. This trunk lean away from the turn may contribute to reduced frontal plane range of motion found in a small sample of lower limb amputees (Golyski and Hendershot, 2018). However, this study analyzed people with TTA and people with a transfemoral amputation as one group. As such, the strategies people with TTA use to maintain balance while turning is unclear.

Similar to straight-line walking, people with TTA have greater intact limb contributions to forward progression. In particular, the intact limb provides greater propulsion compared to the prosthetic limb when on either side of the turn (Ventura et al., 2011). The amputated limb hip joint also generates greater positive hip work compared to the intact limb and to non-amputees when on the inside of the turn. This greater positive hip work is also similar to straight-line walking where greater hip extensor work compensates for the lack of forward progression provided by the amputated leg (Bateni, 2002; Silverman et al., 2008). However, as the greater amputated limb hip work only occurs when the prosthesis is on the inside of the turn, turning with the prosthesis on the inside of a turn may be more challenging than turning with the prosthesis on the outside of a turn for people with TTA.

1.1.3 Biomechanics during seat transfers

Transferring from a seat can be one of the most neuromuscular demanding tasks of daily life due to the need for large force generation and balance control. In addition, there are many factors that can affect the biomechanics of seat transfers. These factors can be divided into chair-related, subject-related, or strategy-related categories (Janssen et al., 2002). The chair-related category includes factors such as chair height, armrests vs. no armrests, type of chair, backrests versus no backrest. The subject-related category includes factors such as age, physical condition/disorder, muscle force capability, footwear. Lastly, the strategy-related category includes factors that can change the biomechanics used to complete the seat transfer such as speed, foot position, and arm movement to name a few. As such, all these factors may affect seat transfer biomechanics for people with TTA.

Seat transfers such as sit-to-stand requires the trunk and legs for control, coordination (Roebroeck et al., 1994), and postural adjustment (Rodosky et al., 1989). The goal of the sit-to-stand transfer is to move the COM forward and upward from sitting to standing while maintaining balance (Roebroeck et al., 1994). Specifically, the trunk is used to accelerate the body COM forward while the knee extensors and hip extensors accelerate the body upward (Jeon et al., 2019). In addition, the tibialis anterior and soleus muscles provide the ankle joint stabilization required for upward momentum transfer (Jeon et al., 2019). Therefore, performing seat transfers may be more challenging for people with TTA as the loss of direct neuromuscular control of the ankle joint may reduce their ability to provide stabilization during the changes in momentum.

Similar to walking and turning, people with TTA are able to perform seat transfers, like sit-to-stand, by using compensatory strategies that lead to asymmetry between limbs. For example, knee extensor activity on the amputated limb is greater compared to the intact limb and non-amputees (Wagner et al., 2020). Moreover, the intact limb has less hip abductor activity compared to non-amputees (Wagner et al., 2020). The greater knee extensor activity on the amputated limb likely compensates for the lack of ankle dorsiflexor function which assists with horizontal braking in a functional ankle joint (Caruthers et al., 2016). The greater intact limb hip abductor activity is likely associated with the weight shift toward the amputated limb at the end of the sit-to-stand movement (Wagner et al., 2020). As people with TTA have asymmetric weight-bearing during sit-to-stand compared to non-amputees (Actis et al., 2018b; Agrawal et al., 2011; Ozyurek et al., 2014), this shift in body weight at the end of the sit-to-stand movement for stabilization is not seen in non-amputees. The greater loading on the intact limb also results in greater intact side knee joint forces compared to the amputated limb (Ferris et al., 2017a; Slajpah et al., 2013), and greater lateral and axial trunk range of motion compared to non-amputees (Actis et al., 2018b).

The compensatory strategies used by people with TTA to transfer from sit-to-stand persist even when seat transfer factors are altered. For example, there is no difference in GRF symmetry or the time to complete the sit-to-stand movement when using arm rests compared to placing hands on the knees (Agrawal et al., 2011). However, compared to non-amputees, transferring from sit-to-stand took longer without armrests compared to with armrests for people with TTA (Agrawal et al., 2011). The difference in rise time is likely due to the lower hip and knee moments needed when using armrests (Janssen et al., 2002) which indicates a reduced demand on the lower limbs. However, while armrests may reduce the hip and knee moments for people with TTA, asymmetrical loading is still prevalent. In particular, the greatest asymmetry is found at the time when body support is being transferred from the seat to the legs (i.e., seat-off). Similarly, transferring from sit-to-stand from different seat heights and at various speeds did not affect the asymmetric vertical GRF, and hip, knee, and ankle joint angles and moments which were greater compared to non-amputees (Slajpah et al., 2013). However, performing the sit-to-stand at a slower speed was found to increase trunk flexion in people with TTA and non-amputees (Slajpah et al., 2013). As people with TTA perform the sit-to-stand movement slower compared to non-amputees, they may have increased trunk compensations to shift the body COM forward. Furthermore, greater trunk lateral bending and axial rotation at seat-off is associated with greater low back compression loads during sit-to-stand found in people with TTA compared to non-amputees using a musculoskeletal model (Actis et al., 2018b).

During the stand-to-sit transfer, people with TTA also have greater weight-bearing on the intact limb compared to the amputated limb resulting in greater asymmetry compared to non-amputees (Agrawal et al., 2011). Similar to the sit-to-stand movement, the greatest asymmetry during stand-to-sit occurs when transferring the body weight from the legs to the chair (Agrawal et al., 2016). In addition, this strategy was not influenced by the use or armrests (Agrawal et al., 2016).

While some studies have investigated the strategies people with TTA use to transfer from sit-to-stand and, to a lesser extent, stand to sit, other seat transfer tasks such as sit-to-walk, and walk-to-sit remain unexplored in people with TTA. As sit-to-walk requires moving from a stable position (two legs for support) to a less stable position (one leg for support) and vice-versa for walk-to-sit, it is important to understand the strategies people with TTA use to complete these tasks.

1.2 Dynamic Balance in People with TTA

An important consideration for improving functional mobility of people with TTA is their ability to control and maintain dynamic balance, or stability during movement. People with TTA have an increased risk of falling and fall-related injuries compared to non-amputees (Kulkarni et al., 1996; Miller et al., 2001b) likely due to lower muscle strength (Pedrinelli et al., 2002), reduced sensory feedback (Kavounoudias et al., 2005), and the inability to actively control the prosthetic ankle joint following a balance disturbance (Buckley et al., 2002; Vanicek et al., 2009a). These functional limitations also contribute to reduced physical activity (Bussmann et al., 2008; Klute et al., 2006) and decreased participation in leisure/social activities (Couture et al., 2010; Miller et al., 2001a) compared to before limb loss. As restriction of activity leads to balance deterioration, reduced endurance, strength, flexibility, and coordination, it appears that fall risk in people with TTA is associated with a cycle between psychological and physical factors (Miller et al., 2001b). Balance confidence is a psychological factor that provides a sensitive measure of fear of falling (Miller et al., 2002) and has been found to be related to functional mobility capacity and participation in social activities (Miller et al., 2001a). As such, people with TTA who have decreased balance confidence have decreased mobility and participation in community activities (Miller et al., 2002).

Balance can be defined as maintaining the line of gravity within the base of support which is the primary requirement during standing. As such, quantifying the capability to maintain standing balance is the primary focus of clinical balance assessments. The Berg Balance Scale (BBS) involves the completion of 14-items that focus on the ability to maintain a static position while performing a postural adjustment (Berg et al., 1992). People with TTA performed worse on the BBS if they had a fear of falling or used a mobility aid (Major et al., 2013). The Functional

Reach Test measures the maximum distance an individual can reach forward while standing in a fixed position (Duncan et al., 1990) and has been used to assess balance ability after balance training over four weeks (Damayanti Sethy et al., 2009). As such, clinical assessments can identify general balance capability and confidence, and gauge the effect of rehabilitation interventions on standing balance. However, these assessments do not provide insight on the mechanisms used to maintain standing balance.

Laboratory measures have been used to further understand how people with TTA maintain standing balance. A common measure is weight-bearing symmetry between the legs during standing which typically finds increased loading on the intact limb for people with TTA (Ku et al., 2014; Nadollek et al., 2002) suggesting altered balance control. Another study used a force plate to investigate the ability to balance on only the intact limb and found that it predicted functional outcomes (Schoppen et al., 2003). Center of pressure-based measures have been used extensively (Buckley et al., 2002; Isakov et al., 1992; Jayakaran et al., 2012; Paráková, 2009) and found that people with TTA generally have greater postural sway compared to non-amputees during standing (Buckley et al., 2002; Isakov et al., 1992; Jayakaran et al., 2012). In addition, perturbation studies have found that people with TTA rely on the intact leg to maintain standing balance (Curtze et al., 2012; Vanicek et al., 2009b). While understanding the mechanisms used to maintain standing balance is important for functional mobility, these mechanisms may not translate to maintaining dynamic balance during movement. Therefore, it is important to understand how people with TTA maintain dynamic balance during functional and dynamic tasks.

For dynamic tasks, more advanced methods have been used such as margin of stability (Hof, 2008) and whole-body angular momentum (Herr and Popovic, 2008). Margin of stability has been used to measure balance during walking on level and irregular surfaces in people with TTA (Beltran et al., 2014; Curtze et al., 2011; Gates et al., 2013; Hak et al., 2013; McAndrew Young et al., 2012; Sinitski et al., 2019). In general, these studies reported that people with TTA were either more mediolaterally stable than controls or showed no differences (Watson et al., 2021). While these results provide insight on walking balance in people with TTA, margin of stability is based on the inverted pendulum theory of standing balance (Hof et al., 2005) which may limit its applicability to other functional tasks. In addition, the human body is represented as a single mass supported by a mass-less leg in the inverted pendulum theory. These assumptions suggest that margin of stability cannot provide insight on how body segments contribute to maintaining dynamic balance during functional tasks.

Whole-body angular momentum is a muscle generated measure that consists of the sum of all segment momenta (Herr and Popovic, 2008). Greater angular momentum away from zero indicates an increased risk of losing balance. As such, angular momentum is tightly regulated during walking (Herr and Popovic, 2008). For people with TTA this regulation differs compared to non-amputees during different walking tasks. For example, people with TTA have greater whole-body angular momentum when walking in a straight-line (Silverman and Neptune, 2011), during stair ascent (Pickle et al., 2014), and when walking on slopes (Pickle et al., 2016) compared to non-amputees. As the time rate of change of angular momentum is the external moment on the body, these differences are attributed to differences in stepping and GRF in people with TTA. As such, prosthetic properties have also been shown to alter angular momentum during turning (Pew et al., 2019; Shell et al., 2017). In addition, increased frontal whole-body angular momentum during walking has been associated with increased margin of stability and lower scores (i.e., decreased balance) on clinical assessments in other populations with gait impairments (Nott et al., 2014; Vistamehr et al., 2016). As such, whole-body angular momentum may be useful for quantifying the risk of losing balance for any functional task.

1.3 Musculoskeletal Modeling and Simulation

While the study of human movement has traditionally relied on observational analyses, understanding the underlying principles of human movement has required advances in technologies to measure biomechanical variables. Some of these variables such as GRFs or muscle activity can be directly measured using sensors while other variables such as joint angles, moments, and joint contact forces require a model of the human body. As muscles generate movement, musculoskeletal models that include models of muscular activation and force generation have been developed to gain insight on human movement. These models are complex as they require accurate representations of the muscles, skeletal geometry, and the connections at the joints. However, these models enable researchers to calculate biomechanical variables that cannot be measured directly or require undesired invasive methods. In addition, these models can be used to simulate human movement to understand how the elements of the musculoskeletal system interact to produce movement (Anderson and Pandy, 2001; Neptune et al., 2001; Neptune et al., 2004; Umberger, 2010; Umberger et al., 2006; Zajac et al., 2002, 2003), to identify which elements affect movement disorders (Hu and Blemker, 2015; Jansen et al., 2014; Silverman and Neptune, 2012), and to evaluate potential treatments (Arnold and Delp, 2005; Fey et al., 2012; Grabke et al., 2019; LaPre et al., 2014; Mansouri et al., 2016).

Depending on the research question, musculoskeletal simulations use many tools to extract variables of human movement. These tools include inverse kinematics, inverse dynamics, and forward dynamics. Inverse kinematics is often used to calculate joint angles by resolving internal joint coordinates from spatial marker positions on skeletal landmarks and rigid body segments. Inverse dynamics is used to reconstruct the net internal forces/torques (e.g., joint moments) from kinematics and known external forces. Forward dynamics identifies the kinematics that result from known internal forces/torques (e.g., muscle forces/activations). A specific case of forward dynamics that is independent of experimental data is predictive simulation which has been used for clinical applications on investigating the effect of different assistive devices (Nguyen et al., 2019) or rehabilitation practices on human movement. Simulations are used to estimate the muscle forces required to generate movement by solving the force distribution problem. The force distribution problem describes the fact that the musculoskeletal system is highly redundant as the internal joint forces are distributed among many individual anatomical structures including muscles, ligaments, and articular surfaces. The redundancy of these anatomical structures presents many more unknowns than independent equations in the simulation framework (Crowninshield, 1978; Crowninshield and Brand, 1981). Thus, to solve for these muscle forces, the model can either be simplified (e.g., ignore some anatomical structures) or use an optimization framework that finds a solution that minimizes/maximizes some process or action, or a combination of both where only anatomical structures deemed important to the research questions are used in an optimization framework. Mathematically, the optimal criterion is expressed as a “cost function” or “objective function”. These objective functions are based on the idea that the central nervous system selects muscles for a given movement according to some criterion (e.g., minimize force, stress, energy, maximize stability, smoothness). Thus, the objective function used for musculoskeletal simulations depends on the goal of the movement being simulated which may be related to muscle function (e.g., minimize muscles forces) or some aspect of the motion (e.g., maximize walking speed).

For people with TTA, these simulations have been used in a wide variety of cases. Specifically, studies have used musculoskeletal simulations with models representing people with TTA to understand individual muscle compensation during walking (Fey et al., 2013; Pickle et al., 2017a; Russell Esposito and Miller, 2018; Silverman and Neptune, 2012; Yoder et al., 2015; Zmitrewicz et al., 2007), joint loading during walking (Fang et al., 2009; Karimi et al., 2017; Koelewijn and van den Bogert, 2016; Silverman and Neptune, 2014; Yoder et al., 2015) and sit-to-stand (Actis et al., 2018b), and the effects of prosthesis properties on a number of variables (Fang et al., 2007; Fang et al., 2009; Fey et al., 2012; Handford and Srinivasan, 2016; LaPre et al., 2014; Pickle et al., 2017a). The variables estimated with simulations for people with TTA, or other pathologies affecting movement, are important as they can provide insight on the underlying causes that lead to the development of secondary conditions in these populations. For people with TTA, including muscle forces in joint contact load estimates can be helpful to understand the development of joint pain, especially at the hip, knee, and low back, which can lead to secondary joint degeneration and osteoarthritis prevalent in this population. In addition, joint contact load estimates simulations that track an individual's movement may provide insight on that individuals risk of developing secondary conditions. Furthermore, simulations that alter model properties may provide valuable insight for rehabilitation or prosthetic design interventions. Thus, using musculoskeletal models to simulate human movement in people with TTA has important clinical implications and may provide insight to improve functional mobility.

1.4 Effect of Prosthetic Alignment on Functional Tasks

As described above, the prosthetist's primary job is to select prosthetic components and align those components to enable functional mobility and ensure comfort throughout daily life. The alignment process has three steps: bench top, static, and dynamic alignment (Krajbich, 2016). Bench top alignment is putting the components together per the manufacturer's specifications before presenting the prosthesis to the patient. Static alignment consists of the patient wearing the prosthesis while standing. The goals for the prosthetists during static alignment are to 1) enable proper and symmetrical loading of the prosthesis, and 2) ensure comfortable pressures at the socket-residuum interface. With these goals in mind, the prosthetist makes adjustments based on their visual assessment and patient feedback on socket fit and comfort. The final step in the alignment process is dynamic alignment. During dynamic alignment, the primary goal is to minimize gait asymmetries during walking. This alignment step involves the prosthetist visually observing the patient walk in a straight-line and asking the patient for their feedback. Thus, the final alignment is dependent on the experience of the prosthetist and the ability of the patient to provide insightful feedback which may affect whether the goals are met. This subjective process can lead to differences in alignment for the same person between prosthetists (Geil, 2002; Zahedi, 1986), different alignments on different days prescribed the same prosthetists (Zahedi, 1986), and thus a range of alignments are deemed acceptable by the user and prosthetist (Blumentritt, 1997; Chow et al., 2006; Sin et al., 2001; Zahedi, 1986).

As alignment affects how the loads transfer from the floor to the rest of the body it can affect the compensatory strategies used by people with TTA during daily life. Moreover, as prosthetist primarily focus on minimizing weight-bearing asymmetry during standing and kinematic asymmetry during walking, many studies have investigated the effect of alignment on standing (Blumentritt et al., 1999; Boone et al., 2012; Isakov et al., 1994; Jia, 2009; Kolarova et al., 2013; Luengas et al., 2017; Paráková, 2009; Seelen et al., 2003; Xiaohong et al., 2005) and walking (Andres and Stimmel, 1990; Beyaert et al., 2008; Chow et al., 2006; Fiedler et al., 2016; Fridman et al., 2003; Geil, 2002; Geil and Lay, 2004; Grumillier et al., 2008; Hannah et al., 1984; Hansen, 2008; Pinzur et al., 1995; Rossi, 1995; Schmalz et al., 2002; Sin et al., 2001; Van Velzen et al., 2005; Yeung et al., 2013) biomechanics.

Prosthetic alignment affects standing biomechanics by affecting the center of pressure and direction of the GRF vector. The ability to remain standing requires control of the center of pressure to maintain upright balance. As such, the lack of prosthetic ankle control in people with TTA results in greater intact limb anterior-posterior forces to maintain standing balance even with altered rotational (Isakov et al., 1994; Paráková, 2009) and translational (Paráková, 2009) alignments. On the amputated limb, alignment changes the location of the center of pressure. For example, sagittal (Jia, 2009; Luengas et al., 2017; Xiaohong et al., 2005) and frontal (Xiaohong et al., 2005) plane rotations between the foot and socket results in posterior center of pressure position compared to the prescribed alignment. As such, these changes in center of pressure also cause changes in the knee position to generate appropriate forces to balance the GRF vector (Xiaohong et al., 2005). The altered knee position requires altered knee muscle activation (Blumentritt et al., 1999) dependent on the alignment. For example, plantarflexion of the prosthetic foot results in increased amputated limb knee flexor activity while dorsiflexion of the foot results in increased knee extensor activity (Blumentritt et al., 1999). As the knee position affects the interaction between the socket and the residual limb, socket comfort is also dependent on alignment during standing. Plantarflexion of the foot reduces anterior tibia pressure but increases sub-patellar pressure while dorsiflexion has the opposite effect (Seelen et al., 2003).

In addition to focusing on socket comfort and reducing asymmetric weight-bearing like during standing (i.e., static alignment), prosthetists also focus on reducing gait asymmetries during walking (i.e., dynamic alignment). To reduce asymmetries, prosthetists adjust alignments based on observable gait parameters as they can be affected by alignment. For example, an alignment with excessive prosthetic foot external rotation will increase stance time, step length, and swing time asymmetry compared to an alignment with less external rotation (Fridman et al., 2003). Internal rotation only decreased single support phase on the amputated limb but also caused reduced knee flexion for both limbs (Beyaert et al., 2008) and increased hip flexion on the intact limb (Grumillier et al., 2008). A posterior foot translation relative to the socket causes reduced amputated limb stance time and increased knee flexion (Andres and Stimmel, 1990). In addition to changes to gait parameters that can be observed, changes in prosthetic alignment have also been shown to affect limb loading (Beyaert et al., 2008; Grumillier et al., 2008; Pinzur et al., 1995; Van Velzen et al., 2005), roll-over shape (Hansen, 2008); oxygen consumption (Schmalz et al., 2002), and socket-residuum pressures (Sanders et al., 1998; Sanders and Daly, 1999; Seelen et al., 2003). While the user may not be able to provide feedback on limb loading, roll-over shape, or oxygen consumption and they cannot be observed by the prosthetist, socket-residuum pressures are important for assessing alignment. While it is not clinically feasible to measure socket pressures, there is special interest in the forces at the socket-residuum interface (i.e., socket reaction moments) as these forces can provide insight on which part of the residual limb is experiencing pressure due to rotation of the prosthesis. In addition, socket reaction moments have been found to be more sensitive to small changes in alignment (Hashimoto et al., 2018a; Hashimoto et al., 2018b; Jonkergouw et al., 2019; Kobayashi et al., 2014a; Kobayashi et al., 2014b; Kobayashi et al., 2013a; Kobayashi et al., 2012, 2013b, 2016).

While alignment is an important process for enabling people with TTA to return to their activity levels and functional mobility before limb loss, the alignment process is not inclusive of all functional tasks. However, our research group found that medial and lateral alignment shifts affect muscle activity during sit-to-stand (Wagner et al., 2020). Therefore, further studies are needed to understand the effect of alignment on other functional tasks.

1.5 Summary of dissertation

Performing tasks such as turning or transferring from a seat are crucial for functional mobility and enabling independence for people with TTA. While turning steps make up 35% of the steps taken daily (Glaister et al., 2007) and 50-60 seat transfers are performed every day (Bussmann et al., 2004; Bussmann et al., 2008), the majority of the literature on biomechanics of people with TTA focus on straight-line walking. As such, clinical care also primarily focuses on assessing straight-line gait asymmetries despite studies showing that people with TTA mostly perform 1-2 min bouts consisting of fewer than 17 steps/min (Klute et al., 2006). Most of these steps are likely turning steps beginning from getting out of a chair. Thus, there is a clinical need for quantifying how the lack of neuromuscular ankle control affects turning and seat transfer tasks so that clinicians can improve functional mobility for people with TTA. Therefore, the purpose of this dissertation is to determine how people with TTA maintain dynamic balance during turning and seat transfers and to quantify the effects of prosthetic alignment during seat transfers.

The **first aim**, presented in Chapter 2, was to **determine how body segments contribute to maintaining dynamic balance during 90-degree turns in people with a unilateral transtibial amputation.** Eight (8) people with a unilateral transtibial amputation and eight age- and sex- matched people without an amputation performed left and right 90-degree step turns. The range of whole-body angular momentum and the positive and negative segment momenta contribution to whole-body angular momentum were compared between turning with the prosthesis on the inside and outside of a turn as well as between people with and without TTA. I hypothesized that there would be a greater range of whole-body angular momentum when turning with the prosthesis on the inside of a turn compared to when the prosthesis was on the outside. I also hypothesized that the intact leg contribution would be greater than the prosthetic leg contribution for both left and right turns. Furthermore, I hypothesized people with TTA would have greater trunk contributions to dynamic balance compared to controls.

Chapter 3 presented the **second aim** which was to **quantify the effect of alignment on dynamic balance during functional tasks in people with a unilateral transtibial amputation.** Ten (10) people with a unilateral transtibial amputation and 10 age- and sex- matched people without an amputation performed the following activities: level ground walking at self-selected, sit-to-stand, stand-to-sit, sit-to-walk, and walk-to-sit. Participants with TTA performed these tasks with 7 different alignments including the prescribed alignment, alignments with 10 mm shifts of the prosthetic foot in the anterior, posterior, medial, and lateral conditions, and ±20 mm shifts in the vertical direction. The range of whole-body angular momentum and the number of adjustments to whole-body angular momentum for each task were compared between the prescribed and all other alignments. The prescribed alignment condition was also compared to controls. I hypothesized that medial/lateral and tall/short alignments would affect frontal and transverse whole-body angular momentum while anterior/posterior alignments would affect sagittal plane whole-body angular momentum.

The **third aim** was to **determine the effects of anterior-posterior alignment shifts on movement strategies used during sit-to-stand** and is presented in Chapter 4. Nine (9) people with a unilateral transtibial amputation and nine age- and sex- matched people without an amputation performed five sit-to-stand trials. Participants with TTA performed the sit-to-stand trials with three different translational alignments. The prescribed alignment was the first alignment followed by 10 mm anterior and posterior alignment shifts. Three-dimensional ground reaction force impulses, sagittal plane knee moments, anterior/posterior center of pressure position, and 3D trunk range of motion between alignments. We also compared dependent measures between the people with TTA with their prescribed alignment and control participants. This work tested that hypothesis that shifting the prosthetic foot posterior to the socket will decrease braking impulse and reduced trunk range of motion compared to the prescribed and anterior alignments.

The **fourth aim**, presented in Chapter 5, was to **determine the effect of prosthetic alignment on hip and low-back joint contact forces during sit-to-stand in people with a unilateral transtibial amputation.** Nine (9) people with a unilateral transtibial amputation performed five self-paced sit-to-stand trials with seven different translational alignments. The prescribed alignment was the first alignment followed by 10-mm anterior, posterior, medial, lateral, and 20-mm tall and short alignment shifts. Kinematics and ground reaction forces for each trial were applied to a musculoskeletal model (Actis et al., 2018a) with trunk, low-back, lower limb musculature, and a transtibial prosthesis (LaPre et al., 2018). The translational alignment of the prosthesis model was changed to match the change in prosthetic alignment of the experimental data. A static optimization algorithm was used to find the muscle activation that minimizes the sum of the squared muscle activations at each instant in time. The muscle forces were then used to calculate the hip and L4-L5 joint reaction contact forces during seat-off to stand. Peak hip and L4-L5 joint contact force magnitude and impulses were compared between alignment conditions and between people with and without TTA. I hypothesized that the tall and short alignment would affect hip and L4-L5 joint contact forces. I also hypothesized that medial/lateral alignment changes would primarily affect hip joint contact forces and that anterior/posterior alignment changes would primarily affect L4-L5 joint contact forces.

Chapter 6 includes a discussion of how a transtibial amputation affects dynamic balance while completing different functional tasks as well as how prosthetic alignment affects seat-transfer tasks. Overall study limitations and suggestions for future research are also discussed in Chapter 6. This work uniquely contributes to our understanding of the biomechanical adaptations during daily life that result from a lower limb amputation. First, it determines how having a lower limb amputation affects segmental contributions to dynamic balance during 90-degree turns by comparing between turns with the prosthesis on the inside of a turn to when it is on the outside. This work also quantifies the effect of acute changes in prosthetic alignment on dynamic balance during various seat-transfer tasks that are important for functional mobility throughout daily life.

Lastly, this work determines the effect of acute changes on prosthetic alignment on movement strategies and joint loads experienced during the sit-to-stand task. Together, these contributions help further understand the effect of lower limb loss on balance during functional tasks of daily living and the role that acute changes in prosthetic alignment may have on the movement strategies used during these tasks.

CHAPTER 2. The Ins and Outs of Dynamic Balance During 90-Degree Turns in People With a Unilateral Transtibial Amputation[1]

2.1 Abstract

The ability to maintain balance when turning is essential to functional and independent living. Due to the lack of neuromuscular ankle control on the prosthetic side in people with a transtibial amputation (TTA), turning is likely more challenging. The purpose of this study was to quantify how people with TTA maintain dynamic balance during 90-degree turns made with the prosthesis on the inside and outside of the turn compared to people without amputation. Eight participants with TTA and eight age-, height-, and sex- matched non-amputee controls performed left and right 90-degree step turns at a self-selected speed. The primary outcomes were range of whole-body angular momentum and positive and negative contributions of six segment groups (head/trunk, pelvis, arms, and legs) to whole-body angular momentum during the continuation stride. Participants with TTA had greater range of frontal- and sagittal-plane whole-body angular momentum when turning with the prosthesis on the inside compared controls. They also had a greater range of whole-body angular momentum in all planes of motion when turning with the prosthesis on the inside compared to outside of the turn. The contributions for the head/trunk and inside and outside legs differed between groups and turns, suggesting altered interactions between segment momenta to compensate for the reduced contribution of the amputated leg. This study

[1] A version of this chapter is published as Nolasco, L. A., Livingston, J., Silverman, A. K., & Gates, D. H. (2021). The ins and outs of dynamic balance during 90-degree turns in people with a unilateral transtibial amputation. *Journal of biomechanics*, 122, 110438. https://doi.org/10.1016/j.jbiomech.2021.110438

provides insight into possible training paradigms to reduce the high incidence of turn related falls in people with TTA and, potentially, ways to alter prosthetic function to promote balance control.

2.2 Introduction

The successful execution of turns is important for functional and independent living. For people with a transtibial amputation (TTA), the lack of neuromuscular ankle control likely makes turning more challenging. The lack of active ankle control reduces the ability to modulate ground reaction forces (GRFs) with the amputated leg, increases reliance on the intact leg (Curtze et al., 2012), and leads to greater temporal-spatial variability compared to non-amputees (Beltran et al., 2014; Gates et al., 2012; Hak et al., 2013; Segal et al., 2015; Sinitski et al., 2019). People with TTA also have decreased amputated leg stability when turning compared to straight-line walking (Segal et al., 2010). Perhaps unsurprisingly, given these deficits, 50% of people with TTA report one or more falls each year (Kim et al., 2019). In addition, people with TTA primarily report falling due to slipping (Kim et al., 2019), which is more common during turning compared to straight-line walking (Yamaguchi et al., 2012). However, the mechanics of how people with TTA maintain dynamic balance while successfully completing a turn is not well understood.

One way to quantify dynamic balance is whole-body angular momentum ($\vec{H}_{WB}$), which is the sum of all body segment momenta about the body center-of-mass (COM). Deviations in $\vec{H}_{WB}$ represent attributes of segment rotations that could lead to a fall if not regulated. As such, decreased regulation of $\vec{H}_{WB}$ is associated with decreased balance (Nott et al., 2014; Vistamehr et al., 2016), suggesting increased fall risk. Prior studies have found that when normalizing $\vec{H}_{WB}$ by dividing by body mass, height, and walking speed, people with TTA have greater frontal and sagittal ranges of $\vec{H}_{WB}$ compared to non-amputees when walking in a straight line (D'Andrea et al., 2014; Silverman and Neptune, 2011) and up an incline (Pickle et al., 2016), suggesting difficulty maintaining balance during these tasks. During walking, non-amputees regulate $\vec{H}_{WB}$ through interactions between segment momenta (Herr and Popovic, 2008) which are altered during turning (Nolasco et al., 2019). The need for altered segment momenta may be due to the asymmetric GRFs (Glaister et al., 2008; Orendurff et al., 2006) and step lengths (Courtine and Schieppati, 2003a, b; Orendurff et al., 2006) used during turning, which affect the net external moment about the COM. As the time rate of change of $\vec{H}_{WB}$ equals the net external moment about the COM, the GRFs and their distance from the COM will affect how $\vec{H}_{WB}$ is regulated. As muscles are the primary generators of segment momenta, regulating $\vec{H}_{WB}$ during turning may be challenging for people with TTA who have reduced muscular control and strength.

People with TTA have different turning mechanics compared to non-amputees partly due to their lack of foot and ankle muscle function. For example, people with TTA have a shorter inside leg stride length when the amputated leg is on the inside of a circular turn compared to non-amputees (Segal et al., 2011). Moreover, the difference in medial-lateral GRFs between the inside and outside leg is greater compared to non-amputees, regardless of whether the prosthesis is on the inside or outside of a turn (Segal et al., 2011; Ventura et al., 2011). People with TTA also generate both medial and lateral ground reaction impulses on the inside leg during the stance phase of a circular turn whereas non-amputees only generate a medial impulse (Segal et al., 2011; Ventura et al., 2011). Shifting the ground reaction impulse from lateral to medial is associated with a longer effective limb length on the inside of a turn, (Segal et al., 2011) suggesting that people with TTA shift their trunk segments from leaning toward the turn to leaning away from the turn

(Golyski and Hendershot, 2018). This altered strategy is likely used to maintain the COM within the base of support and suggests muscular control of the trunk has an important role in maintaining dynamic balance during turning. Given these altered turning mechanics in people with TTA, previous studies have investigated balance during turning in this population (Pew et al., 2019; Segal et al., 2010; Shell et al., 2017). However, how whole-body dynamic balance differs between people with TTA and non-amputees during turning remains unclear. Furthermore, it is unknown how the movement of individual segments contribute to regulating dynamic balance in people with TTA.

The purpose of this study was to 1) quantify how people with TTA maintain dynamic balance during a 90-degree turn compared to non-amputees and 2) determine how balance regulation during turning is affected by whether the prosthesis is on the inside or outside of the turn. We expected people with TTA would have greater range of $\vec{H}_{WB}$ compared to non-amputees. We also expected the lack of force modulation from the amputated leg would result in a greater range of $\vec{H}_{WB}$ while on the inside of the turn compared to the outside. Lastly, we also hypothesized that the amputated leg would have a lower contribution to $\vec{H}_{WB}$ than the intact leg and that the trunk would contribute less when the prosthesis was on the inside of the turn.

2.3 Methods

Participants

Eight people with a unilateral transtibial amputation (TTA) were recruited through local prosthetists, while eight age-, height-, and sex- matched non-amputees were recruited through a local online database (https://umhealthresearch.org/) (Table 2.1). All participants were screened to ensure they did not have a history of cardiovascular or neurological disease, uncorrected vision problems, take medication that affected their ability to walk, or have any mental capacity impairment that would negatively affect verbal communication. Potential participants with TTA were also excluded if they could not walk independently for at least 10 minutes at a time for at least two months prior to data collection. No participants with TTA reported any significant injury or pathology of their intact leg. Potential participants without an amputation were also excluded if they had a history of significant musculoskeletal injuries that affected their ability to walk. All participants provided their written informed consent to participate in this institutionally approved study.

Experimental Protocol

All participants performed 90-degree turns at a self-selected speed, to the right and then to the left until 10 step-turns were collected. In an open lab space, participants were instructed to turn around a cone as if they were "turning around a hallway corner." When instructed to "Go", participants walked along a 2.4 m walkway, turned around a cone, and came to a stop after taking 3 steps following the last turning step. This turn was chosen as it is within the range of common turning angles performed in daily life (Sedgman et al., 1994). We focused on step-turns as they are more common than spin-turns during activities of daily living (Glaister et al., 2007) and people with TTA prefer step-turns over spin-turns (Golyski and Hendershot, 2018). Each turn was visually determined to be a step-turn if the first step taken after body rotation was taken with the outside leg (Glaister et al., 2008). The turn was repeated if it was a spin-turn. All participants wore their own comfortable walking shoes, and participants with TTA used their prescribed prosthesis (Table 2.1).

Table 2.1. Participant demographics.

TTA	Age (years)	Sex	Weight (kg)	Height (m)	Amputated Side	Prosthetic foot	K-Level	Cause	Control	Age (years)	Sex	Weight (kg)	Height (m)
1	25	F	63.9	1.66	L	Endolite Elite 2	K4	Cancer	1	21	F	66.0	1.70
2	56	M	99.6	1.70	L	Ossur LP Variflex with EVO	K4	Trauma	2	52	M	69.6	1.69
3	58	M	101.8	1.78	R	College Park Velocity	K4	Trauma	3	48	M	116.1	1.91
4	63	M	121.9	1.81	R	College Park TruStep	K3	Trauma	4	62	M	84.4	1.81
5	65	M	90.4	1.67	R	Ottobock Triton 1C66	K3	Trauma	5	61	M	79.8	1.78
6	31	M	93.4	1.82	L	Freedom Innovations Maverick	K3	Trauma	6	27	M	69.4	1.75
7	61	M	115.6	1.88	R	Fillaeur AllPro	K4	Trauma	7	53	M	108.4	1.78
8	56	M	109.6	1.82	L	Freedom Innovations Agilix	K3	Trauma	8	54	M	99.8	1.97
Mean ± SD	51.9 ± 15.2	-	99.53 ± 17.94	1.77 ± 0.08	-	-	-	-	-	47.25 ± 15.15	-	86.69 ± 19.2	1.80 ± 0.10

A 20-camera motion capture system (Motion Analysis, Santa Rosa, CA) tracked full-body motion at 120 Hz using a marker set that allows six degrees-of-freedom at each joint (Collins et al., 2009; Wilken et al., 2012) with 67 reflective markers. For the TTA group, the amputated side markers were placed so they mirrored the intact side.

Data Analysis

Turning was divided into initiation, continuation, and termination strides (Figure 2.1). The continuation stride was defined as the stride of the pivot foot, which for a step-turn is the foot on the outside of a turn. The initiation and termination strides were the strides taken with the inside leg before and after the continuation stride, respectively.

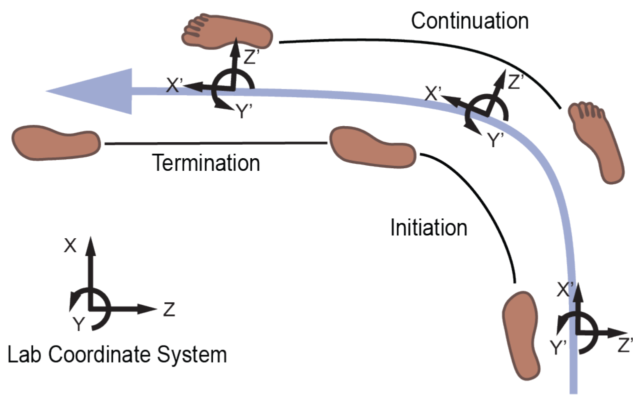

Figure 2.1. Turning strides for a left 90-degree turn for a person with a left side amputation with approximate body-based reference frame rotated throughout the turn relative to the lab coordinate system.

Marker trajectories were filtered using a 4th-order low-pass Butterworth filter using a cutoff frequency of 6 Hz. A 13-segment model consisting of the head, trunk, upper arms, forearms, pelvis, thighs, shanks, and feet was developed in Visual3D (C-Motion, Germantown, MA) with segment inertial properties based on previous work (Dempster, 1955; Hanvan, 1964). The inertial properties for the prosthesis were adjusted according to (Ferris et al., 2017b). Whole-body COM location was calculated from the 13-segment model. Segment ($\vec{H}_i$) and whole-body ($\vec{H}_{WB}$) angular momenta were calculated as:

$$\vec{H}_i = [(\vec{r}_i^{COM} - \vec{r}_{body}^{COM}) \times m_i(\vec{v}_i^{COM} - \vec{v}_{body}^{COM}) + I_i\vec{\omega}_i] \qquad (1)$$

$$\vec{H}_{WB} = \sum_{i=1}^{n} \vec{H}_i \qquad (2)$$

where $\vec{r}_i^{COM}$ and $\vec{v}_i^{COM}$ are the position and velocity of the i-th segment's COM, $\vec{r}_{body}^{COM}$ and $\vec{v}_{body}^{COM}$ are the position and velocity of the whole-body COM, m_i is the segment mass (Dempster, 1955),

I_i is the segment moment of inertia, $\vec{\omega}_i$ is the segment angular velocity, and n is the total number of segments. All angular momenta were normalized by dividing by body mass (kg), height (m), and average walking speed (m/s) across all three strides (Nolasco et al., 2019). Time was normalized to 0-100% of a stride.

We transformed $\vec{H}_{WB}$ and $\vec{H}_i$ vectors from the global reference frame (X, Y, Z) into the reference frame aligned with the participants' path trajectory (X', Y', Z') at each instant to interpret angular momentum in the context of the anatomical planes of motion (Nolasco et al., 2019). The angle of rotation was defined as the angle between the heading direction and the global reference frame, where the heading direction was the forward component of the COM linear velocity vector in the transverse plane.

For all participants we calculated the average range of $\vec{H}_{WB}$ across trials in each plane of motion for the continuation stride. We grouped segments together into six segment groups: head/trunk, pelvis, right arm (upper arm, forearm), left arm, right leg (thigh, shank, foot), and left leg. For each segment group, we calculated positive and negative contributions to $\vec{H}_{WB}$ as a percentage according to:

$$\%\,Positive\ Contribution_{seg} = 100 * \frac{\int(\vec{H}_{seg}>0)}{\int(\sum_{seg}^{6}(\vec{H}_{seg}>0))} \quad (3)$$

$$\%\,Negative\ Contribution_{seg} = 100 * \frac{\int(\vec{H}_{seg}<0)}{\int(\sum_{seg}^{6}(\vec{H}_{seg}<0))} \quad (4)$$

where $\vec{H}_{seg}$ is the angular momentum of each segment group (Figure 2.2).

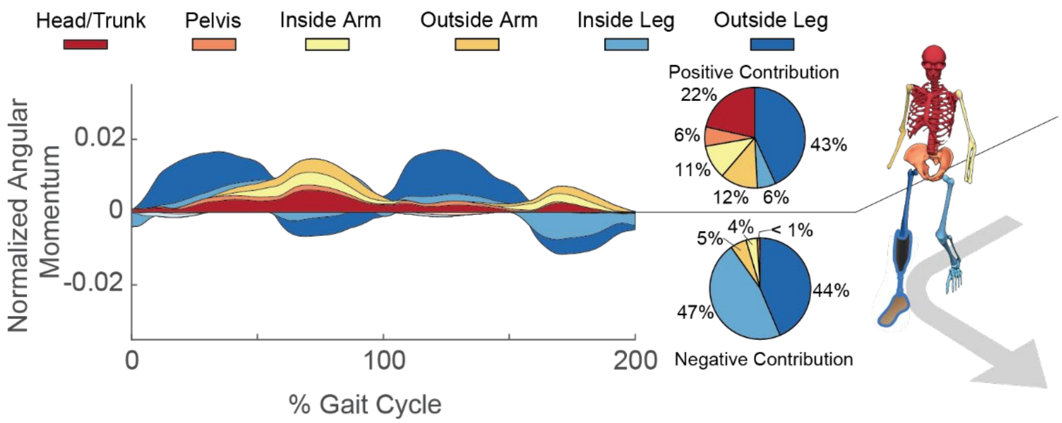

Figure 2.2. Visualization of the positive and negative segment contributions to positive and negative whole-body angular momentum, respectively.

Statistical Analysis

The primary dependent measures were the range of $\vec{H}_{WB}$ and positive and negative $\vec{H}_{seg}$ contributions during the continuation stride. Secondarily, as speed can affect segment momenta generation (Bennett et al., 2010; D'Andrea et al., 2014; Silverman and Neptune, 2011), we compared walking speed between turns using a series of paired t-tests and between groups using a series of independent t-tests. Turns were defined as inside and outside turns, indicating where the prosthesis, for participants with TTA, and the non-dominant leg, for controls, was located during the turn. We tested for differences in the range of $\vec{H}_{WB}$ using a linear mixed model for each plane (frontal, sagittal, transverse) with Group (TTA, controls), Turn (Inside, Outside) and the interaction (Group×Turn) as fixed effects and participants as a random effect with random intercepts. The same linear mixed model was used to compare positive and negative segment contributions to $\vec{H}_{WB}$ for each segment group. All linear mixed models converged successfully, and model results are shown in the appendix (Table A.1-Table A.3). We used contrasts to make pairwise comparisons and applied a Sidak correction for multiple comparisons. All statistical analyses were performed using R 3.6.3 (R Core Team, Vienna, Austria), with $\alpha = 0.05$.

2.4 Results

Demographics and Walking Speed

There were no significant differences between groups in age ($p = 0.55$), height ($p = 0.50$), or body mass ($p = 0.19$). There were no significant differences in walking speed between inside (1.05 ± 0.07 m/s) and outside (1.05 ± 0.08 m/s) turns for controls ($p = 0.866$), between groups ($p = 0.264$), or between inside (0.99 ± 0.14 m/s) and outside (1.03 ± 0.15 m/s) turns for participants with TTA ($p = 0.097$).

Whole-body Angular Momentum

During the continuation stride, there were significant Group×Turn interactions for the range of whole-body angular momentum ($\vec{H}_{WB}$) in all planes of motion ($p < 0.003$). Specifically, participants with TTA had a greater frontal and sagittal range of $\vec{H}_{WB}$ compared to controls during inside turns ($p < 0.047$; Figure 2.3; Table A4). For participants with TTA, the range of $\vec{H}_{WB}$ was greater for inside turns compared to outside turns in all planes of motion ($p < 0.003$), while this comparison was significant for controls only in the sagittal plane ($p < 0.001$).

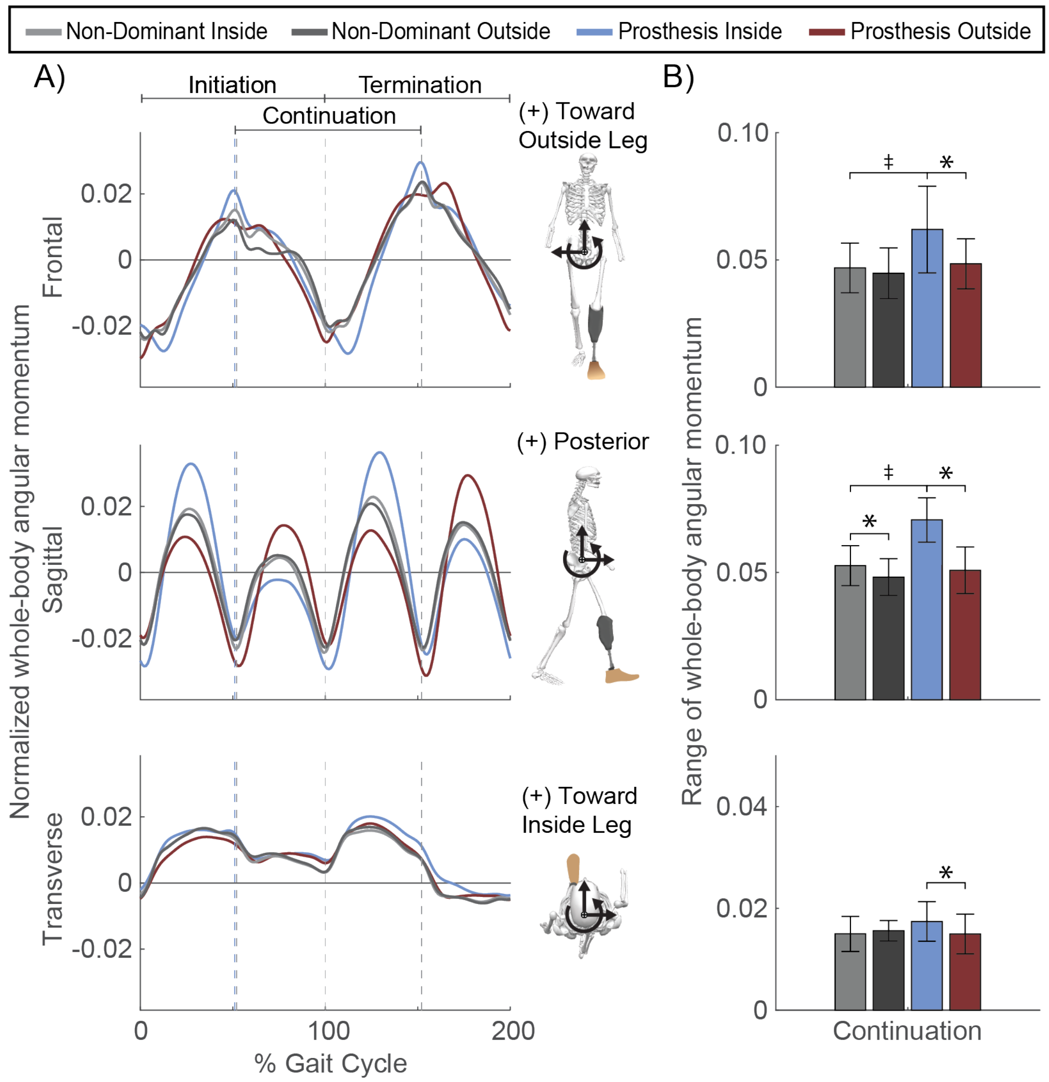

Figure 2.3. A) Dimensionless whole-body angular momentum, $\vec{H}_{WB}\left(\frac{kg\cdot m^2}{s\cdot\left(kg\cdot m\cdot\frac{m}{s}\right)}\right)$, trajectories during turning for all strides (initiation, continuation, termination) in the three planes of motion. The dashed vertical lines correspond to the beginning or end of a specific turning stride. **B)** Range of whole-body angular momentum, $\vec{H}_{WB}\left(\frac{kg\cdot m^2}{s\cdot\left(kg\cdot m\cdot\frac{m}{s}\right)}\right)$, over the gait cycle of the continuation stride. Significant differences between groups are shown with (‡) and between turns with (*).

Head/Trunk and Pelvis Contributions to Whole-body Angular Momentum

In the frontal plane, there was a significant Group×Turn interaction for overall head/trunk contribution to $\vec{H}_{WB}$ (p < 0.001). Controls had greater positive contribution during inside turns compared to outside turns (p < 0.001), while participants with TTA had greater positive contribution during outside turns compared to inside turns (p < 0.001; Figure 2.4; Table A5). For negative contribution, participants with TTA had greater contribution to $\vec{H}_{WB}$ during outside turns compared to inside turns (p < 0.004).

In the sagittal plane, there was a significant Group×Turn interaction for the positive head/trunk contribution (p < 0.001). Specifically, participants with TTA had greater positive contribution to $\vec{H}_{WB}$ during inside turns compared to controls (p < 0.036) and to outside turns (p < 0.001). The negative contribution across groups was greater during inside turns compared to outside turns (p = 0.003).

In the transverse plane, there was a significant Group×Turn interaction only for the negative head/trunk contribution (p = 0.017). While controls had greater negative contribution during inside turns compared to outside turns (p = 0.004), the magnitude of contribution for either turn was close to zero (Figure 2.4).

The positive transverse-plane pelvis contribution to $\vec{H}_{WB}$ had a significant Group×Turn interaction (p < 0.001) where participants with TTA had greater contribution during outside turns compared to inside turns (p <0.001). All other significant pelvis contributions were close to zero (Figure 2.4; Figure 2.5) and are presented in the appendix (Table A.4-Table A.5).

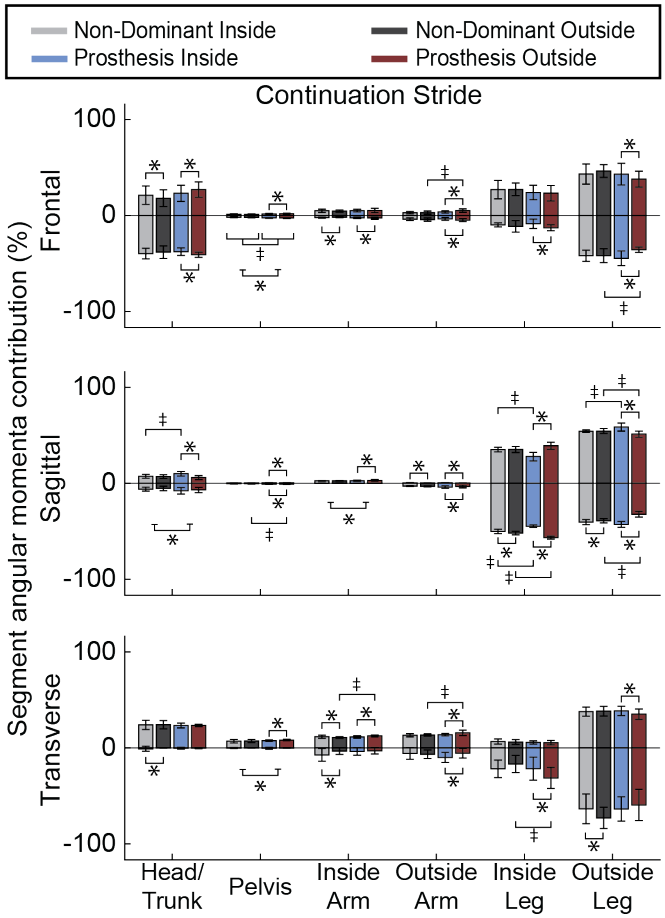

Figure 2.4. Percent contribution of each segment group as a percent of whole-body angular momentum, $\overrightarrow{\boldsymbol{H}}_{\boldsymbol{WB}}\left(\frac{\boldsymbol{kg \cdot m^2}}{\boldsymbol{s \cdot \left(kg \cdot m \cdot \frac{m}{s}\right)}}\right)$, for the continuation stride in the three planes of motion. Positive segment contribution is the percent of positive whole-body angular momentum ($\boldsymbol{Positive}\ \overrightarrow{\boldsymbol{H}}_{\boldsymbol{WB}}$) a segment group contributes. Negative segment contribution is the percent of negative whole-body angular momentum ($\boldsymbol{Negative}\ \overrightarrow{\boldsymbol{H}}_{\boldsymbol{WB}}$) a segment group contributes. Significant differences between groups are shown with (‡) and between turns with (*).

Arm Contribution to Whole-body Angular Momentum

There was a significant Group×Turn interaction for the positive contribution of the inside and outside arms ($p < 0.001$) in the transverse plane. Participants with TTA had greater contribution during outside turns compared to controls ($p < 0.025$) and inside turns ($p < 0.001$). Controls also had greater inside arm contribution during inside turns compared to outside turns (p

< 0.001). For the outside arm negative contribution, participants with TTA had greater contribution during inside turns compared to outside turns ($p < 0.001$). All other significant arm contributions with magnitudes close to zero (Figure 2.4; Figure 2.5) are presented in the appendix (Table A.4-Table A.5).

Leg Contribution to Whole-body Angular Momentum

In the frontal plane, there was a significant Group×Turn interaction for the inside leg negative contribution to $\vec{H}_{WB}$ ($p = 0.028$), which was greater for participants with TTA during inside turns compared to outside turns ($p < 0.001$). For the outside leg, both negative and positive contributions had significant Group×Turn interactions ($p < 0.001$). Both positive and negative contributions were greater during inside turns compared to outside turns for the participants with TTA ($p < 0.001$). Furthermore, the outside leg negative contribution was greater for controls compared to participants with TTA during outside turns ($p = 0.046$).

There were significant Group×Turn interactions for overall sagittal-plane inside and outside leg contributions ($p < 0.001$; Figure 2.4). Both inside and outside legs shared significant differences in positive and negative contribution between turns for participants with TTA, in positive contribution between groups for inside turns, and in negative contribution between turns for controls. However, the comparisons within each leg were opposite between legs. For example, participants with TTA had less positive inside leg contribution compared to controls ($p < 0.001$) while the positive outside leg contribution was greater than controls ($p = 0.008$) during inside turns. Similar for controls, the inside leg negative contribution during inside turns was less compared to outside turns ($p < 0.001$), while the outside leg negative contribution was greater during inside turns compared to outside turns ($p = 0.003$). For participants with TTA, the inside leg had greater overall contribution during outside turns compared to inside turns ($p < 0.001$), while the outside leg had the opposite relationship ($p < 0.001$). In addition, participants with TTA had greater inside leg negative contribution compared to controls during outside turns ($p < 0.001$) while controls had greater negative contribution compared to participants with TTA during inside turns ($p < 0.001$). Furthermore, controls had greater overall outside leg contributions compared to participants with TTA during outside turns ($p < 0.049$).

In the transverse plane, the inside leg negative contribution, and outside leg overall contributions had significant Group×Turn interactions. For the inside leg, participants with TTA had greater negative contribution during outside turns compared to controls ($p = 0.007$) and inside turns ($p < 0.001$). For the outside leg, controls had greater negative contribution during outside turns compared to inside turns ($p < 0.001$). For positive outside leg contribution, participants with TTA had greater contribution during inside turns compared to outside turns ($p < 0.001$).

2.5 Discussion

This study quantified differences in whole-body angular momentum ($\vec{H}_{WB}$) in people with TTA compared to non-amputees and determined how the regulation of $\vec{H}_{WB}$ was affected by the location of the prosthesis with respect to the turn. In the frontal plane, our hypothesis that people with TTA would have a greater range of $\vec{H}_{WB}$ than controls was supported only during inside turns (Figure 2.3). In addition, as found in circle turning (Pew et al., 2019; Shell et al., 2017), we found participants with a greater range of frontal $\vec{H}_{WB}$ during inside turns than outside turns (Figure 2.3), supporting our hypothesis. Participants with TTA also had greater frontal-plane head/trunk contribution during outside turns compared to inside turns, in agreement with our hypothesis. This increase was likely used to compensate for the reduced contribution of the amputated leg, which was likely due to the smaller vertical GRF and altered inertial properties of the prosthesis. As the range of frontal-plane $\vec{H}_{WB}$ correlates with lower clinical balance scores and other laboratory-based balance measures in people post-stroke (Vistamehr et al., 2016), our findings suggest that people with TTA may be at a greater risk of losing balance when turning with the amputated leg on the inside of a turn.

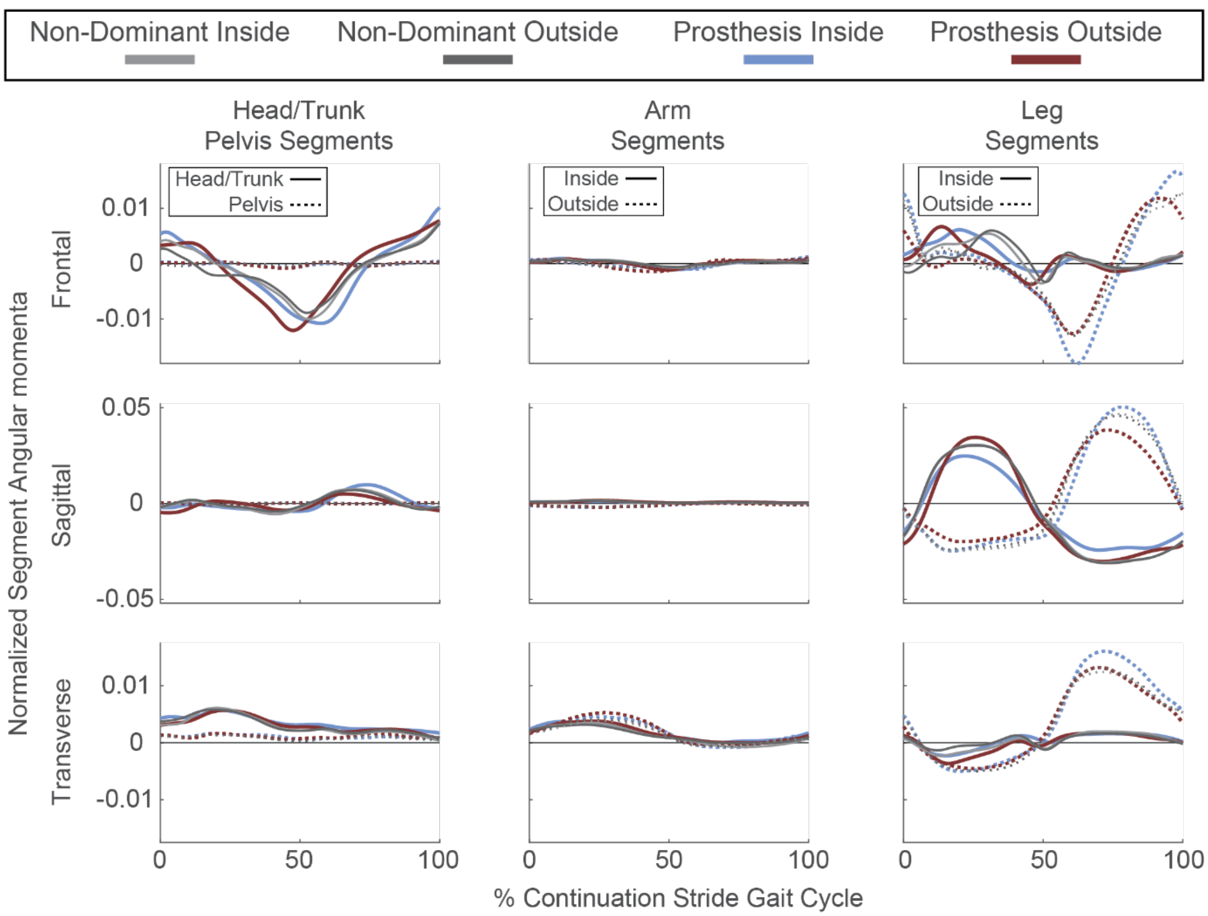

Figure 2.5. Dimensionless segment angular momentum $\left(\frac{\boldsymbol{kg\cdot m^2}}{\boldsymbol{s\cdot\left(kg\cdot m\cdot\frac{m}{s}\right)}}\right)$ during the continuation stride gait cycle in the three planes of motion for the head/trunk, pelvis, inside arm and leg, and outside arm and leg. Segment momenta for participants with TTA turning with the prosthesis on the inside and outside of the turn is shown in blue and red, respectively. Segment momenta for controls turning with the non-dominant leg on the inside and outside of the turn are show in light gray and dark gray, respectively.

Also partially supporting our hypothesis, people with TTA had a greater range of sagittal $\vec{H}_{WB}$ compared to controls during inside turns (Figure 2.3). Control participants also had a greater range of sagittal $\vec{H}_{WB}$ during inside compared to outside turns (Figure 2.3). The difference between turns for controls is likely due to the, greater propulsion of the dominant leg compared to the non-dominant leg when on the outside of a turn (Strike and Taylor, 2009). In support of our hypothesis, people with TTA had differences in the range of sagittal $\vec{H}_{WB}$ between turns (Figure 2.3). The greater range of sagittal $\vec{H}_{WB}$ during inside turns is likely due to body segment compensations for the reduced propulsion and altered inertial properties of the amputated leg. Consistent with the greater intact leg propulsion on either side of the turn (Ventura et al., 2011), we found greater intact leg positive and negative contributions to sagittal $\vec{H}_{WB}$ compared to the amputated leg regardless of turn direction (**Figure 2.4**), further supporting our hypothesis. Participants with TTA also had greater head/trunk contribution to positive sagittal $\vec{H}_{WB}$ during inside turns compared to controls and outside turns (**Figure 2.4**), which did not support our hypothesis. As the continuation stride is taken with the intact leg during inside turns, the positive head/trunk contribution occurs when the intact leg is in the swing phase and the amputated leg is in the stance phase (Figure 2.5). During the stance phase in non-amputee gait, the soleus muscle transfers energy to the trunk (Neptune et al., 2001) and contributes to negative sagittal $\vec{H}_{WB}$ (Neptune and McGowan, 2011). Thus, the inability of the passive prosthesis on the inside of the turn to fully replicate the soleus function likely resulted in more positive head/trunk contribution to a greater range of sagittal $\vec{H}_{WB}$. This result is further supported by prior findings where people with TTA reduced the range of sagittal $\vec{H}_{WB}$ when walking on level ground (D'Andrea et al., 2014) and down an incline (Pickle et al., 2016) when using a powered prosthesis that generates positive ankle power, similar to the soleus. Powered prostheses may assist in reducing the head/trunk contribution to positive sagittal $\vec{H}$ during inside turns and therefore may improve balance control in the sagittal plane.

In the transverse plane, our hypothesis was not supported as there were no differences in the range of $\vec{H}_{WB}$ between groups. For participants with TTA, the range of $\vec{H}_{WB}$ was greater during inside turns compared to outside turns (Figure 2.2), which can be attributed to the greater intact leg contribution to positive transverse $\vec{H}_{WB}$ (Figure 2.4), supporting our hypotheses. This greater intact leg contribution occurs from the end of the stance phase to end of the continuation gait cycle (Figure 2.5). Thus, the greater range of $\vec{H}_{WB}$ is likely the result of the greater propulsion (Ventura et al., 2011), stride length (Segal et al., 2011), and altered inertial properties of intact leg compared to the amputated leg. In addition, during outside turns participants with TTA had greater arm contributions toward the turn (i.e., positive) (Figure 2.4), likely to counteract the greater intact leg contribution away from the turn (i.e., negative) and resulting in a range of $\vec{H}_{WB}$ similar to controls. As transverse-plane $\vec{H}_{WB}$ has to be generated toward the inside of the turn (Nolasco et al., 2019), these different segment interactions between inside and outside turns suggest altered strategies between turn directions for people with TTA.

In this study, we chose to investigate balance during turning in people with TTA using their own passive prosthesis as it is most representative of how they turn in daily life. These devices likely had differing inertial and mechanical properties (Table 2.1), both of which may influence

$\vec{H}_{WB}$. While we accounted for the altered inertial properties of the amputated leg using a general model which estimates inertial properties based on body mass and segment length (Ferris et al., 2017b) we acknowledge that the actual values likely differed from this estimation. Mechanical properties such as sagittal- and transverse-plane compliance or energy-return can also influence $\vec{H}_{WB}$. Prior work found that people with TTA had reduced vertical GRF impulse and range of frontal $\vec{H}_{WB}$ during inside circle turns with a more compliant prosthetic foot (Shell et al., 2017). Another study found that decreasing the stiffness of a torsion adapter increased frontal $\vec{H}_{WB}$ range but did not affect GRF impulses during inside circle turns (Pew et al., 2019). In addition, powered prostheses provide active ankle power and have altered inertial properties due to the increased mass of powered components. Together these factors affect $\vec{H}_{WB}$ by altering the contributions of the trunk and amputated leg (Pickle et al., 2019). While the devices used in this study likely had a variety of compliances, no participants used torsional adaptors or powered prostheses.

One potential limitation in the generalizability of our findings is the relatively small sample size, which may limit the ability to detect differences. Thus, we have provided confidence intervals to assess the magnitude of the differences for all comparisons (Table A1, A2). We also focused on step-turns which may not generalize to all turns made in daily life (e.g., spin-turns). However, our participants predominately chose step-turns which is consistent with a previous study (Golyski and Hendershot, 2018). Another limitation is the homogenous cohort of highly mobile individuals classified as K3/K4 level on the Medicare Functional Classification scale. High-functioning participants are more likely to perform successful 90-degree turns with little risk of losing balance compared to less mobile individuals. For example, individuals with lower activity have a lower balance confidence (Mandel et al., 2016) and greater fear of falling (Miller et al., 2002), which may affect turning in daily life. Therefore, we may expect greater differences in $\vec{H}_{WB}$ regulation during turning and between turns from these individuals. However, understanding how highly mobile individuals perform turns provides insight on strategies that should be developed in individuals with lower balance confidence.

2.6 Conclusion

This study examined how people with TTA maintain dynamic balance during 90-degree turns. We found that the regulation of $\vec{H}_{WB}$ during 90-degree turns depends on the side of the turn the prosthesis is on. During outside turns, people with TTA are able to alter their head/trunk, leg, and arm contributions to $\vec{H}_{WB}$ to compensate for the inability of a passive prosthesis to replicate the biological function of an ankle joint. In contrast, we found that during inside turns, people with TTA do not normalize $\vec{H}_{WB}$. Future work should explore how practicing turns during rehabilitation may improve balance and functional mobility in daily life for people with an amputation of varying mobility levels.

CHAPTER 3. Transtibial Prosthetic Alignment Has Small Effects on Whole Body Angular Momentum During Functional Tasks

3.1 Abstract

Due to the loss of ankle function, many people with a transtibial amputation (TTA) have difficulty maintaining balance during functional tasks. Prosthetic alignment affects ground reaction forces and center of pressure, two factors important for maintaining balance. This effect suggests that alignment may play a role in how people with TTA maintain dynamic balance. As such, we explored dynamic balance, measured as whole-body angular momentum ($\vec{H}_{WB}$), in people with and without TTA, during several functional tasks including walking and transitioning in and out of a chair. Participants with TTA completed all tasks with their prescribed alignment and six shifted alignments, including ±10 mm anterior/posterior, medial/lateral, and ±20 mm in the vertical direction. Alignment had little effect on the range of $\vec{H}_{WB}$ or the number of balance adjustments depending on the task and plane of motion. Participants with TTA had a greater range of $\vec{H}_{WB}$ during walking, sit-to-stand, and stand-to-sit compared to controls. Participants with TTA made fewer adjustments to $\vec{H}_{WB}$ compared to controls during sit-to-stand and stand-to-sit while making more adjustments during walk-to-sit. These results suggest that highly mobile individuals with TTA can adjust to generate and regulate similar levels of $\vec{H}_{WB}$ despite small and acute changes in alignment during various activities of daily living. Our findings also suggest that highly functional people with TTA have reduced frontal plane balance control and require greater generation of

transverse and sagittal plane $\vec{H}_{WB}$ compared to non-amputees during sit-to-stand and stand-to-sit. These results may be useful for understanding the planes of motion where regulation of $\vec{H}_{WB}$ should be improved for seat transfers in people with TTA. Future work should also investigate whether these effects extend to lower functioning individuals.

3.2 Introduction

To achieve functional independence in daily life, it is necessary to effectively control balance. Balance control can be challenging for people with lower limb amputation (Ku et al., 2014) and likely leads to their high risk of falling (Hunter et al., 2017; Kulkarni et al., 1996) and reports of falling during activities of daily living (Chihuri and Wong, 2018). For people with a unilateral transtibial amputation (TTA), deficits in neuromuscular function result in altered walking strategies compared to non-amputees (Fey et al., 2010; Sanderson and Martin, 1997) and a limited ability to make rapid balance adjustments (Curtze et al., 2012; Ku et al., 2014; Olensek et al., 2021). To accommodate the loss of ankle function, people with TTA rely on the intact leg to maintain balance during standing (Ku et al., 2014) and perturbation recovery (Curtze et al., 2012).

The ability to control balance on the prosthetic leg may be affected by prosthetic alignment. Specifically, the orientation of the prosthetic foot relative to the socket can affect how forces are transferred through the body by altering the location of the center of pressure (COP) and direction of the ground reaction force (GRF). For example, differences in GRF distance (i.e., moment arm) from the knee joint due to changes in sagittal rotation (i.e., plantarflexion/dorsiflexion) (Blumentritt et al., 1999) and translation (i.e., anterior/posterior) of the prosthetic foot altered knee muscle activity during standing (Blumentritt et al., 1999) and the stance phase of walking

(Blumentritt et al., 2001) compared to a prescribed alignment. In addition, dorsiflexing and lengthening the prosthesis affected the reaction latency to standing platform translations (Paráková, 2009). Similarly, alignment affects joint kinetics during walking (Jonkergouw et al., 2016), and COP location (Nolasco et al., 2020), GRF (Nolasco et al., 2020), and muscle activity (Wagner et al., 2020) during sit-to-stand. While these studies suggest that alignment will affect balance control, those that have explicitly measured balance during standing with eyes closed (Isakov et al., 1994) and center of gravity target matching during standing (Kolarova et al., 2013) did not report differences. However, lack of differences between alignments in these studies may be due to the focus on standing tasks, which may require smaller adjustments to maintain balance compared to dynamic movement tasks, or the choice of balance measure.

One measure of dynamic balance that incorporates COP, GRF, and kinematics is whole-body angular momentum. Whole-body angular momentum ($\vec{H}_{WB}$) is altered to maintain and/or restore dynamic balance through muscle activation that accelerates body segments and produces GRFs. Adjustments to $\vec{H}_{WB}$ can be described by the net external moment about the body center of mass (COM), which equals the time rate of change of whole-body angular momentum. As such, changes in GRF and moment arm (i.e., COM to COP distance) can affect how whole-body angular momentum is controlled and/or regulated. For example, the reduced prosthetic limb vertical GRF during walking is associated with a greater frontal range of $\vec{H}_{WB}$ (Silverman and Neptune, 2011). This greater range of $\vec{H}_{WB}$ suggests people with TTA have a greater risk of losing balance as a larger external moment is required to restore large values of $\vec{H}_{WB}$. Though not quantified previously, generating and arresting $\vec{H}_{WB}$ may also be challenging during other functional tasks such as sit-to-stand where people with TTA have greater COM sway velocity compared to non-amputees (Ozyurek et al., 2014). Not to mention, the reduced COP control with the amputated leg compared to the intact leg during standing (Ku et al., 2014; Rusaw, 2019) and sit-to-stand (Nolasco et al., 2020) in people with TTA may require altered generation and regulation of $\vec{H}_{WB}$ in the sagittal plane. Due to the generation of sagittal plane $\vec{H}_{WB}$ and regulation of medial/lateral $\vec{H}_{WB}$ for balance required during seat transfers, controlling $\vec{H}_{WB}$ may be more difficult for people with TTA. Furthermore, how prosthetic alignment influences generation and regulation of $\vec{H}_{WB}$ during these functional tasks, remains unclear.

The purpose of this study was to investigate how acute changes in prosthetic alignment affect $\vec{H}_{WB}$ during functional tasks of daily living including walking and transitioning in and out of a chair. To provide context for the findings, we also compared $\vec{H}_{WB}$ between people with and without a unilateral transtibial amputation. Due to the relationship between the external moment about the COM and $\vec{H}_{WB}$, we expected medial/lateral and tall/short alignments to affect frontal and transverse $\vec{H}_{WB}$ and anterior/posterior alignments to affect sagittal plane $\vec{H}_{WB}$. Secondarily, to gain additional insight into how individuals controlled their balance their balance *throughout* the movement, we measured the number of $\vec{H}_{WB}$ direction changes for each task.

3.3 Methods

Participants

We recruited 10 people with a unilateral transtibial amputation (TTA) through a local orthotics and prosthetics clinic (Table 3.1). All participants were screened to ensure they did not have a history of neurological or cardiovascular disease, uncorrected vision problems, take medication that affected their ability to walk, or any mental capacity impairment that would negatively affect verbal communication. Participants were excluded if they were not able to walk independently for at least 10 minutes at a time for at least two months prior to data collection. No participants reported any significant injury or pathology of their intact leg. To participate in this institutionally approved study, all participants provided their written informed consent.

Table 3.1. Participant demographics.

TTA	Age (years)	Sex	Weight (kg)	Height (m)	Cause	Prosthetic Foot	K-Level	Control	Age (years)	Sex	Weight (kg)	Height (m)
1	57	M	94.08	1.85	Dysvascular	College Park Velocity	K3	1	52	M	69.63	1.69
2	25	F	63.94	1.66	Cancer	Endolite Elite 2	K4	2	21	F	66.00	1.70
3	56	M	99.56	1.70	Trauma	Ossur LP Variflex with EVO	K4	3	42	M	87.54	1.79
4	28	M	84.12	1.87	Trauma	Freedom Innovations Agilix	K4	4	27	M	97.07	1.87
5	58	M	101.84	1.78	Trauma	College Park Velocity	K4	5	48	M	116.12	1.91
6	63	M	121.94	1.81	Trauma	College Park TruStep	K3	6	62	M	84.37	1.81
7	65	M	90.43	1.67	Trauma	Ottobock Triton 1C66	K3	7	61	M	79.83	1.78
8	31	M	93.40	1.82	Trauma	Freedom Innovations Maverick	K3	8	29	M	79.38	1.80
9	61	M	115.55	1.88	Trauma	Fillaeur All Pro	K4	9	53	M	108.41	1.78
10	56	M	109.61	1.82	Trauma	Freedom Innovations Agilix	K3	10	54	M	99.79	1.97
Mean ± SD	50 ± 16	-	97.45 ± 16.59	1.79 ± 0.08	-	-	-	Mean ± SD	45 ± 15	-	88.81 ± 16.32	1.81 ± 0.09

Experimental Protocol

Each participant's initial prosthetic alignment condition was prescribed by their certified prosthetist. Translational alignment changes relative to each participant's prescribed alignment were performed by a certified prosthetist in the anterior-posterior, medial-lateral, and vertical directions. For the anterior, posterior, medial, and lateral conditions, the prosthetist shifted the foot relative to the socket by 10 mm. For the tall and short conditions, the prosthetist changed the height of the prosthesis by 20 mm. The alignment changes were verified using a flexible measuring tape by the prosthetist. After the prescribed condition, the remaining alignment conditions were completed randomly, in pairs (i.e., anterior and posterior, medial and lateral, tall and short) and participants were not informed of the change that had been made. For each alignment condition, participants were instructed to complete a series of functional tasks including walking over level ground at a self-selected speed, five sit-to-stand and stand-to-sit trials at a self-selected pace, and a single Timed-up-and-Go (TUG) assessment (Schoppen et al., 1999) (Figure 3.1). For all tasks requiring a chair, participants were seated on a backless and armless stool that was height-adjusted so that their hips and knees were at approximately 90 degrees and their feet placed hips-width apart.

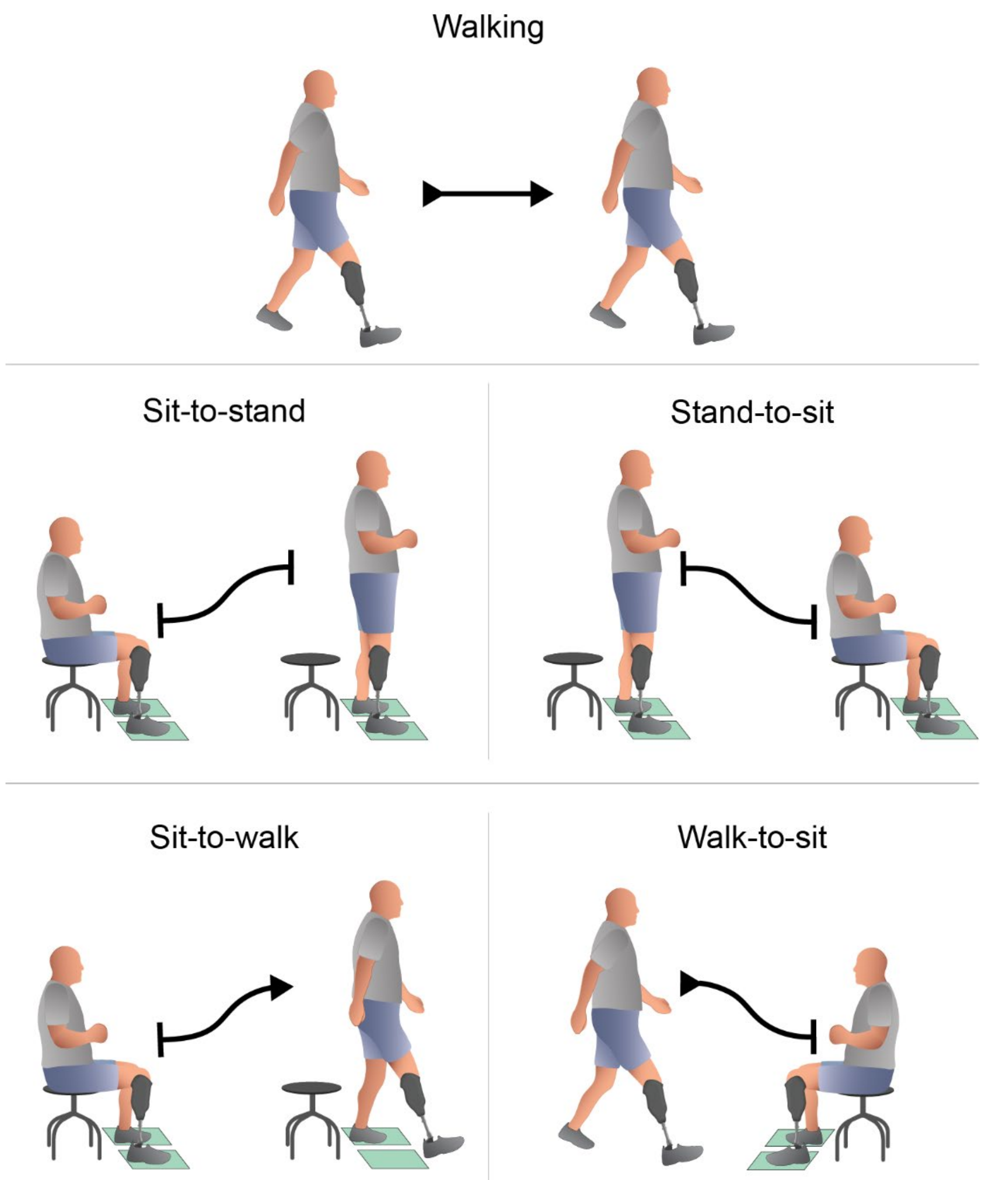

Figure 3.1. Visual representation of the tasks analyzed in this study: walking (amputated side heel strike – heel strike), sit-to-stand, stand-to-sit, sit-to-walk, and walk-to-sit.

A marker set that tracks six degrees-of-freedom at each joint (Collins et al., 2009; Wilken et al., 2012) and a 20-camera motion capture system (Motion Analysis, Santa Rosa, CA) tracked body motion at 120 Hz during all trials. Forty-five reflective markers were placed on body landmarks including the C7, sternum, xiphoid process, T10, and bilaterally on the acromion, iliac crest, anterior and posterior superior iliac spines, greater trochanter, lateral and medial tibial epicondyles, lateral and medial malleoli, and 2nd and 5th metatarsals. For the amputated side, markers were placed so they mirrored the intact side. Marker clusters were placed bilaterally on the thighs and shanks while a single marker was placed on each heel for segment tracking.

Data Analysis

Marker trajectories were filtered using a 4th-order low-pass Butterworth filter using a cutoff frequency of 6 Hz. A 9-segment model consisting of the head, trunk , pelvis, thighs, shanks, and feet was developed in Visual3D (C-Motion, Germantown, MA) with segment inertial properties based on previous work (Dempster, 1955; Hanvan, 1964). The inertial properties for the prosthesis were adjusted according to (Ferris et al., 2017b). Whole-body COM location was calculated as the weighted sum of all 9-segments.

The TUG assessment was divided into sit-to-walk and walk-to-sit tasks occurring at the beginning and end of the assessment, respectively. Analyzed tasks included walking, sit-to-stand, stand-to-sit, sit-to-walk, walk-to-sit. We measured the time taken to complete each task from the start to end events defined in Table 3.2.

Table 3.2. Protocol and event definitions for each task.

Task	Protocol	Event definition	
		Start	**End**
Walk	Self-selected speed over 8-m level ground walkway	Amputated side heel strike	Subsequent amputated side heel strike
Sit-to-stand	Self-selected pace	Initiation of positive horizontal body COM velocity (Jones et al., 2016; Nolasco et al., 2020; Wagner et al., 2020)	Termination of vertical and horizontal body COM velocity (Jones et al., 2016; Nolasco et al., 2020; Wagner et al., 2020)
Stand-to-sit	Self-selected pace	Initiation of sagittal trunk angular movement (angular acceleration crosses zero)	Moment when posterior body COM velocity reaches zero
Sit-to-walk	At a comfortable pace, stand up from a chair and walk toward a cone 3-meters away (beginning of TUG)	Initiation of positive horizontal body COM velocity (Jones et al., 2016; Nolasco et al., 2020; Wagner et al., 2020)	First initial contact after gait initiation (Jones et al., 2016; Kerr et al., 2004)
Walk-to-sit	At a comfortable pace, walk to a chair and sit down (end of TUG)	Moment when foot (2nd metatarsal) is within 61 cm of the chair (last component of the component-TUG (Clemens et al., 2018a))	Moment when posterior body COM velocity reaches zero

Whole-body angular momentum ($\vec{H}_{WB}$) was calculated for all tasks between start and end events as:

$$\vec{H}_i = [(\vec{r}_i^{COM} - \vec{r}_{body}^{COM}) \times m_i(\vec{v}_i^{COM} - \vec{v}_{body}^{COM}) + I_i\vec{\omega}_i] \quad (1)$$

$$\vec{H}_{WB} = \sum_{i=1}^{n} \vec{H}_i \quad (2)$$

where $\vec{r}_i^{COM}$ and $\vec{v}_i^{COM}$ are the position and velocity of the i-th segment's COM, $\vec{r}_{body}^{COM}$ and $\vec{v}_{body}^{COM}$ are the position and velocity of the whole-body COM, m_i is the segment mass (Dempster, 1955), I_i is the segment moment of inertia, $\vec{\omega}_i$ is the segment angular velocity, and n is the total number of segments. Because the walk-to-sit task included a turn, we rotated the angular momentum reference frame from the laboratory reference frame to the pelvis reference frame about the vertical axis. This rotation ensured that angular momentum would be interpreted in the anatomical planes of motion for all tasks. For all other tasks, angular momentum was defined in the laboratory reference frame. All angular momenta were normalized by dividing by body mass (kg), and height (m). We then calculated the peak-peak range of $\vec{H}_{WB}$ in all planes of motion from 0-100% of each task.

The range of $\vec{H}_{WB}$ describes the extremes of $\vec{H}_{WB}$ generated during motion, and thus represents how $\vec{H}_{WB}$ is generated across a full movement at steady-state. As it represents the extrema of the movement, it does not describe how $\vec{H}_{WB}$ is controlled continuously throughout the task. In addition, the idea that a smaller range of $\vec{H}_{WB}$ suggests better balance regulation likely does not apply to tasks that require active $\vec{H}_{WB}$ generation (e.g., sit-to-stand). To better understand how participants actively controlled $\vec{H}_{WB}$ throughout each task, we quantified the number of adjustments in $\vec{H}_{WB}$ *(nAdj)* made throughout the movement. This was calculated as the number of zero-crossings of the time-derivative of $\vec{H}_{WB}$ in each plane of motion.

Statistical Analysis

For each task, we tested for differences in the task duration, range of $\vec{H}_{WB}$, and *nAdj* between alignment conditions using separate linear mixed models for each plane of motion with alignment as the fixed effect and participants as a random effect. Significant main effects were explored by comparing each alignment condition to the prescribed condition, using estimated marginal means with specific contrasts and a Sidak correction for multiplicity. In addition, we compared all outcome variables between controls and people with TTA (prescribed condition only) using a series of Welch's t-test. All statistical analyses were performed using R 3.6.3 (R Core Team, Vienna, Austria), with a significance level of $\alpha = 0.05$.

3.4 Results

Time to complete task

There were no differences in the task duration between alignments for walking, sit-to-walk, or walk-to-sit ($p > 0.062$; Table 3.3). Participants completed the sit-to-stand task significantly slower with their prescribed alignment compared to all other alignments ($p < 0.004$). Similarly, participants completed stand-to-sit slower with their prescribed alignment compared to all conditions, except the posterior alignment ($p = 0.059$). There were no significant differences in task duration between groups for any task ($p > 0.075$; Appendix B).

Table 3.3. Average (standard deviation) time taken to complete a task.

	Condition							
Task	Prescribed	Anterior	Posterior	Medial	Lateral	Tall	Short	Controls
Walk	1.13 (0.06)	1.13 (0.05)	1.12 (0.03)	1.13 (0.05)	1.13 (0.06)	1.16 (0.06)	1.13 (0.06)	1.15 (0.09)
Sit-to-stand	2.30 (0.32)	**2.05 (0.18)***	**2.02 (0.18)***	**1.92 (0.22) ***	**2.02 (0.19)***	**1.99 (0.23)***	**1.99 (0.23)***	2.45 (0.26)
Stand-to-sit	1.95 (0.30)	**1.77 (0.17)***	1.79 (0.19)	**1.74 (0.19)***	**1.80 (0.20)***	**1.79 (0.20)***	**1.73 (0.20)***	1.99 (0.39)
Sit-to-walk	1.57 (0.20)	1.56 (0.26)	1.52 (0.29)	1.50 (0.24)	1.41 (0.29)	1.50 (0.21)	1.52 (0.27)	1.46 (0.25)
Walk-to-sit	3.29 (0.61)	3.00 (0.30)	3.17 (0.55)	3.14 (0.46)	3.00 (0.43)	3.29 (0.37)	3.13 (0.54)	2.85 (0.41)

Frontal plane angular momentum

There were significant main effects of alignment for the range of frontal $\vec{H}_{WB}$ during walking at a self-selected speed (p = 0.005; Figure 3.2) and sit-to-stand (p = 0.017). During walking there was a greater range of $\vec{H}_{WB}$ with the tall alignment compared to the prescribed alignment (p = 0.023). During sit-to-stand, there was a greater range of $\vec{H}_{WB}$ with the lateral alignment compared to the prescribed alignment (p = 0.037). There were no significant effects of alignment for the other tasks (p > 0.120). Compared to controls, participants with TTA had a greater range of $\vec{H}_{WB}$in the frontal plane during sit-to-stand (p = 0.002) only.

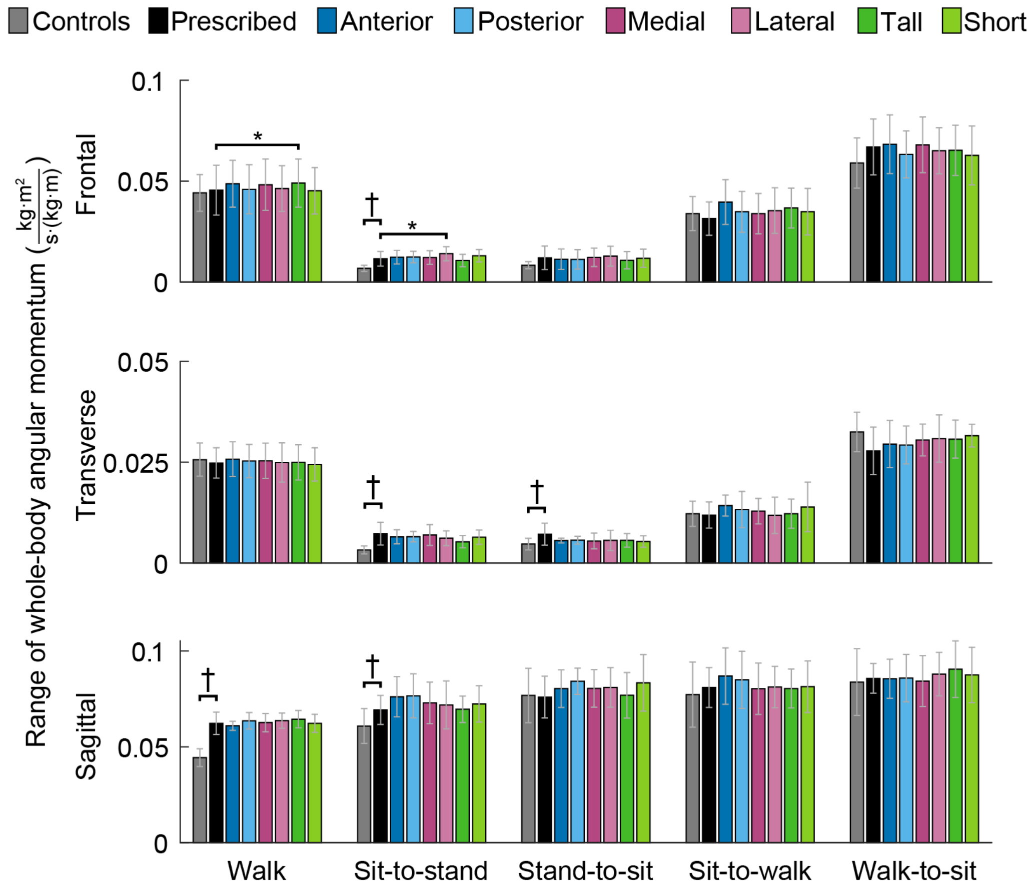

Figure 3.2. Range of $\vec{H}_{WB}$ ($\frac{kg \cdot m^2}{s \cdot (kg \cdot m)}$) normalized to height (m) and weight (kg) in all planes of motion for each task. *Significant differences compared to the prescribed alignment. †Significant differences between controls and people with TTA.

There was no significant effect of alignment on the number of adjustments (*nAdj*) of frontal plane $\vec{H}_{WB}$ for any task ($p > 0.061$; Figure 3.3). Participants with TTA had a smaller *nAdj* compared to controls for the sit-to-stand ($p = 0.015$) and stand-to-sit ($p = 0.002$) tasks.

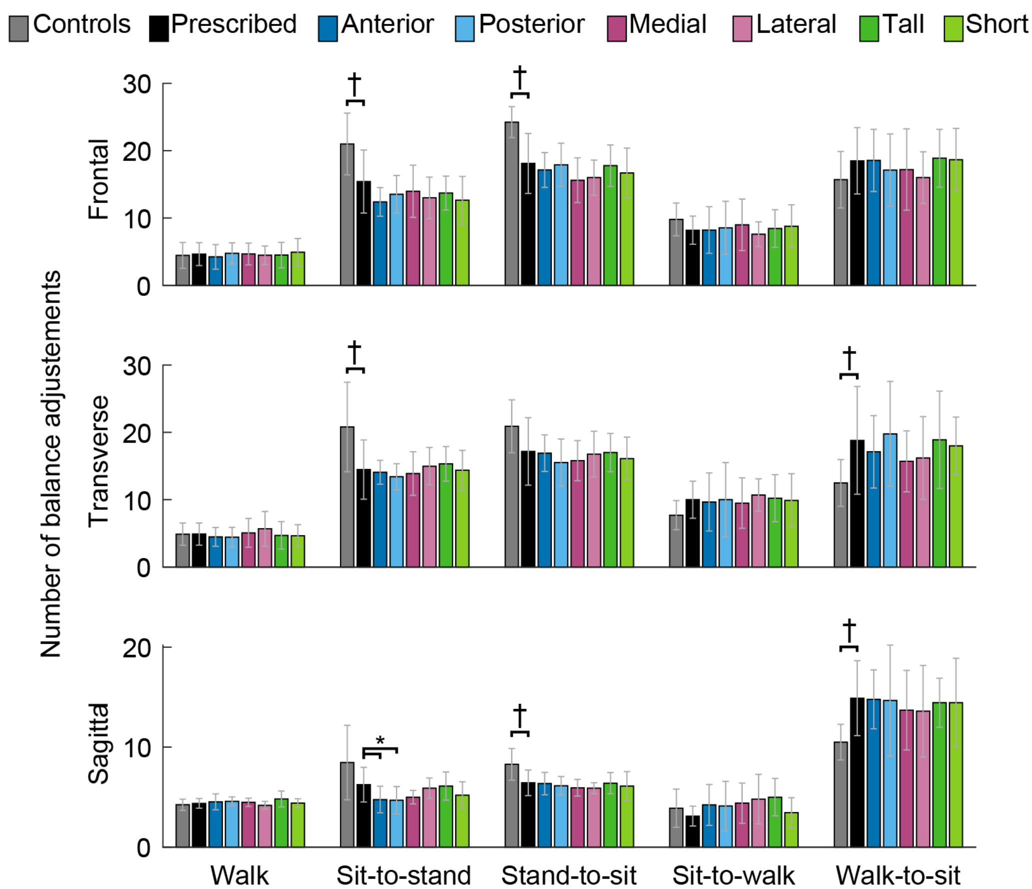

Figure 3.3. Number of zero-crossings in the time-derivative of $\boldsymbol{\vec{H}_{WB}}$ for each task in each plane of motion (*nAdj*). *Significant differences compared to the prescribed alignment. †Significant differences between controls and people with TTA.

Transverse plane angular momentum

The transverse range of $\vec{H}_{WB}$ was not different between alignments for any task ($p > 0.091$; Figure 3.2). There was a significant difference between groups, however. People with TTA had a greater range of $\vec{H}_{WB}$ compared to controls during the sit-to-stand ($p = 0.001$) and stand-to-sit ($p = 0.025$) tasks.

There was no significant effect of alignment on the *nAdj* in the transverse plane for any task ($p > 0.063$; Figure 3.3). Compared to controls, people with TTA had a greater *nAdj* during sit-to-stand ($p = 0.024$) and smaller *nAdj* during walk-to-sit ($p = 0.041$).

Sagittal plane whole body angular momentum

There was no significant effect of alignment on the sagittal range of $\vec{H}_{WB}$ for any task ($p > 0.176$; Figure 3.2). Compared to controls, participants with TTA had a greater range of $\vec{H}_{WB}$ in the sagittal plane during walking ($p < 0.001$) and sit-to-stand ($p = 0.038$).

There was a significant main effect of alignment on the *nAdj* during sit-to-stand ($p = 0.016$; Figure 3.3). Both anterior ($p = 0.048$) and posterior ($p = 0.034$) alignments had a smaller *nAdj* compared to the prescribed alignment. Compared to controls, people with TTA had a smaller *nAdj* during the stand-to-sit ($p = 0.011$) task and a greater *nAdj* during the walk-to-sit ($p = 0.005$) task. As the goals and performance of these tasks are different, we expect significant differences in the angular momentum generation used to complete them successfully for both groups. However, we also expect the ranges to be greater in people with TTA compared to controls for all tasks. While expecting differences in angular momentum between different tasks may be intuitive, quantifying how they differ will provide insight into the amount of balance control required to complete common functional tasks for clinicians. Specifically, as clinicians assess balance ability through timed clinical tests, having knowledge of which tasks may be more challenging for people with TTA during daily life may assist in selecting prosthetic components, setting alignment, and/or developing rehabilitation training to improve balance control during specific tasks.

3.5 Discussion

This study investigated the effect of prosthetic alignment on whole body angular momentum ($\vec{H}_{WB}$) during functional tasks of daily living in people with TTA. Our expectation that frontal and transverse $\vec{H}_{WB}$ would be affected by medial/lateral and short/tall alignment changes was partially supported. Only frontal plane range of $\vec{H}_{WB}$ was affected by tall and lateral alignment changes during the walk and sit-to-stand tasks, respectively. The larger frontal range of $\vec{H}_{WB}$ when walking with a tall prosthesis may be due to greater lateral trunk motion (Gardas and Shah, 2020) and/or greater vertical GRF on the intact leg (Azizan et al., 2018; White et al., 2004) that can be caused by leg length discrepancy. Even though the range of $\vec{H}_{WB}$ was larger in the tall condition, participants made a similar number of adjustments to frontal plane $\vec{H}_{WB}$ (*nAdj*) in each alignment condition. This result suggests that people with TTA did not actively attempt to restore their angular momentum to that observed in the prescribed alignment during walking. During sit-to-stand, the greater range of frontal $\vec{H}_{WB}$ with the lateral alignment may also be a result of greater lateral trunk motion and intact leg reliance. Prior work found that a lateral alignment increased medial GRFs compared to the prescribed alignment (Wagner et al., 2020), due to the wider base of support. This wider base contributes to increased moment arm for the intact side vertical GRF (which also increases), and thus leads to increased $\vec{H}_{WB}$. Importantly, people may be able to counteract this momentum to some degree by increasing lateral trunk lean toward the intact leg.

Our hypothesis that sagittal plane $\vec{H}_{WB}$ would be affected by anterior/posterior alignment changes was also partially supported. While sagittal range of $\vec{H}_{WB}$ did not differ across alignments statistically, the anterior and posterior alignments had greater sagittal range of $\vec{H}_{WB}$ on average compared to the prescribed alignment during sit-to-stand (Figure 3.4). As braking and propulsive forces contribute to sagittal plane $\vec{H}_{WB}$, the altered braking and propulsive forces with anterior and posterior alignments (Nolasco et al., 2020) may result in altered regulation of $\vec{H}_{WB}$. In addition, there were fewer *nAdj* made with the anterior and posterior alignments compared to the prescribed alignment during sit-to-stand. This effect of sagittal plane changes in alignment may be due to the greater sagittal plane $\vec{H}_{WB}$ required during sit-to-stand relative to the other planes of motion (Figure 3.3). While the sagittal plane $\vec{H}_{WB}$ generation was also greater than other planes of motion during stand-to-sit, the lack of alignment effect for this task may be due to controlling balance being less of a priority during this task. As losing balance while sitting down results in completing the task faster or falling into the chair, maintaining balance while sitting down may not be a priority during the task. In addition, the faster completion times with all alignments compared to the prescribed alignment during sit-to-stand and stand-to-sit tasks may also contribute to these differences. While we randomized the order of the alignment changes, the prescribed alignment was performed first for all participants. Thus, participants may have become familiar with performing the tasks over the course of the collection. However, only the times to complete sit-to-stand and stand-to-sit tasks were affected. This may suggest that participants were able to learn a faster strategy to complete the sit-to-stand and stand-to-sit tasks but not for the other tasks. With more acclimation time for different alignments, participants may also adjust their strategies to complete the sit-to-stand and stand-to-sit tasks. As such, our results suggests that the anterior/posterior orientation of the prosthetic foot may affect the strategies used by people with TTA to complete the sit-to-stand and stand-to-sit movements.

Similar to prior literature (Silverman and Neptune, 2011), we found that people with TTA have a greater sagittal range of $\vec{H}_{WB}$ compared to non-amputees during walking. We also found that people with TTA had a greater range of $\vec{H}_{WB}$ compared to non-amputees in all planes of motion during sit-to-stand and in the transverse plane during stand-to-sit. Collectively, this suggests that people with TTA have decreased dynamic balance compared to non-amputees during a range of tasks. This altered balance is likely the result of the greater reliance on the intact leg for force generation and balance control (Curtze et al., 2012; Ku et al., 2014). For sit-to-stand (Agrawal et al., 2011) and stand-to-sit (Agrawal et al., 2016), people with TTA have greater weight bearing on the intact leg compared to non-amputees, which contributes to angular momentum between legs that do not fully counteract each other. As previously mentioned, greater lateral trunk lean toward the intact leg during sit-to-stand (Actis et al., 2018b; Nolasco et al., 2020) may be a compensatory strategy to help counteract the $\vec{H}_{WB}$ generated toward the amputated leg in the frontal plane, thus mitigating the greater range of $\vec{H}_{WB}$. Along with the greater trunk lean, people with TTA also have greater axial trunk rotation during sit-to-stand (Actis et al., 2018b; Nolasco et al., 2020) which may contribute to the greater transverse range of $\vec{H}_{WB}$ found in the current study.

In this study, the range of $\vec{H}_{WB}$ and *nAdj* were quantified to gain insight on different aspects of dynamic balance. The range of $\vec{H}_{WB}$ captures the extrema of the $\vec{H}_{WB}$ generated during a movement task. As this difference between the extrema of $\vec{H}_{WB}$ is regulated to be close to zero during walking (Herr and Popovic, 2008), studies have often used the range of $\vec{H}_{WB}$ to gain insight on the risk of losing balance. However, the range of $\vec{H}_{WB}$ does not offer insight into *how* $\vec{H}_{WB}$ is controlled throughout a movement. In addition, maintaining an average $\vec{H}_{WB}$ at zero is not a goal for non-steady state tasks that require $\vec{H}_{WB}$ generation for task completion like turning (Nolasco et al., 2019) or sit-to-stand. As such, other measures are needed to assess balance control during non-steady tasks. For example, a previous study quantified angular momentum smoothness as the number of peaks in $\vec{H}_{WB}$, where fewer peaks suggest more smooth movement, to investigate the relationship between impaired coordination in people post-stroke and turning performance (Lewallen et al., 2021). As each peak of $\vec{H}_{WB}$ represents a direction change of $\vec{H}_{WB}$ generation, the *nAdj* also represents the number of peaks in $\vec{H}_{WB}$, but is calculated differently. The method used to calculate *nAdj* in the current study accounts for a change in direction of $\vec{H}_{WB}$ generation such that the smallest changes in direction of $\vec{H}_{WB}$ made around zero are taken into account. While fewer peaks in $\vec{H}_{WB}$ were associated with smoother movement in people post-stroke during turning (Lewallen et al., 2021), we found fewer *nAdj* (i.e., more smooth) were made when there was a greater range of $\vec{H}_{WB}$ (i.e., greater risk of balance loss) between alignments and groups during non-steady state tasks (Figure 3.2; Figure 3.3). This finding is likely a result of attempting to regulate $\vec{H}_{WB}$ back to zero after completing a movement requiring $\vec{H}_{WB}$ generation (e.g., sit-to-stand in the sagittal plane; Figure 3.4) and/or regulating $\vec{H}_{WB}$ near zero (e.g., sit-to-stand in the frontal and transverse planes; Figure 3.4). Therefore, the greater *nAdj* in non-amputees suggests that they generate minimal $\vec{H}_{WB}$ and actively adjust to have tight bounds around zero while people with TTA have fewer *nAdj* due to the greater $\vec{H}_{WB}$ generated. As such, both the greater range of $\vec{H}_{WB}$ and fewer *nAdj* suggest that people with TTA may have difficulty regulating $\vec{H}_{WB}$ during non-steady state tasks compared to non-amputees. Moreover, different measures other than the range of $\vec{H}_{WB}$ can provide insight on how dynamic balance is maintained during non-steady state tasks.

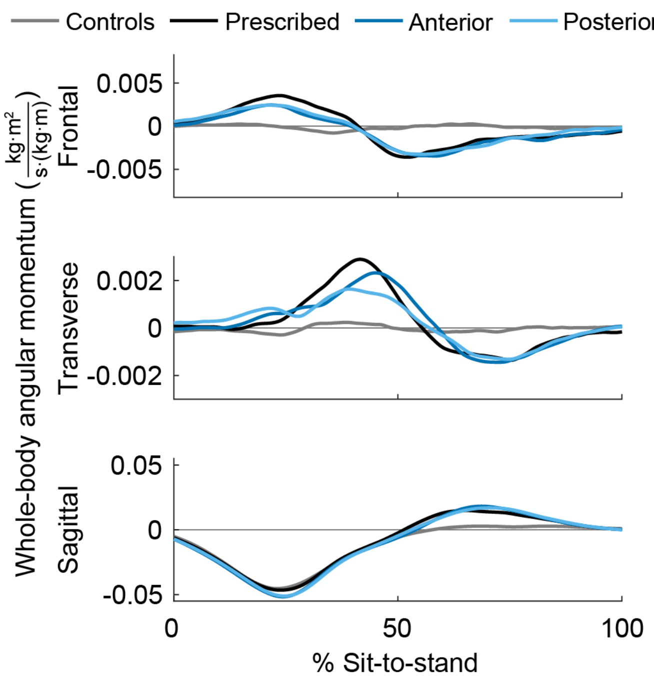

Figure 3.4. Average whole-body angular momentum ($\vec{\boldsymbol{H}}_{\boldsymbol{WB}}$; $\frac{\boldsymbol{kg \cdot m^2}}{\boldsymbol{s \cdot (kg \cdot m)}}$) in all planes of motion during sit-to-stand for controls and participants with TTA. For participants with TTA, the prescribed, anterior, and posterior alignments are plotted.

There were no differences in $\vec{H}_{WB}$ across alignments for the stand-to-sit, sit-to-walk, or walk-to-sit tasks. For stand-to-sit, failing to maintain balance would likely result in falling into the chair. As getting down to the chair is the goal, maintaining balance during stand-to-sit may be a lower priority suggesting that small differences in prosthetic alignment are less likely to affect balance during this task. For sit-to-walk and walk-to-sit, the varying strategies observed across participants likely masked any effect of alignment. For example, the leading leg during sit-to-walk was not consistent across participants where some chose to lead with the intact leg, and some chose to lead with the amputated leg. In addition, while the leading leg was consistent across alignments for most participants, this was not the case for all participants. For walk-to-sit, different strategies were used during the turn-to-sit portion of the task where some participants completed the turn before beginning the sitting motion, while others began lowering their body COM while turning. As these walk-to-sit strategies are associated with different turning strategies (i.e., step vs spin turns) (Weiss et al., 2016) that have different implications for balance (Taylor et al., 2005), future studies should focus on understanding the implications of these different strategies in people with TTA.

Limitations to this study may affect its generalizability. As we wanted to capture how participants performed functional tasks in daily life, we did not control for prosthetic foot type. Prosthetic foot properties may affect $\vec{H}_{WB}$ during different tasks (Shell et al., 2017); however, our within-participant design compared changes in alignment with the same prosthetic foot, mitigating this potential concern. The small participant cohort included high functioning (i.e., K3/K4; Table 3.1) individuals who had the strength and agility to perform these functional tasks comfortably. Given their high functional mobility, it possible that they could more easily adjust to small changes in alignment than lower functioning individuals could. Future work should investigate how different alignments affect the strategies used to perform functional tasks between people with TTA of different functional mobility levels.

3.6 Conclusion

This study investigated whether acute changes in prosthetic alignment affect $\vec{H}_{WB}$ during functional tasks. While we found a few alignments resulted in altered $\vec{H}_{WB}$ compared to the prescribed alignment during walking and sit-to-stand, these differences were small in magnitude and no other tasks were affected by prosthetic alignment. Thus, functionally mobile people with TTA are able to mitigate the effect of small and acute changes in alignment on $\vec{H}_{WB}$ during most seat transfer tasks. This is important as small changes in alignment may be frequently experienced in daily life due to pistoning of the residual limb in the socket, residual limb volume fluctuations, and/or differences in footwear. Compared to non-amputees, we found that people with TTA have greater range of $\vec{H}_{WB}$ and smaller number of adjustments to $\vec{H}_{WB}$ during sit-to-stand and stand-to-sit. These results add to the existing literature suggesting people with TTA have a reduced ability to control balance and require altered movement strategies to perform functional tasks. Future work should explore whether acute changes in prosthetic alignment have a similar effect on dynamic balance during tasks that require unilateral body support and force generation such as turning, walking on stairs, and walking at different speeds.

CHAPTER 4. Effects of Anterior-Posterior Shifts in Prosthetic Alignment on the Sit-To-Stand Movement in People with a Unilateral Transtibial Amputation[2]

4.1 Abstract

The sit-to-stand movement can be challenging for people with a transtibial amputation (TTA). The alignment of the prosthesis may influence the movement strategies people with TTA use to transfer from sit-to-stand by affecting foot placement. The purpose of this study was to determine how shifting the prosthetic foot anterior and posterior relative to the socket affects movement strategies used to transfer from sit-to-stand. To aid in interpretation, we compared movement strategies between people with and without TTA. Nine people with TTA and nine sex-, and age-matched non-amputee controls completed five self-paced sit-to-stand trials. With the posterior alignment, participants with TTA had 1) smaller braking GRF impulse on the prosthetic side and greater impulse on the intact side compared to the anterior alignment, 2) no significant differences between limbs which suggests greater braking impulse symmetry compared to anterior and prescribed alignments, and 3) smaller axial trunk range of motion compared to the prescribed alignment. There were also differences between participants with TTA and controls in braking GRF impulse, knee extension moment, anterior/posterior center of pressure position, and lateral

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and axial trunk range of motion. Based on these results, shifting the prosthetic foot posterior to the socket may be a useful tool to reduce braking impulse asymmetry and trunk motion in people with TTA during sit-to-stand. Thus, prosthetic alignment can have important implications for the comfort and ability of people with TTA to transfer from sit-to-stand as well as for development of secondary health conditions like low back pain, which is associated with compensatory movements.

4.2 Introduction

Standing up from a seated position is a common activity, performed approximately 60 times per day (Bussmann et al., 2008; Dall and Kerr, 2010), and is a vital functional activity for living independently. Performing a sit-to-stand movement can be physically demanding for those with disabling conditions, as it requires use of the trunk and legs for control, coordination (Roebroeck et al., 1994), and postural adjustment (Rodosky et al., 1989). For people with a unilateral transtibial amputation (TTA), performing this movement may be even more challenging. Specifically, the loss of direct neuromuscular control of the ankle joint may prevent people with TTA from providing the stabilization required for generating upward momentum (Jeon et al., 2019) as the center of mass (COM) moves away from the base of support during sit-to-stand. As a result, people with TTA have altered movement strategies compared to people without an amputation during sit-to-stand (Actis et al., 2018; Agrawal et al., 2016; Özyürek et al., 2014).

Kinematic and kinetic modifications necessary for completion of sit-to-stand in people with TTA have potential to be associated with negative clinically-relevant outcomes, such as challenges accomplishing sit-to-stand, poor balance leading to greater risk of falling, residual limb discomfort or injury, and greater intact limb knee loading. For example, people with TTA may have difficulty generating the braking (backward) impulses used during sit-to-stand. In people without an amputation these braking impulses are generated by the ankle muscles (Jeon et al., 2019) to oppose the propulsion created from pushing off of the seat, which moves the body forward (Hirschfeld et al., 1999). As such, reduced braking forces are associated with greater forward COM velocity (Jeon et al., 2019). The loss of neuromuscular control of the prosthetic side ankle in people with TTA may therefore lead to reduced postural control during sit-to-stand (Özyürek et al., 2014). In addition to the difficulty of generating the required forces for the sit-to-stand movement, people with TTA likely adjust their movement to avoid residual limb discomfort. Specifically, the smaller internal knee extension moment observed on the prosthetic side during sit-to-stand (Šlajpah et al., 2013) may be to avoid an increase in force of the distal anterior tibia onto the anterior socket wall which could cause discomfort and/or injury. Avoiding residual limb discomfort is also likely one reason why people with TTA increase weight-bearing on the intact leg when transferring their weight from the chair to the lower limbs (Actis et al., 2018; Agrawal et al., 2016; Özyürek et al., 2014). This asymmetric weight-bearing is also associated with greater lateral and axial trunk range of motion during sit-to-stand compared to people without an amputation (Actis et al., 2018). Weight-bearing asymmetry also leads to greater intact limb knee joint forces compared to the prosthetic side during sit-to-stand (Ferris et al., 2017a; Šlajpah et al., 2013), which are concerning given the high prevalence of osteoarthritis in the intact limb (Morgenroth et al., 2012; Norvell et al., 2005).

Optimizing foot placement may mitigate the negative consequences of the sit-to-stand strategies used by people with TTA. For example, placing both feet in a posterior position can reduce the forward COM displacement and velocity, as well as braking forces compared to a more anterior position in people without an amputation (Jeon et al., 2019). As an example from a different population with neuromuscular control deficits, people with hemiparesis reduce lateral trunk motion (Duclos et al., 2008; Lecours et al., 2008), center of pressure distance (Duclos et al., 2008; Han et al., 2015), weight-bearing asymmetry (Brunt et al., 2002; Roy et al., 2006) and increase affected side knee moment (Lecours et al., 2008) when placing the affected leg posterior to the unaffected leg. While altered foot placement could affect how people with TTA transfer from sit-to-stand, the prosthesis itself may limit the ability to actively alter foot position. Specifically, the prosthetic socket adds a kinematic constraint on posterior foot placement, as knee flexion can be limited by pressure of posterior brim of the socket against the popliteal fossa. Even if posterior foot placement is achieved, the position of the foot would likely cause altered residual limb pressure within the socket. However, foot position can be altered in people with TTA by altering the relative positions of the foot and socket in the sagittal plane during the prosthetic alignment process. Therefore, adjusting alignment may be useful in mitigating the challenges faced by people with TTA during sit-to-stand.

The purpose of this study was to determine how altering the translational alignment of a prosthesis in the sagittal plane affects the strategies used by people with TTA to transfer from sit-to-stand. Specifically, we compared GRF impulse, sagittal plane knee joint moment, anterior/posterior center of pressure position, and trunk range of motion between alignments. We hypothesized that a posterior alignment would decrease braking GRF impulse, increase prosthetic side knee moment, provide a more posterior center of pressure on the prosthetic side, and reduce trunk range of motion compared to prescribed and anterior alignments. For further interpretation of these findings, we compared people with TTA to age- and sex- matched non-amputee controls.

4.3 Methods

Participants

We recruited nine people with a unilateral transtibial amputation (TTA) through the University of Michigan Orthotics and Prosthetics Center and nine age- and sex- matched non-amputee Controls through an online database (https://umhealthresearch.org/) (Table 4.1). Potential participants with TTA were excluded if they could not walk independently for 10 minutes at a time for at least two months prior to data collection, and if the length of their residual limb prevented performing alignment adjustments. No participants with TTA reported any significant injury or pathology of their intact limb. Additional exclusion criteria for participants with TTA and control participants included, 1) taking medication that affected their ability to walk, 2) neurologic or cardiovascular disease, 3) uncorrected vision problems, and 4) mental capacity impairment that would negatively affect verbal communication. All participants provided their written informed consent to participate in this institutionally approved study.

Table 4.1. Participant demographics.

TTA	Age (years)	Sex	Mass (kg)	Height (m)	Cause of amputation	Prosthetic Foot	K-Level*	Controls	Age (years)	Sex	Mass (kg)	Height (m)
1	57	M	94.1	1.85	Dysvascular	College Park Velocity	K3	1	52	M	69.6	1.69
2	25	F	63.9	1.66	Cancer	Endolite Elite 2	K4	2	21	F	66.0	1.70
3	56	M	99.6	1.70	Traumatic	LP Variflex with EVO foot	K4	3	42	M	87.5	1.79
4	28	M	84.1	1.87	Traumatic	Freedom Innovations Agilix	K4	4	27	M	97.1	1.87
5	58	M	101.8	1.78	Traumatic	College Park Velocity	K3	5	48	M	116.1	1.91
6	63	M	121.9	1.81	Traumatic	TruStep	K4	6	62	M	84.4	1.81
7	65	M	90.4	1.67	Traumatic	Ottobock Triton 1C66	K3	7	61	M	79.8	1.78
8	61	M	115.6	1.88	Traumatic	Fillaeur All Pro	K4	8	53	M	108.4	1.78
9	56	M	109.6	1.82	Traumatic	Freedom Innovations Agilix	K4	9	54	M	99.8	1.97
Mean (SD)	52 (14.9)	-	97.9 (17.5)	1.78 (0.08)	-	-	-	-	47 (14.3)		89.9 (16.9)	1.81 (0.09)

*K-level is the Medicare Functional Classification level which can range from K0-K4

Experimental Protocol

Participants completed five self-paced sit-to-stand trials. At the start of the first trial, they were seated on a backless chair, with their hips and knees aligned at approximately 90 degrees of flexion, and feet placed hips width apart. Each foot and the chair were placed on separate force plates (AMTI Inc., Watertown, MA).

As part of a larger study, participants with TTA performed five sit-to-stand trials with three different alignments. The prescribed alignment was the first condition followed by altered translational alignment where the prosthetist shifted the foot 10-mm anterior or posterior relative to the socket from the prescribed alignment. The distance was verified using a flexible measuring tape. The anterior and posterior shifts were performed in random order and participants were not informed of what change had been made.

A 20-camera motion capture system (Motion Analysis Corp., Santa Rosa, CA, USA) tracked whole body kinematics at 120 Hz using a six degree of freedom marker set (Collins et al., 2009; Wilken et al., 2012) while ground reaction forces (GRFs) were collected simultaneously at 1200 Hz. Forty-five reflective markers were placed on bony landmarks including the C7, sternum, xiphoid process, T10, and bilaterally on the acromion, iliac crest, anterior and posterior superior iliac spines, greater trochanter, lateral and medial tibial epicondyles, lateral and medial malleoli, and 2nd and 5th metatarsals. For knee, ankle, and foot landmarks that were not present on the prosthetic side, markers were placed by visually mirroring the marker locations on the intact side (e.g., placing the medial malleolus marker on the prosthetic foot at the same height and relative position of the marker on the intact side). We also placed 4-marker clusters bilaterally on the thighs and shanks and a single marker on both heels.

Data Analysis

Marker position and GRF data were filtered using 4th-order low-pass Butterworth filters with cutoff frequencies of 6 Hz and 10 Hz, respectively. A 9-segment model including the head, trunk, pelvis, thighs, shanks, and feet was constructed in Visual3D (C-Motion, Germantown, MD, USA) using segment markers and landmarks to estimate inertial properties (Dempster and Aitkens, 1995). In addition, as prosthesis inertial properties (e.g., mass, center of mass location, etc.) differ to that of a biological limb that affect the calculation of the body center of mass, prosthesis inertial properties were adjusted according to (Ferris et al., 2017b). Local coordinate systems for each joint were defined based on International Society of Biomechanics guidelines (Wu, 2002). GRF data were normalized to body weight. Internal knee joint moments were calculated using inverse dynamic calculations and were normalized to body mass. We also calculated the anterior/posterior center of pressure position relative to the heel using the force plates and heel markers. Seat-off was defined as the instant the participant was no longer on the chair determined by when the force plate underneath the chair was unloaded.

Statistical Analyses

We first verified that the movements were performed consistently in position and timing. The time to complete sit-to-stand was consistent across trials and conditions; however, participants significantly adjusted their foot position after the first trial. Therefore, we averaged across the last four trials for each condition (see Appendix C for details). The primary dependent measures were the average posterior, anterior, medial, and vertical GRF impulse, the sagittal plane knee moment at seat-off, anterior/posterior center of pressure position at seat-off, and the trunk range of motion in each plane for each condition. We tested for differences in trunk range of motion across alignment conditions using a series of single factor within-subjects ANOVAs. The remaining dependent measures for the TTA group were compared between sides (intact, prosthetic) and alignment conditions (prescribed, anterior, posterior) using a series of 3x2 within-subjects ANOVAs. Secondarily, we compared the TTA group's prescribed alignment to the Control group using a series of 2-factor mixed model ANOVAs where group (TTA/Control) was the between-subjects factor and side (Intact/Dominant, Prosthetic/Non-dominant) was the within-subjects factor. A series of Welch's t-tests were used to compare trunk range of motion between groups. Significant main effects and interactions were explored using Estimated Marginal Means with a Sidak correction for multiple comparisons. We also calculated effects sizes using Cohen's *d* and 95% confidence intervals for the differences (Appendix C). All statistical analyses were performed using R 3.6.3 (R Core Team, Vienna, Austria, 2019).

4.4 Results

There were no significant differences in age ($p = 0.44$), mass ($p = 0.34$), or height ($p = 0.50$) between people with TTA and Controls (Table 4.1).

Ground Reaction Force Impulse

For participants with TTA, there was a significant main effect of alignment on propulsive (forward) GRF impulse ($p < 0.001$) (Figure 4.1). Propulsive GRF impulse was greater with the prescribed alignment compared to the anterior ($p < 0.001$; 95% CI [0.002, 0.005], $d = 0.69$) and posterior ($p < 0.001$; 95% CI [0.001, 0.004]; $d = 0.62$) alignments.

There was also a significant alignment × side interaction for the braking (backward) GRF impulse ($p = 0.020$). Braking GRF impulse was greater on the prosthetic side compared to the intact side with the anterior ($p = 0.006$; 95% CI [0.008, 0.038]; $d = 1.93$) and prescribed alignments ($p = 0.021$; 95% CI [0.003, 0.033]; $d = 1.51$). In addition, the braking GRF impulse on the prosthetic side was greater for the anterior alignment compared to the posterior alignment ($p = 0.017$; 95% CI [0.001, 0.013]; $d = 0.50$). On the intact side, the braking GRF impulse was greater for the posterior alignment compared to the anterior alignment ($p = 0.031$; 95% CI [0.001, 0.012]; $d = 0.66$). In addition, the vertical GRF impulse was greater on the intact side compared to prosthetic side across all alignments ($p = 0.009$; 95% CI [0.05, 0.25]; $d = 1.06$).

Similar to the alignment comparison, there was a significant side effect for vertical GRF impulse where the intact/dominant side vertical GRF impulse was greater compared to the prosthetic/non-dominant side across groups ($p = 0.020$; 95% CI [0.015, 0.016]; $d = 0.46$). There were also significant group and side effects for braking GRF impulse. Specifically, participants with TTA had greater braking GRF impulse compared to controls ($p = 0.006$; 95% CI [0.003, 0.015]; *d* = 0.80) while the prosthetic/non-dominant side had greater braking GRF impulse compared to the intact/dominant side (p = 0.003; 95% CI [0.005, 0.019]; *d* = 1.08).

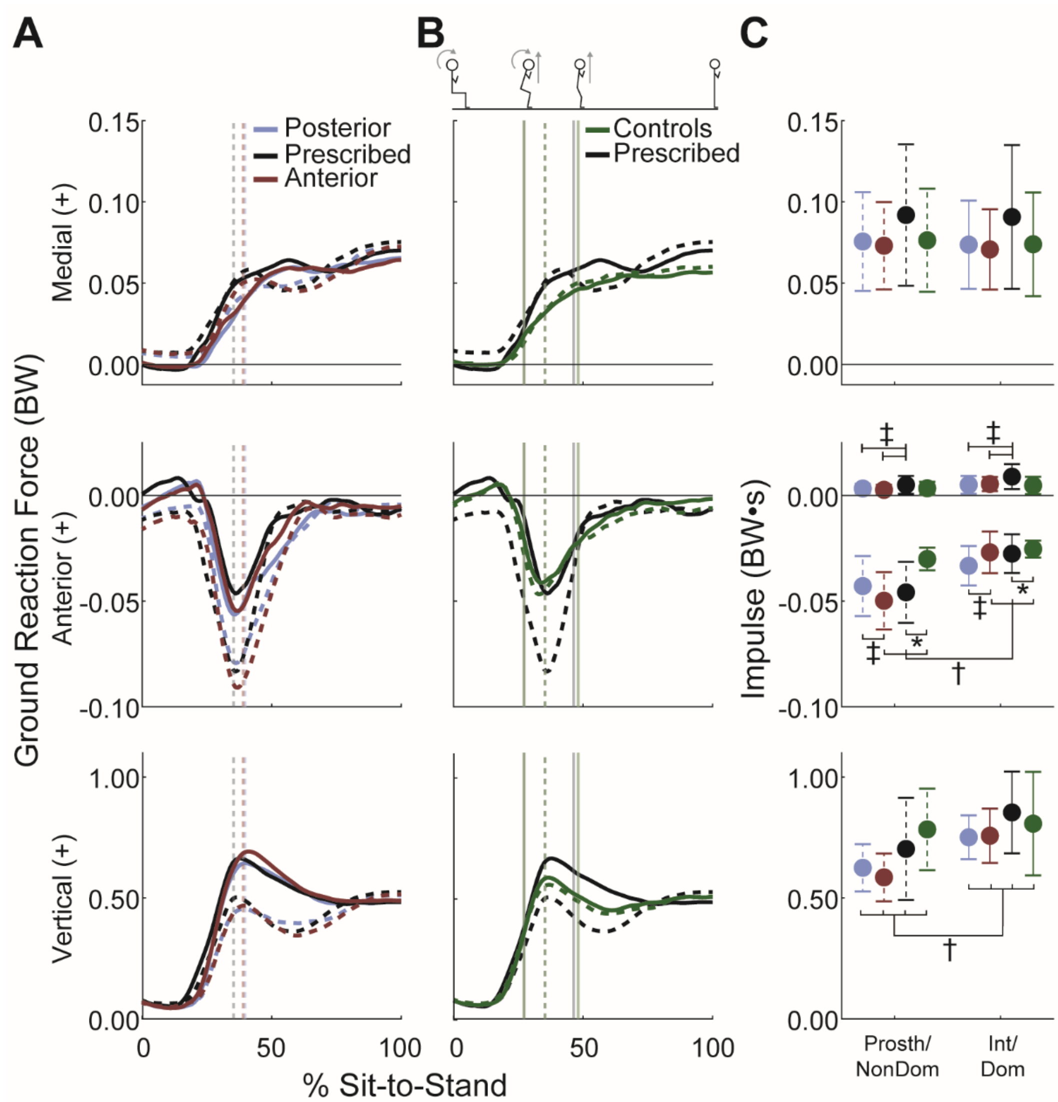

Figure 4.1. Average ground reaction forces (GRFs) during sit-to-stand compared **A)** between posterior (blue), anterior (red), and prescribed (black) alignment conditions and **B)** between the prescribed alignment (black) for people with TTA and Controls (green). **C)** Average positive and negative GRF impulse for Intact/Dominant (Int/Dom) and Prosthetic/Non-Dominant (Prosth/NonDom) sides for TTA and Controls. The Intact/Dominant sides (solid lines) and the Prosthetic/Non-dominant sides (dashed lines) are shown. The vertical solid lines indicate the beginning and end of the transition phase (see Appendix C) while the vertical dashed line indicates the instance of seat-off. Significant differences between the prescribed condition for

people with TTA and Controls are shown as (*), significant differences between alignments are shown as (‡) and differences between legs are shown as (†).

Knee Moments

There was no significant effect of alignment for the knee joint moment at seat-off (p = 0.694; Figure 4.2), but the intact side knee extension moment was greater than the prosthetic side ($p < 0.001$; 95% CI [0.493, 0.894]; $d = 3.03$).

There was a significant group × side interaction ($p < 0.001$). Participants with TTA had a lower knee extension moment on the prosthetic side compared to the intact side ($p < 0.001$; 95% CI [0.565, 0.785]; $d = 3.47$) and controls ($p < 0.001$; 95% CI [0.370, 0.725]; $d = 2.73$). There were no differences between the intact side of participants with TTA and the dominant side of controls ($p = 0.306$; 95% CI [-0.268, 0.09]; $d = 0.55$).

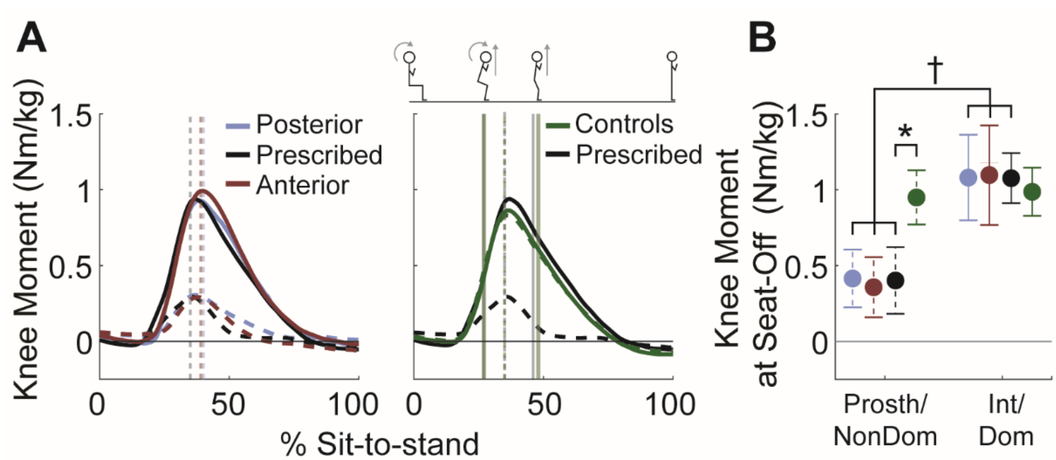

Figure 4.2. A) Average internal sagittal plane knee moments (Nm/kg) during sit-to-stand for the posterior (blue), anterior (red), and prescribed (black) alignments and Controls (green). Solid lines represent the Intact/Dominant (Int/Dom) side while dashed lines represent the Prosthetic/Non-Dominant (Prosth/NonDom) side. The vertical solid lines indicate the beginning and end of the transition phase (see Appendix C) while the vertical dashed line indicates the instance of seat-off. **B)** Average joint moment (Nm/kg) at the time of seat-off for each side. Error bars represent ± one standard deviation. Significant differences between the prescribed condition for people with TTA and Controls are shown as (*), significant differences between alignments are shown as (‡) and differences between legs are shown as (†).

Center of Pressure Position

There were no significant differences in center of pressure position between alignment conditions at the time of seat-off ($p = 0.180$; Figure 4.3). There was a significant main effect of side where the prosthetic side center of pressure position was more anterior than the intact side ($p < 0.001$; 95% CI [0.043, 0.102]; $d = 2.41$).

There was a significant group × side interaction for the center of pressure position at the time of seat-off ($p < 0.001$). The center of pressure position was more anterior on the prosthetic side compared to the intact side ($p < 0.001$; 95% CI [0.061, 0.093]; $d = 2.82$) and controls ($p < 0.001$; 95% CI [0.065, 0.111]; $d = 3.26$). There were no differences between the intact side of participants with TTA and the dominant side of controls ($p = 0.323$; 95% CI [-0.035, 0.012]; $d = 0.55$).

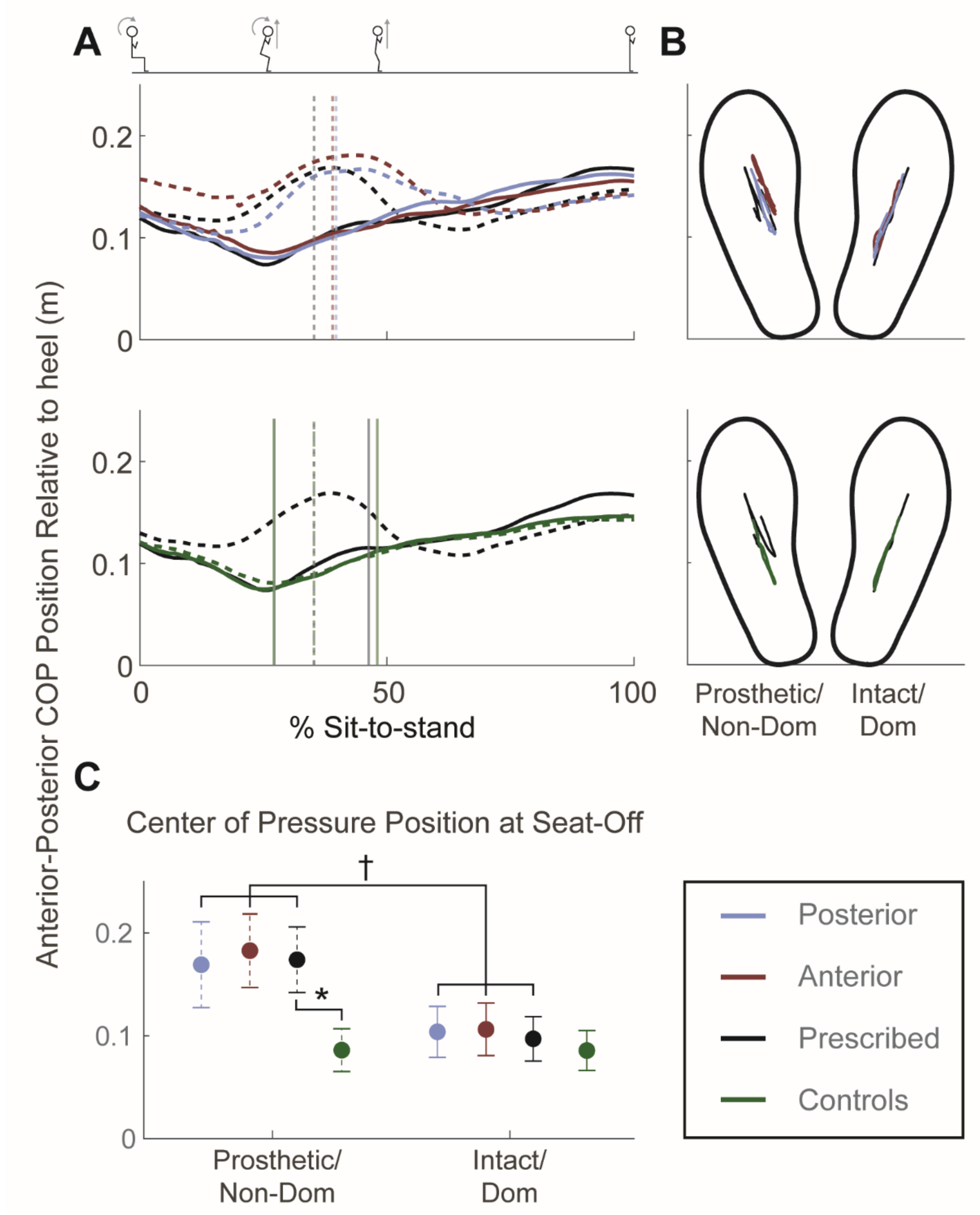

Figure 4.3. A) Average anterior-posterior center of pressure position (m) relative to the heel for each foot throughout sit-to-stand for the posterior (blue), anterior (red), and prescribed (black) alignments and Controls (green). The Intact/Dominant sides (solid lines) and the Prosthetic/Non-dominant sides (dashed lines) are shown. The vertical solid lines indicate the beginning and end of the transition (see Appendix C) phase while the vertical dashed line indicates the instance of seat-off. **B)** center of pressure trajectory in the medial-lateral and anterior-posterior directions. **C)** Average center of pressure position (m) relative to the heel at seat-off for each side (Prosthetic/Non-dominant and Intact/Dominant). Error bars represent the standard deviation.

Significant differences between the prescribed condition for people with TTA and Controls are shown as (*), significant differences between alignments are shown as (‡) and differences between legs are shown as (†).

Trunk Range of Motion

There was a significant main effect of alignment for trunk axial rotation range of motion ($p = 0.042$) where range of motion was smaller in the posterior alignment compared to the prescribed alignment ($p = 0.017$; 95% CI [0.329, 3.74]; $d = 1.26$; Figure 4.4). Compared to controls, participants with TTA had greater lateral lean ($p = 0.003$; 95% CI [0.594, 2.48]; $d = 1.64$) and axial rotation ($p < 0.001$; 95% CI [2.12, 5.39]; $d = 2.40$) range of motion.

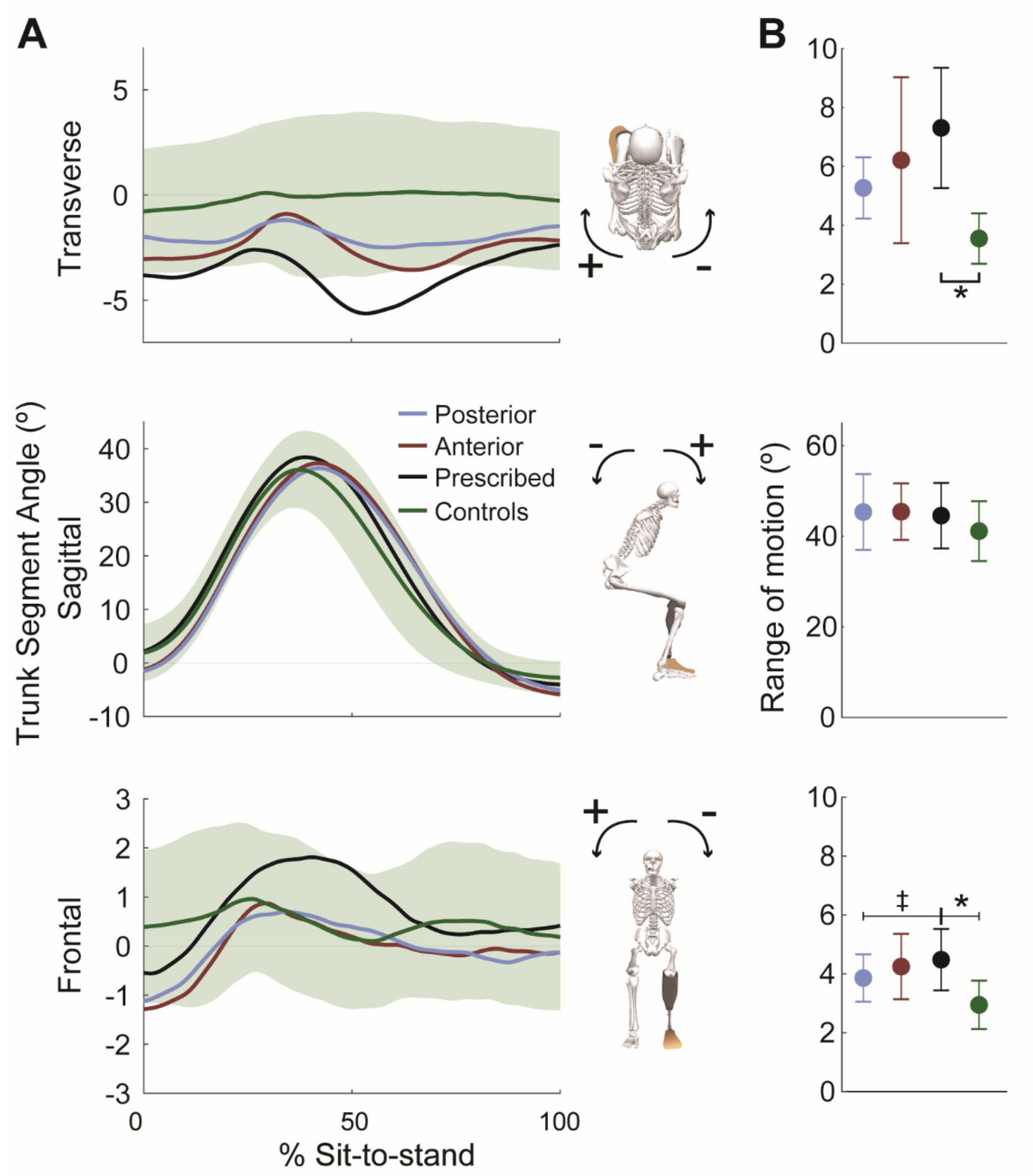

Figure 4.4. A) Average trunk angles in the three planes of motion during sit-to-stand for people with TTA using a posterior (blue), anterior (red), and prescribed (black) alignments and for Controls (green). The shaded region indicates ± one standard deviation for Controls. **B)** Average trunk range of motion in each plane where error bars indicate ± one standard deviation. Significant differences between the prescribed condition for people with TTA and Controls are shown as (*) and significant differences between alignments are shown as (‡).

4.5 Discussion

This study determined how altering the sagittal translational alignment of a prosthesis affected sit-to-stand in people with TTA. Our hypothesis that a posterior alignment would decrease braking ground reaction force (GRF) impulse compared to the prescribed alignment was not supported. We did find greater propulsive impulse with the anterior and posterior alignment conditions compared to the prescribed condition. While the differences between alignment for propulsive impulse might suggest a different strategy, there were no differences between people with and without TTA. In addition, a propulsive impulse from the feet during sit-to-stand may not be clinically relevant as it only occurs at the beginning of the movement and has a peak propulsive force of < 1% BW (Hirschfeld et al., 1999). While there was no difference in braking impulse between the posterior and prescribed alignments, there was a smaller braking impulse with the posterior alignment compared to anterior alignment on the prosthetic side (Figure 4.1). In addition, the braking impulse on the intact side with the posterior alignment was greater compared to the anterior alignment. These differences demonstrate that the braking impulses for each leg are more similar to each other with a posterior alignment compared to the anterior alignment. Furthermore, braking impulse was significantly greater on the prosthetic side compared to the intact side with the anterior and prescribed alignments, while there was no statistically significant difference between legs for the posterior alignment. Thus, our results suggest that a posterior alignment may have greater braking impulse symmetry compared to the anterior and prescribed alignments. In addition, while braking impulse is required to stabilize the propulsive impulse from the buttocks pushing from a chair during sit-to-stand (Hirschfeld et al., 1999), people with TTA generated greater braking impulse compared to people without an amputation. Considering that impulse is the change in momentum, people with TTA likely use greater braking impulse to control the forward momenta of the trunk and thighs (Pai and Rogers, 1991) to transition from sit-to-stand safely.

Our hypotheses that a posterior alignment would increase prosthetic side knee moment and provide a more posterior center of pressure position were also not supported. Similar to previous studies (Ferris et al., 2017a; Šlajpah et al., 2013), our participants with TTA performed the sit-to-stand movement asymmetrically with greater knee extension moments on the intact side compared to the prosthetic side, in all alignments, and compared to control participants (Figure 4.2). As joint moments are affected by both the magnitude of the GRF and the center of pressure position, the combination of a smaller vertical GRF, greater braking GRF, and more anterior center of pressure position is consistent with the smaller prosthetic side knee moment. These differences in prosthetic side mechanics may enable people with TTA to transfer from sit-to-stand and they likely aid in avoiding large forces of the distal tibia on the anterior socket wall to prevent discomfort and/or injury.

Asymmetric weight-bearing has also been associated with greater lateral and axial trunk range of motion during sit-to-stand in people with TTA compared to people without an amputation (Actis et al., 2018). Based on prior work in people with hemiparesis during sit-to-stand (Duclos et al., 2008; Lecours et al., 2008), we expected that a posterior alignment of the prosthesis would decrease lateral trunk range of motion. While this hypothesis was not supported, there was a trend toward decreased lateral trunk range of motion with the posterior alignment compared to the prescribed alignment ($p = 0.071$). In addition, participants with TTA decreased trunk axial rotation with the posterior alignment compared to the prescribed alignment (Figure 4.4). Compared to controls, participants with TTA had greater lateral lean and axial rotation. This has important implications as greater trunk axial rotation and lateral lean range of motion may lead to greater low back loads (Actis et al., 2018) thereby increasing the risk of low back pain, joint degeneration, and/or injury. Given that the posterior alignment decreased axial trunk range of motion, altering alignment may be one strategy to address low back issues in people with TTA. However, the magnitude of change in range of motion and load needed to mitigate this secondary condition remains unknown.

In this study, we chose to shift prosthetic alignment from that clinically optimized by a prosthetist, rather than using a standard "neutral" baseline for all participants. While this choice may have increased the variability between individuals, it is more clinically-relevant in the context of prosthetic alignment as a single "neutral" alignment for all people with TTA may not exist (Blumentritt, 1997; Chow et al., 2006; Sin et al., 2001; Zahedi et al., 1986). In addition, alignment is likely dependent on the prosthetic foot which differed between participants in this study (Table 4.1). Investigating the prescribed alignment for people with TTA provides insight on the movement strategies used in daily life.

In addition, we chose to make relatively small changes in alignment to ensure that the participant was comfortable and to avoid excessive compensatory movement due to pain or discomfort. The 10-mm shift was within the range of "acceptable alignments" previously reported (Sin et al., 2001) and was within the range used in prior studies exploring the influence of alignment changes on walking mechanics (Boone et al., 2013; Kobayashi et al., 2013). Although larger alignment shifts may have led to different results, we aimed to stay within the realm of what was felt to be most clinically relevant.

A potential limitation in the generalizability of our findings is the standardization of the task. Here we restricted arm use and specified initial joint angles. In spite of this instruction, participants adjusted their foot movement after the first trial of each set of five sit-to-stand trials, which may be more representative of how they would choose to perform the sit-to-stand movement. Another factor relating to generalizability is our relatively small and homogeneous cohort, which was limited to highly mobile individuals (i.e., K3/K4; Table 4.1). More mobile individuals typically have greater strength and agility, which likely enables them to adjust to different alignments more easily. The small sample size may also contribute to Type II errors in cases where we did not find statistically significant differences. In contrast, the number of comparisons made may increase the likelihood of Type I errors where we did find significance. Thus, we have worked to balance the likelihood of these potential errors. To account for Type I errors, we performed Sidak corrections on all comparisons. Given the possibility of Type II errors, we have provided effect sizes and confidence intervals to assess the magnitude of the differences for all comparisons. These results can also be used to power future studies.

4.6 Conclusion

Altering sagittal plane prosthetic alignment can affect how people with TTA transfer from sit-to-stand. The differences between alignments in braking GRF impulse and axial trunk range of motion may be important as postural stability during sit-to-stand depends on controlling the forward momentum generated by the trunk and buttocks with the forces at the feet (Hirschfeld et al., 1999). Moreover, using alignment to reduce excessive trunk motion may be useful to reduce the risk of low back pain common in people with TTA (Kulkarni et al., 2005; Sivapuratharasu et al., 2019). In addition, these differences were found in spite of small changes in alignment, suggesting that it may be important to evaluate sit-to-stand in addition to walking during the clinical alignment process. Future work should explore the effects of prosthetic alignment on sit-to-stand in a larger cohort, including individuals with a lower functional capacity.

CHAPTER 5. The Effect of Acute Changes in Transtibial Prosthetic Alignment on Hip and Low Back Joint Loads During Sit-To-Stand

5.1 Abstract

Hip joint and low-back pain are common secondary conditions affecting people with a transtibial amputation (TTA). A greater reliance on the intact limb during daily activities is associated with asymmetric hip muscle activation and altered trunk movement patterns which may be risk factors for joint pain and degeneration. Sit-to-stand is one task necessary for independent living, requires substantial hip and trunk range of motion, and is completed asymmetrically by people with TTA. As prosthetic alignment can affect muscle activity during sit-to-stand and muscles are important contributors to joint loading, we investigated the effect of prosthetic alignment on hip and low-back joint contact forces in people with TTA during sit-to-stand. Kinematics, ground reaction forces, and muscle activity data were collected from nine people with a unilateral transtibial amputation and nine age- and sex- matched non-amputees during five self-paced sit-to-stand trials. Participants with TTA completed the sit-to-stand task with their prescribed alignment and six shifted alignments, including ±10 mm anterior/posterior, ±10 medial/lateral, and ±20 mm in the vertical direction. Hip and L4-L5 joint contact force magnitude peak and impulse were calculated with a musculoskeletal model of the torso, lumbar spine, pelvis, lower limbs, and 294 musculotendon actuators using a static optimization framework in OpenSim. There were no differences in hip and L4-L5 joint contact force measures between alignments or between groups. Participants with TTA had a greater peak hip joint contact force on the intact side

hip compared to the amputated side hip near seat-off across all alignments. These results suggest that small and acute changes in prosthetic alignment do not affect hip and low-back joint loads during sit-to-stand in functionally mobile people with TTA. Future work should investigate how longer exposure to prosthetic alignment changes affect cumulative joint loading over time and across different activities of daily living.

5.2 Introduction

People with a unilateral transtibial amputation commonly report pain in the lower-limb joints (Friberg, 1984; Struyf et al., 2009) and have high rates of chronic low back pain (Kulkarni et al., 2005). These secondary conditions may be caused by compensatory movements used by people with TTA throughout daily life. In particular, the greater reliance on the intact leg for weight-bearing (Agrawal et al., 2011; Nolan and Lees, 2000; Nolan et al., 2003) and balance (Curtze et al., 2012; Ku et al., 2014) places greater forces on the intact leg joints compared to the amputated leg in daily activities like walking (Molina Rueda et al., 2013; Sharifmoradi et al., 2017; Yu et al., 2014) and sit-to-stand (Nolasco et al., 2020; Slajpah et al., 2013; Wagner et al., 2020). Greater cumulative loading over time is likely to cause joint pain and cartilage degeneration, which are more common in people with TTA than non-amputees (Struyf et al., 2009). In addition, the lack of ankle muscle function in people with TTA leads to altered amputated side hip muscle strength (Croisier et al., 2001; Nolan, 2009) and activation (Schmalz et al., 2001), creating asymmetric hip mechanics. These altered hip mechanics likely contribute to joint pain and degeneration in this population (Gailey et al., 2008). As asymmetry between intact and amputated side joint mechanics also affect trunk control (Yoder et al., 2019), the greater reliance on the intact leg also results in altered trunk motion and low-back muscle activity in people with TTA (Butowicz et al., 2018). These altered low back mechanics likely contribute to altered low-back loads (Actis et al., 2018b; Yoder et al., 2015) and result in the high rates of low-back pain in people with TTA (Kulkarni et al., 2005). As sit-to-stand is a physically demanding task requiring control and coordination of the trunk and lower extremities, hip and low-back forces play an important role in successful completion of the task.

People with TTA complete the sit-to-stand task asymmetrically. Due to greater reliance on the intact leg, people with TTA rise from a chair by moving the body center of mass toward the intact leg (Wagner et al., 2020) by leaning their trunk over the intact leg (Actis et al., 2018b; Nolasco et al., 2020; Slajpah et al., 2013). The asymmetry between intact and amputated sides is also apparent in lower limb kinematics and kinetics. Specifically, ankle, knee, and hip angle asymmetry is greater in people with TTA compared to non-amputees when rising from different seat heights at different speeds (Slajpah et al., 2013). Similarly, sagittal ankle and knee joint moments are greater on the intact side compared to the amputated side (Nolasco et al., 2020; Slajpah et al., 2013). In contrast, the amputated side hip has a greater sagittal joint moment compared to the intact side (Slajpah et al., 2013), which may be due at least in part to greater amputated side rectus femoris activity compared to the intact side and non-amputees (Wagner et al., 2020). As both intact and amputated side rectus femoris activity increase when shifting the prosthetic foot medial relative to the socket and gluteus medius activity decreased with a lateral shift (Wagner et al., 2020), medial/lateral changes in prosthetic alignment may affect the muscle contributions to joint mechanics during sit-to-stand. While prosthetic alignment affects how the force is transferred from the ground throughout the body, altered muscle activity has been shown to be the primary factor affected by small changes in prosthetic alignment during standing (Blumentritt et al., 1999) and sit-to-stand (Wagner et al., 2020). In addition, anterior/posterior changes in alignment affect sit-to-stand trunk kinematics (Nolasco et al., 2020), which may affect low-back muscle activity. As muscle forces are large contributors to joint contact forces (Correa et al., 2010), the muscle activity differences between prosthetic alignments may contribute to altered joint loading in people with TTA.

As asymmetric, large, and repeated joint forces are mechanical risk factors for developing lower extremity (Hurwitz et al., 2001) and low-back (Kulkarni et al., 2005) joint degradation and pain, measuring joint contact forces can provide insight related to these secondary conditions in people with TTA. However, *in vivo* joint contact forces are nearly impossible to measure directly and thus modeling approaches can be used to estimate muscle and joint forces non-invasively. Using these musculoskeletal modeling approaches, prior work has identified greater peak intact knee joint contact force compared to the amputated leg in people with TTA (Silverman and Neptune, 2014). Similarly, people with TTA generally have greater hip joint contact forces on the intact leg compared to the amputated leg during walking (Sharifmoradi et al., 2017; Yu et al., 2014) and running (Sepp et al., 2020). Moreover, people with TTA have greater low-back loads compared to non-amputees during walking (Yoder et al., 2015) and sit-to-stand (Actis et al., 2018b). As these studies found muscle forces to be major contributors to the estimated joint contact forces, it is important to consider muscle forces in joint force estimations. In addition, while muscle forces are altered with different alignments (Fang et al., 2007) and lead to altered knee ligament forces with different alignments (Fang et al., 2009), it is unclear how prosthetic alignment affects hip and low-back joint contact loads. While another study reported decreased intact leg hip joint and increased low-back intersegmental forces with a shorter prosthesis (Yu et al., 2014), the joint force estimate did not take muscle forces into account.

The purpose of this study was to determine the effect of prosthetic alignment on hip and low-back (L4-L5) joint contact forces in people with TTA. To provide context for the findings, we also compared hip and low-back forces between people with and without TTA. We expected hip joint contact forces to be affected by medial/lateral and tall/short alignments, while low-back forces would be affected by anterior/posterior and tall/short alignments.

5.3 Methods

Participants

We recruited nine people with a unilateral transtibial amputation (TTA) through a local orthotics and prosthetics clinic and nine age- and sex- matched controls through an online database (https://umhealthresearch.org/) (Table 5.1). Potential participants were screened to ensure they did not have a history of neurological or cardiovascular disease, uncorrected vision impairments, take medication that affected their ability to walk, or any mental capacity impairment that would negatively affect verbal communication. Participants were excluded if they were unable to walk independently for at least 10 minutes at a time for at least two months prior to data collection. No participants with TTA reported any significant injury or pathology of their intact leg. This study was approved by the institutional IRB and all participants provided their written informed consent.

Table 5.1. Participant demographics

TTA	Age (years)	Sex	Weight (kg)	Height (m)	Cause	Prosthetic Foot	K-Level*	Control	Age (years)	Sex	Weight (kg)	Height (m)
1	57	M	94.1	1.85	Dysvascular	College Park Velocity	K3	1	52	M	69.6	1.69
2	25	F	63.9	1.66	Cancer	Endolite Elite 2	K4	2	21	F	66.0	1.70
3	56	M	99.6	1.70	Trauma	Ossur LP Variflex with EVO	K4	3	42	M	87.5	1.79
4	28	M	84.1	1.87	Trauma	Freedom Innovations Agilix	K4	4	27	M	97.1	1.87
5	58	M	101.8	1.78	Trauma	College Park Velocity	K4	5	48	M	116.1	1.91
6	63	M	121.9	1.81	Trauma	College Park TruStep	K3	6	62	M	84.4	1.81
7	65	M	90.4	1.67	Trauma	Ottobock Triton 1C66	K3	7	61	M	79.8	1.78
8	31	M	93.4	1.82	Trauma	Freedom Innovations Maverick	K3	8	29	M	79.4	1.80
9	56	M	109.6	1.82	Trauma	Freedom Innovations Agilix	K3	9	54	M	99.8	1.97
Mean (SD)	50 (16)	-	97.5 (16.6)	1.79 (0.08)	-	-	-	**Mean (SD)**	45 (15)	-	88.8 (16.3)	1.81 (0.09)

*K-level is the Medicare Functional Classification level which can range from K0 to K4

Experimental Protocol

All participants completed five self-paced sit-to-stand trials beginning from a seated position on a backless chair with their hips and knees aligned at approximately 90 degrees of flexion with a goniometer, and feet placed approximately hips-width apart. Each foot and the chair were placed on separate force plates (AMTI Inc., Watertown, MA, USA).

Participants with TTA initially performed the sit-to-stand trials with the prosthetic alignment prescribed by their certified prosthetist. Translational alignment changes were made by a certified prosthetist relative to each participant's prescribed alignment in the anterior-posterior directions by 10 mm, medial-lateral directions by 10 mm, and the vertical direction by 20 mm.

These alignment changes were defined as the change in foot position relative to the socket. The alignment changes were verified using a flexible measuring tape by the prosthetist. After the prescribed condition, the remaining alignment conditions were completed randomly, in pairs (i.e., anterior and posterior, medial and lateral, tall and short) and participants were not informed of the change that had been made.

For all trials, a full-body marker set that tracks six degrees-of-freedom at each joint (Collins et al., 2009; Wilken et al., 2012) and a 20-camera motion capture system (Motion Analysis, Santa Rosa, CA) tracked body motion at 120 Hz. Forty-five reflective markers were placed on body landmarks including the C7, sternum, xiphoid process, T10, and bilaterally on the acromion, iliac crest, anterior and posterior superior iliac spines, greater trochanter, lateral and medial tibial epicondyles, lateral and medial malleoli, and 2^{nd} and 5^{th} metatarsals. For the amputated side, markers were placed so they mirrored the intact side. Marker clusters were placed bilaterally on the thighs and shanks while a single marker was placed on each heel for segment tracking. Surface electromyography (EMG) sensors (Delsys, Inc., Boston, MA) were placed bilaterally on thoracic paraspinals (TP), lumbar paraspinals (LP), gluteus medius (GM), rectus femoris (RF), vastus lateralis (VL), biceps femoris long head (BF), medial gastrocnemius (MG), and tibialis anterior (TA). For participants with TTA, EMG sensors were placed unilaterally on the intact side MG, TA, soleus (SOL) and lateral gastrocnemius (LG). EMG and ground reaction forces (GRFs) were collected at 1200 Hz.

Data Processing

Marker trajectories and GRF data were filtered using a 4^{th}-order low-pass Butterworth filter. Marker trajectories were filtered with a cutoff frequency of 6 Hz while the GRFs were filtered with a cutoff frequency of 10 Hz. A 9-segment model consisting of the head, trunk, pelvis, thighs, shanks, and feet was developed in Visual3D (C-Motion, Germantown, MA) for model scaling and to determine the inverse kinematics solution for each trial. The model had 6 degrees of freedom (DOF) between the ground and pelvis, three rotational DOFs between the pelvis and trunk, three rotational DOFs between the pelvis and each femur, one rotational DOF between the femur and shank (tibia and fibula), and 2 DOFs between each shank and foot (i.e., ankle plantarflexion/dorsiflexion and inversion/eversion). The inverse kinematics solution for each sit-to-stand trial was computed from the initiation of forward center of mass velocity until the termination of anterior and mediolateral center of mass velocity using a least-squares optimization algorithm (Lu and O'Connor, 1999).

Musculoskeletal Modeling and Simulation

We used a previously developed bipedal musculoskeletal model (Actis et al., 2018a) with lower limbs from Delp et al. (Anderson and Pandy, 1999, 2001; Delp et al., 1990; Yamaguchi and Zajac, 1989), lumbar spine and torso from Christophy et al. (Christophy et al., 2012), torso muscle strengths from Bruno et al. (Bruno et al., 2015), body mass distributions from Winter (Winter, 2009), and the same DOFs as the Visual3D model in OpenSim 4.2 (simtk.org). The intervertebral joint motion of the five lumbar vertebrae was defined with a linear function based on overall trunk-pelvis angles (Christophy et al., 2012). This model includes 294 Hill-type musculotendon actuators consisting of a contractile element representing active muscle fibers, a series elastic element representing tendons, and parallel elastic elements representing passive fiber stiffness where muscle contraction was governed by force-length-velocity relationships (Zajac, 1989). This model was altered to represent a unilateral transtibial amputation by removing all bodies and musculotendon actuators distal to the affected tibia and transecting the affected tibia and fibula by modifying the mass, inertia, and graphics to represent an amputation at 50% of the limb length (LaPre et al., 2018). A generic socket and pylon segment was articulated with the affected tibia through a reversed joint with 4 DOFs including flexion/extension, abduction/adduction, axial rotation, and axial translation (pistoning) (LaPre et al., 2018). An ankle-foot prosthesis was attached to the end of the pylon with a single DOF pin joint representing foot flexion (LaPre et al., 2018). The location of the ankle-foot prosthesis was changed according to the different anterior/posterior and medial/lateral alignment conditions by translating the foot relative to the socket-pylon segment. For the tall/short alignments, the length of the prosthesis model was scaled based on the scaling factors exported from Visual3D.

All models were scaled in size and body mass for each participant based on the static, standing trial. The inverse kinematics solution from Visual3D, filtered GRF data, and scaled models were then input to a residual reduction algorithm to ensure dynamic consistency between the inverse kinematics solution and GRFs. Then a static optimization using interior point optimization (Wächter and Biegler, 2005) was used to solve for muscle activations at each instant in time for the duration of the simulation by minimizing the sum of muscle activations squared:

$$J = \sum_{m=1}^{294} (a_m)^2$$

subject to the following constraints for $j = 1{:}k$:

$$\sum_{m=1}^{n} [a_m f(F_m^o, l_m, v_m)] r_{mj} = \tau_j$$

where a_m is the activation of muscle m, f is the maximum muscle force dependent on its maximum isometric force (F_m^o), length (l_m), and velocity (v_m), r_{mj} is the moment arm of the muscle m about joint j, n is the number of muscles spanning the joint j, k is the total number of joints in the model, τ_j and is the net joint torque at joint j. It is important to note that this static optimization algorithm ignores the parallel elastic elements of the musculotendon actuator by assuming an inextensible tendon. For models with TTA, the prosthetic ankle joint was actuated with idealized torque actuators that reflected the planarflexion/dorsiflexion and subtalar pronation/supination net joint moments computed from inverse dynamics.

Data Analysis

EMG data were demeaned, band-pass filtered between 30 and 500 Hz, rectified and low-pass filtered with a cutoff frequency of 2.5 Hz (Actis et al., 2018a; Actis et al., 2018b; Drake and Callaghan, 2006). EMG signals were then normalized to their peak value during each sit-to-stand trial. Muscle activations from the simulations were low-pass filtered with a 4th-order Butterworth filter with a 2.5 Hz cutoff frequency and grouped according to anatomical location (Table 5.2). The resulting linear envelopes were normalized to peak activation for each trial and visually compared to experimentally collected EMG to ensure similar onset and offset times. We also evaluated residual forces and moments from the residual reduction algorithm as a metric of simulation quality.

Individual muscle forces determined from static optimization were used to calculate the three-dimensional magnitude of the total hip joint contact force (HJCF) and L4-L5 contact force (LBCF) from the seat-off to standing portion of the sit-to-stand task for each trial. All joint contact forces were low-pass filtered using a Butterworth filter with a 6 Hz cutoff frequency and normalized to body weight for each participant. Peak LBCF, peak HJCF, and LBCF and HJCF impulse calculated as the time integral of LBCF from seat-off-stand, were computed for each trial, then averaged across trials for each participant. For participants with TTA, these measures were computed for each alignment condition.

Table 5.2. Anatomical grouping of musculotendon actuators within the model compared to EMG signals**.**

Muscle signals recorded through EMG	Grouped musculotendon actuators	# of modeled musculotendon actuators per side
Thoracic paraspinals	Iliocostalis pars thoracis, longissimus pars thoracis	29
Lumbar paraspinals	Iliocostalis pars lumborum, longissimus pars lumborum	9
Gluteus medius	Gluteus medius (three compartments) Gluteus minimus (three compartments)	6
Biceps femoris	Biceps femoris long head, semimembranosus, semitendinosus, gracilis	4
Vastus lateralis	Vastus lateralis, vastus intermedius, vastus medialis	3
Rectus femoris	Rectus femoris	1
Medial gastrocnemius	Medialis gastrocnemius, lateral gastrocnemius	2
Tibialis anterior	Tibialis anterior, extensor digitorum longus, extensor hallucis longus, peroneus tertius	4

Statistical Analysis

We tested for differences in peak HJCF, peak LBCF, HJCF impulse, and LBCF impulse between alignments using separate linear mixed models with alignment as a fixed effect and participants as a random effect. For impulse measures we also included the time taken to complete the movement as a covariate. To test differences between intact and amputated side peak HJCF and HJCF impulse we added side (Intact/Dominant – Amputated/Non-Dominant) and alignment × side interaction as fixed effects. We also used a linear mixed model with group (TTA - Controls) and side as fixed factors and participants as a random effect to test between groups, sides, and the group × side interaction for the prescribed alignment only. Significant main effects were explored using estimated marginal means with specific contrasts and a Sidak correction for multiplicity. For significant alignment main effects, we compared each alignment condition to the prescribed condition. All statistical analyses were performed using R 3.6.3 (R Core Team, Vienna, Austria), with a significance level of $\alpha = 0.05$.

5.4 Results

Two participants with TTA did not complete all alignment conditions due to time constraints. P1 did not complete the tall/short alignment conditions while P8 did not complete the anterior/posterior alignment conditions. In addition, the simulation did not run for the short alignment trials for P8 likely due to the knee hyperextension observed during those trials.

Simulation Quality

The average root-mean-squared (RMS) residual forces in the model across all trials and participants were 2.02 %BW in the anterior/posterior direction, 3.28 %BW in the vertical direction, and 1.16 %BW in the medial/lateral direction. The vertical force had the largest residual force with an average peak of 9 %BW (max of 16 %BW in the TTA model with a lateral alignment). The average RMS residual moments were 0.70 %BW-m in the frontal plane, 0.45 %BW-m in the transverse plane, and 2.01 %BW-m in the sagittal plane. The residual moment was greatest in the sagittal plane with an average peak of 5.44 %BW-m (max of 11.02 %BW-m in the TTA Model with the medial alignment). Average RMS kinematics adjusted by the residual reduction algorithm were less than 5 mm for pelvis position and less than 0.51° for all joint angles. The largest difference in kinematics occurred at the knee with average peak knee flexion adjustment of 1° a maximum of 4° in the TTA model with the prescribed alignment. There was generally good visual agreement between average muscle activations and EMG activity (Figure 5.1). However, the control model bilateral rectus femoris, vastus lateralis, and medial gastrocnemius differed in activation pattern compared to EMG. For the TTA model, the rectus femoris and medial gastrocnemius activations on the intact side differed from EMG while the vastus lateralis activation on the amputated side differed from EMG. The agreement between model activation and EMG was similar between all alignment conditions.

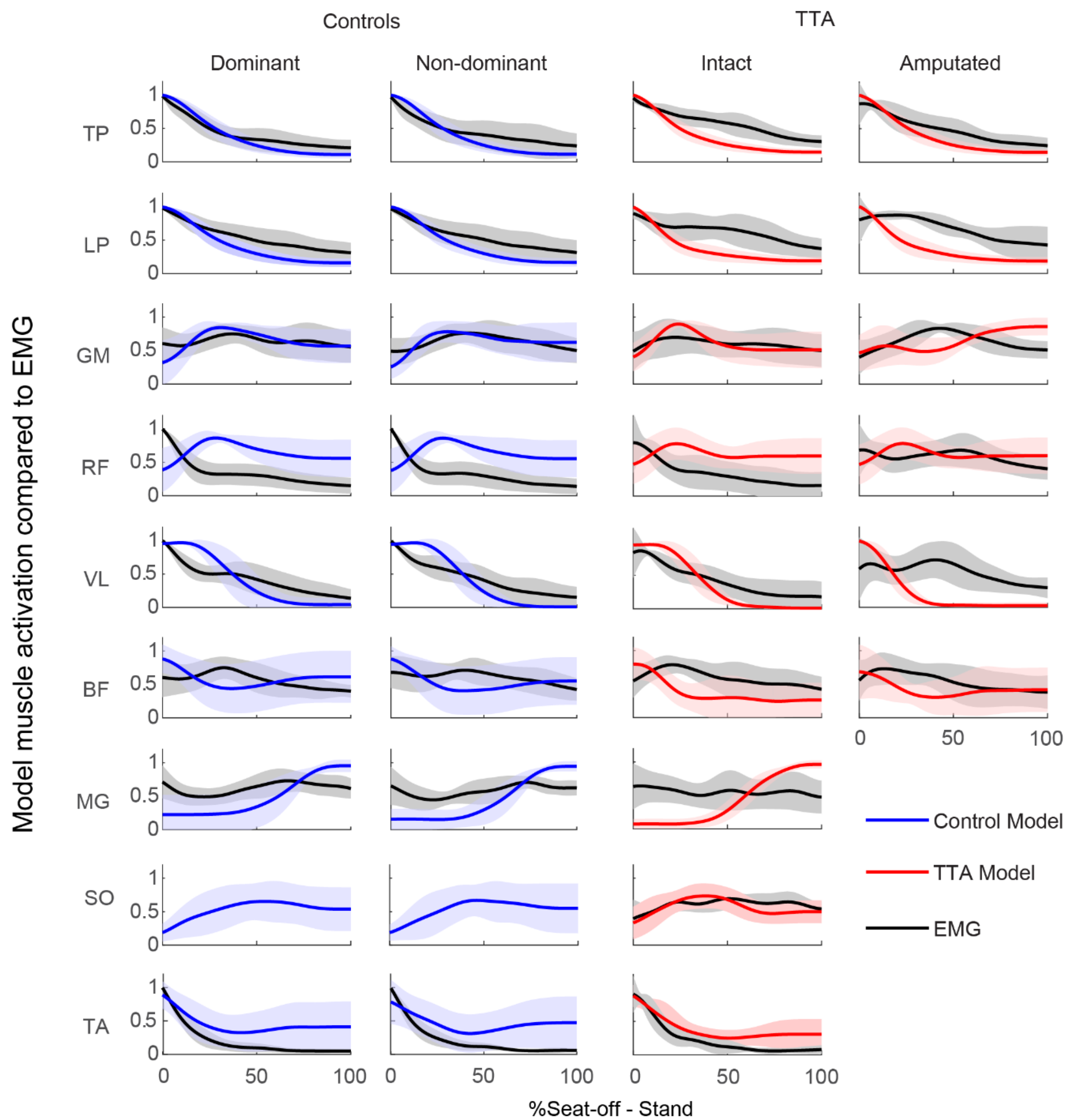

Figure 5.1. Experimental EMG and simulated muscle activations (average ± standard deviation) during seat-off-to-stand for controls (blue) and participants with TTA (red) with each alignment.

Peak hip joint contact forces

Within the TTA group, there was no significant main effect of alignment ($p = 0.743$) or alignment × side interaction effect ($p = 0.453$) for peak hip joint contact force (HJCF) (Figure 5.2).

There was a significant main of side (p = 0.004) where the intact side peak HJCF was greater compared to the amputated side (Figure 5.3).

When comparing people with TTA and controls, there was no significant main effect of group (p = 0.789), side (p = 0.302), or group × side interaction (p = 0.342) for peak HJCF.

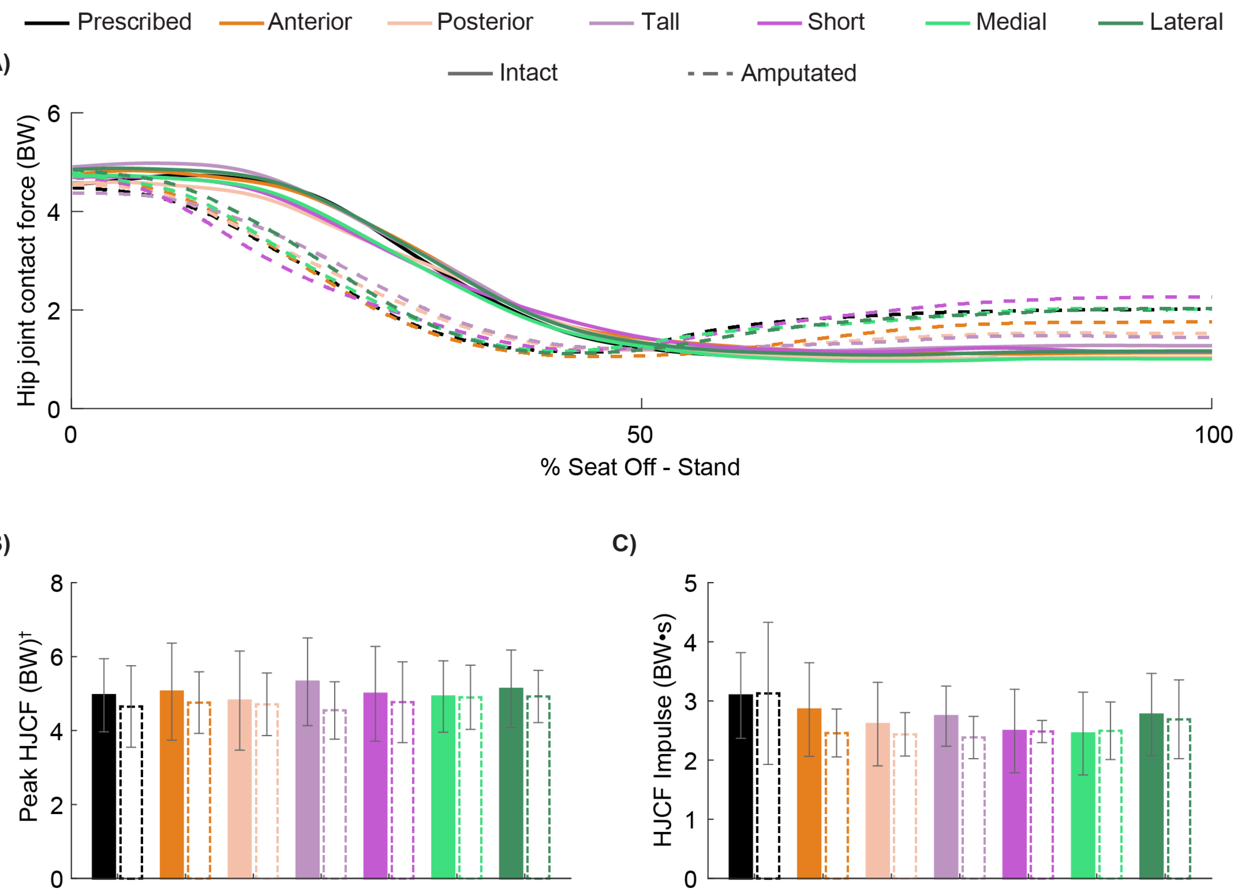

Figure 5.2. A) Average HJCF magnitude across the seat-off-stand motion for all alignments. B) Average peak intact and amputated side HJCF (BW) for all alignments. C) Average intact and amputated side HJCF impulse (BW·s) for all alignments. Error bars represent ± 1 standard deviation. †Significant main effect of side.

Hip joint contact impulse

Within the TTA group, there were no significant main effects of alignment (p = 0.948) or side (p = 0.102), or alignment × side interaction (p = 0.632) for HJCF impulse (Figure 5.2).

When comparing between groups, there were no significant group (p = 0.509), side (p = 0.894), or group × side (p = 0.974) effects for HJCF impulse (Figure 5.3). The co-variate of task duration was significant for both comparisons (p < 0.001).

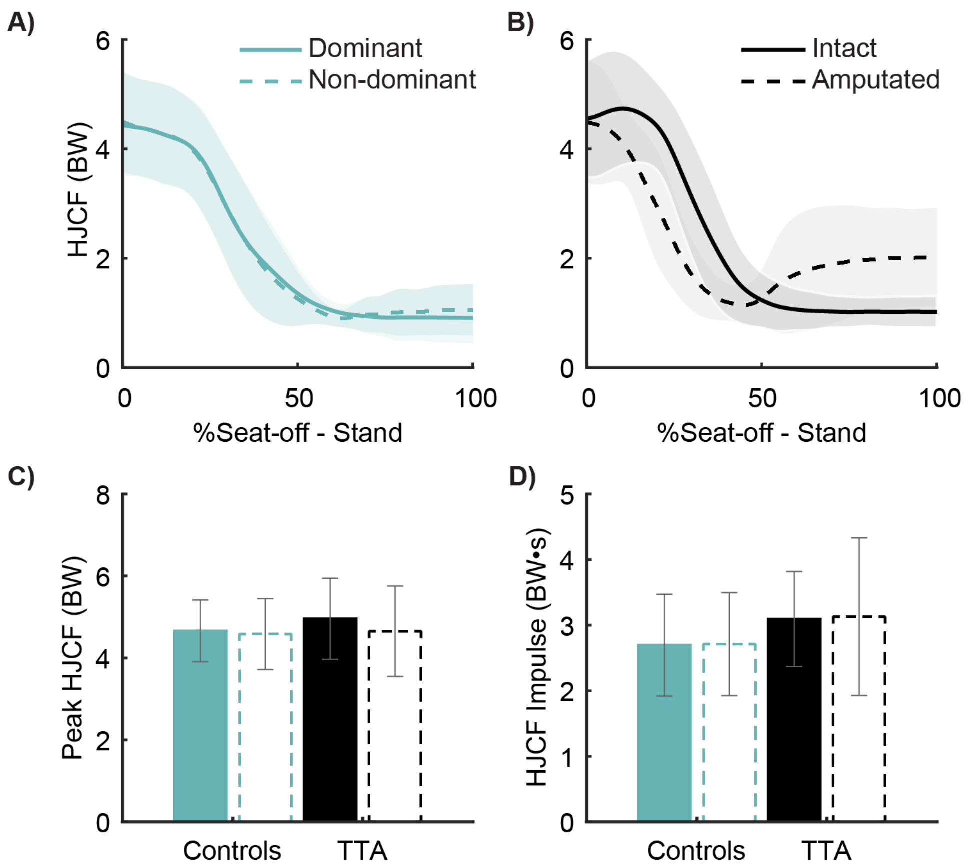

Figure 5.3. Average HJCF (BW) across the seat-off-stand motion for A) controls and B) participants with TTA. C) Average peak HJCF (BW) for controls and participants with TTA. D) Average HJCF impulse (BW·s) for controls and participants with TTA. Error bars represent ± 1 standard deviation.

Peak L4-L5 joint contact forces

There was no significant main effect of alignment for the peak L4-L5 joint contact force (LBCF) ($p = 0.728$; Figure 5.4). There was also no significant difference in peak LBCF between groups ($p = 0.071$).

L4-L5 joint contact impulse

There was no significant main effect of alignment ($p = 0.142$) for the LBCF impulse Similarly, LBCF impulse did not have a significant difference between groups ($p = 0.111$). Task duration was a significant co-variate in both comparisons ($p < 0.001$).

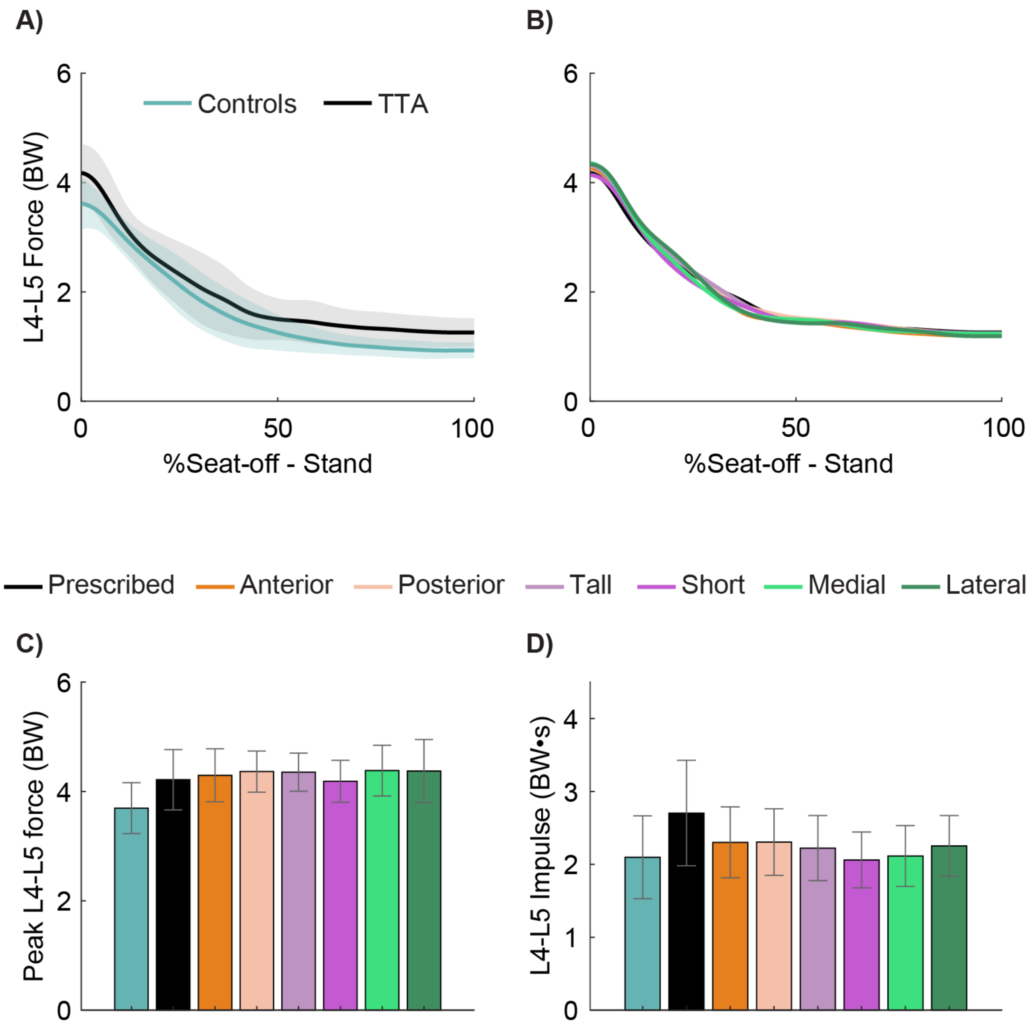

Figure 5.4. L4-L5 joint contact force magnitude during seat-off-to-stand for A) controls and participants with TTA and B) all alignment conditions. C) Peak L4-L5 joint contact forces and D) L4-L5 impulse. Error bars represent ± 1 standard deviation.

5.5 Discussion

The purpose of this study was to determine the effect of prosthetic alignment on hip and low-back joint contact forces during sit-to-stand in people with TTA. Contrary to our expectations, prosthetic alignment did not alter the joint contact forces at the intact and amputated side hips or the L4-L5 joint. Specifically, the peak hip joint contact force (HJCF) and peak L4-L5 joint contact force (LBCF) right after seat-off were similar across alignment conditions. While the average

HJCF and LBCF impulses are 0.23 BWs and 0.4 BWs are respectively greater when the sit-to-stand task was completed with the prescribed alignment compared to the other alignments (Figure 5.2-C and Figure 5.4-D), this difference was due to the longer duration of the sit-to-stand task with the prescribed alignment. As such, when accounting for the duration of the task, prosthetic alignment did not affect HJCF or LBCF impulses statistically.

The findings of this study contradict a prior study on the effect of alignment during walking. In that study, shortening the length of the pylon by 1% of the original pylon length decreased hip intersegmental force asymmetry and resulted in low L5-S1 intersegmental forces during walking (Yu et al., 2014). However, the average differences found in intersegmental forces during walking between different pylon lengths were quite small (< 0.03 BW for hip, < 0.05 BW for L5/S1 joints) (Yu et al., 2014). While it is unclear whether these differences during walking were statistically significant, our similar results of an average difference of 0.04 BW for hip and 0.03 BW for L4-L5 peak joint contact forces between the prescribed and short alignments during sit-to-stand were not significant. As such, it is likely that vertical changes in prosthetic alignment do not affect hip and low-back joint forces during walking or sit-to-stand. However, it is unclear whether these differences in forces are clinically significant.

Across all alignments we found a greater peak HJCF on the intact side compared to the amputated side (Figure 5.2-B). This result is in agreement with previous studies which found greater intact leg HJCF during walking (Sharifmoradi et al., 2017) and running (Sepp et al., 2020) in people with TTA. The altered muscle properties at the hip joints in people with TTA likely contribute to this difference in HJCF between intact and amputated side hips. For example, people with TTA have decreased amputated side hip abductor and quadricep muscles strength (Hewson et al., 2020; Nolan, 2009, 2012), which may contribute to the greater reliance on the intact leg across different activities. In addition, people with TTA have decreased gluteus medius activity at the beginning of sit-to-stand due to shifting the body center of mass toward the intact side (Wagner et al., 2020). Moreover, the amputated side gluteus medius activity increases as people with TTA shift their weight back to the amputated side while completing the sit-to-stand task (Wagner et al., 2020). The amputated side rectus femoris also remains active at the end of the sit-to-stand motion whereas the intact side does not (Wagner et al., 2020). This altered hip muscle activity during sit-to-stand supports our simulation results as the amputated side HJCF increased at the end of the sit-stand motion while the intact side HJCF plateaued (Figure 5.3-B).

There were no differences in hip and low-back joint peak forces or impulse between people with TTA and non-amputees. While some differences in HJCF between people with and without TTA have been found during running (Sepp et al., 2020), most studies in this population have only focused on the differences between intact and amputated sides (Sharifmoradi et al., 2017; Yu et al., 2014). At the low-back, previous studies have found significantly greater L4/L5 compression loads in people with TTA compared to non-amputees during walking (Yoder et al., 2015) and sit-to-stand (Actis et al., 2018b). While the LBCF in our study only trended toward significance ($p = 0.071$), the difference in average LBCF between participants with TTA and controls in our study (0.52 BW) was similar to the difference in axial L4/L5 compression force between people with TTA and non-amputees (0.57 BW) (Actis et al., 2018b). As we compared the vector magnitude of LBCF instead of the axial compression force, differences in shear forces between people with and without TTA may contribute to greater variability within groups in our study. While it is unclear if these differences between groups is clinically significant, the combination of higher peak

LBCF and greater trunk movement during sit-to-stand (Actis et al., 2018b; Nolasco et al., 2020) may be a risk factor for people with TTA developing low-back pain.

Our estimated joint contact force magnitudes were greater than those of previous experimental and modeling studies. *In vivo* measurements of average peak HJCF during sit-to-stand ranged between 1.90- and 3.91-times body weight (Bergmann et al., 2016; Bergmann et al., 2001; Stansfield et al., 2003) while our estimates were approximately 4.6-times body weight for controls and 4.95-times body weight on the intact side for participants with TTA. The greater HJCFs estimated here may be due to the imposed sit-to-stand strategy. Here, participants were asked to complete the task with arms crossed over their chest. Given this requirement, our participants may have used greater trunk and hip flexion strategy to complete the task. Prior studies have found that peak HJCF increases with greater hip flexion during sit-to-stand (Inai et al., 2018). For the low-back, our average peak LBCF of 3.69 BW for controls and 4.21 BW for participants with TTA include the shear forces at the L4/L5 joint which explain why they are slightly greater than L4-L5 axial compression forces previously estimated (controls: 3.41 BW; TTA: 3.98 BW) (Actis et al., 2018b). While there are discrepancies in joint contact force magnitudes between studies, our modeling approach was consistent across groups and alignment conditions, which resulted in consistent loading patterns between groups and within participants, respectively.

One limitation of the modeling approach to estimate joint loading was the differences between measured EMG and model muscle activations, particularly for the rectus femoris and medial gastrocnemius (Figure 5.1). This discrepancy may be caused by the high degree of hip and knee flexion during sit-to-stand. As the lower limbs of the model were developed for simulating walking gait, the moment arms of these musculotendon actuators may not represent realistic muscle line of actions during the sit-to-stand movement. Even during the seat-off-stand portion simulated in this study which begins at a less flexed position, it was visually apparent in the OpenSim GUI that the rectus femoris and gastrocnemius musculotendon actuators passed through the bone geometries. Future work may mitigate this effect by incorporating wrapping surfaces for these muscles during movements with high degrees of hip and knee flexion (Catelli et al., 2019; Lai et al., 2017). Despite this limitation, major contributors to HJCF (i.e., gluteus medius) and LBCF (i.e., thoracic and lumbar paraspinals) had good visual agreement between model activations and EMG across all alignments, giving us confidence in our simulation results. Another limitation is the modeling of the same prescribed alignment of the prosthesis for all participants. As each participant has a different prescribed alignment, the force estimates may not be representative of the actual forces experienced by each individual. However, we also adjusted the position of the foot relative to the socket in the model for each alignment condition by the same magnitude of the alignment change made experimentally. As such, our results still provide insight on the effect of prosthetic alignment on the joint contact forces.

5.6 Conclusion

Acute changes in translational prosthetic alignment did not affect peak HJCF or LBCF during sit-to-stand in our participants with TTA. These results suggest that acute changes in prosthetic alignment that may occur for short durations due to factors such as residual limb volume changes and footwear, are not likely to affect hip and low-back joint loading during sit-to-stand. The greater intact side peak HJCF compared to the amputated side supports prior work that found similar results during other functional tasks of daily living. As such, the cumulative asymmetric loading between legs across different activities during daily life may contribute to the development of hip joint pain. Future work should investigate the effect of prosthetic alignment during tasks such as turning, walking up and down stairs, and transitioning between tasks as these tasks can have a greater duration of joint loading, force generation, and/or body support on each leg separately.

CHAPTER 6. Discussion

Living an independent lifestyle requires the ability to perform various functional tasks throughout daily life, such as rising from a chair or turning a corner. For people with a unilateral transtibial amputation (TTA), the missing ankle muscle function and associated increased reliance on the intact leg can affect their ability to remain functionally mobile. Existing literature has primarily focused on characterizing the altered movement mechanics during straight-line gait, with only a few studies investigating the movement strategies used by people with TTA during turning (Golyski and Hendershot, 2018; Orendurff et al., 2006; Segal et al., 2011; Shell et al., 2017; Ventura et al., 2011) and seat transfers (Agrawal et al., 2011; Agrawal et al., 2016; Ferris et al., 2017a; Ozyurek et al., 2014; Slajpah et al., 2013). This dissertation addressed this important gap in the literature by investigating how people with TTA maintain dynamic balance during 90-degree turns (Chapter 2) and seat transfers (Chapter 3).

Functional mobility can also be affected by the alignment of the prosthesis. Previous literature has primarily focused on the effect of prosthetic alignment on straight-line gait despite functional mobility requiring the ability to perform all types of tasks including seat transfers. Chapter 3-5 addressed this important gap in the literature by focusing on understanding the effect of prosthetic alignment on the strategies used by people with TTA to maintain dynamic balance and how joint loading is affected during seat transfer tasks. The results of this work also improved our understanding of how people with TTA adapt to the missing ankle function during different functional tasks in several ways. First, our findings demonstrated that people with TTA alter their generation and regulation of $\vec{H}_{WB}$ during functional tasks which has implications for balance control and the ability to complete these tasks. Second, hip joint loading is asymmetric during sit-to-stand supporting previous research on other tasks that reported greater intact side joint loading. Lastly, highly functional people with TTA are able to adjust to acute changes in prosthetic alignment such that the strategies used with the different alignments do not affect dynamic balance or joint loading.

As the function of ankle muscles are important for balance, a TTA can affect a person's ability to maintain dynamic balance during daily life. Whole-body angular momentum ($\vec{H}_{WB}$) is one measure that can be used to describe dynamic balance during a movement. More specifically, the range of $\vec{H}_{WB}$ is used to describe how well angular momentum is regulated where a larger range suggests less regulation and thus, greater risk of being unable to counteract angular momentum generated in one direction. Based on this measure, previous literature has demonstrated that people with TTA have a decreased ability to maintain balance during walking on level ground (D'Andrea et al., 2014; Silverman and Neptune, 2011), on inclines (Pickle et al., 2016), and stairs (Pickle et al., 2014) compared to non-amputees. Our results in Chapters 2 and 3 extend this knowledge by demonstrating that having a TTA also affects dynamic balance during turning and sit-to-stand tasks. In Chapter 2 we found that turning with the prosthesis on the inside of the turn resulted in a greater range of $\vec{H}_{WB}$ compared to turning with the prosthesis on the outside of the turn and compared to non-amputees. We found this was due to the greater angular momentum contribution from the intact leg and trunk which was likely related to the lack of impulse of the prosthetic ankle during push-off and low confidence in body support on the amputated leg. Similarly, the greater reliance on the intact leg for body weight support and associated altered trunk motion during sit-to-stand contributes to the greater range of $\vec{H}_{WB}$ in people with TTA compared to non-amputees found in Chapter 3. As the trunk is the largest body segment, trunk adjustments to control balance may be the primary compensatory mechanism to maintain dynamic balance for people with TTA. Thus, to create rehabilitation protocols that improve dynamic balance in functional tasks it is important to understand what factors affect trunk compensations in people with TTA. For example, previous work found that the forces generated by a robotic prosthesis that mimics soleus muscle forces can contribute to anterior trunk angular momentum (D'Andrea et al., 2014; Pickle et al., 2017b), thereby decreasing the range of whole-body angular momentum when walking on inclines compared to a passive prosthesis. As such, future investigations should focus on understanding which prosthesis-related factors may affect how people with TTA maintain dynamic balance during daily life.

In Chapter 4, we found that prosthetic alignment affected how the force was transferred from the ground through the rest of the body during sit-to-stand. Given this, we expected prosthetic alignment to be a factor that affected $\vec{H}_{WB}$ during functional tasks. In Chapter 3, we found small differences in $\vec{H}_{WB}$ with the anterior and posterior alignments during sit-to-stand which were supported by the differences in anterior/posterior ground reaction forces with these alignments in Chapter 4. However, these acute changes in prosthetic alignment did not affect $\vec{H}_{WB}$ for other seat-transfer tasks. During sit-to-stand, the differences in $\vec{H}_{WB}$ were small in magnitude and it is unclear if they are clinically significant. While studies in other populations have demonstrated a relationship between $\vec{H}_{WB}$ and clinical balance measures, more studies are needed to fully understand and identify magnitude thresholds for the relationship between angular momentum and balance deficiencies.

The participants for these studies had a Medicare functional classification level of K3 or K4 indicating high level of functional mobility. Greater function is typically associated with greater strength and agility. This participant characteristic likely contributed to the small differences between alignment conditions, as participants were able to use sufficient compensatory strategies to minimize the effect of the alignment change on $\vec{H}_{WB}$ in Chapter 3 and joint loading in Chapter 5. The magnitude of the alignment changes used for these studies were chosen to be within an acceptable range previously reported in the literature (Zahedi, 1986). In addition, these changes are also representative of changes that may occur in daily life due to pistoning of the residual limb within the prosthetic socket, residual limb volume fluctuations, and/or footwear form factor. While larger changes in prosthetic alignment outside of an acceptable range may have greater effects on the strategies used to complete functional tasks, the small changes in alignment made in these studies were detectable by the participants. As these changes in alignment were made acutely, participants were likely able to adjust how they complete the different functional tasks based on how the prosthesis felt such that dynamic balance and joint loading were not affected. Therefore, future work should investigate how long-term changes in alignment may affect the strategies used to complete functional tasks.

Joint and low-back pain are experienced by people with TTA at greater rates compared to the general population making it important to understand which factors contribute to these secondary conditions. Pain is multifactorial and can be caused by interrelated social, biological, psychological, and mechanical factors (Sivapuratharasu et al., 2019). As people with TTA primarily rely on the intact leg which leads to asymmetric and altered movement patterns, mechanical factors are one of the greatest contributors to pain development in this population

(Devan et al., 2015; Gailey et al., 2008; Norvell et al., 2005; Wasser et al., 2020). Specifically, greater intact leg loading is thought to be associated with a greater risk for intact knee and hip joint pain and degeneration in people with TTA. Supporting this idea, previous studies have found that people with TTA have greater intact side hip and knee contact forces during walking (Sharifmoradi et al., 2017) and running (Sepp et al., 2020). Our results in Chapter 5 extends this knowledge as we found people with TTA also have a greater intact side hip joint contact force magnitude compared to the amputated side during the sit-to-stand task regardless of prosthetic alignment. Thus, the overall cumulative intact side joint loading during various activities of daily life likely contributes to hip joint pain in people with TTA. To avoid experiencing pain, people with joint pain typically avoid physical activity which contributes to functional limitations and leads to reduced community participation over time. In addition, the altered loading patterns and reduced physical activity may contribute to the development of joint osteoarthritis in the long-term affecting quality of life. Improving amputated side hip muscle strength is one idea that may have the potential to improve asymmetric loading for people with TTA. However, while improving amputated side muscle strength can increase physical activity participation in people with TTA (Nolan, 2012), active people with TTA who are likely to have less asymmetry in hip muscle strength (Nolan, 2009) still have asymmetric joint moments during walking (Nolan and Lees, 2000) and joint contact forces during running (Sepp et al., 2020). This is likely due to the altered hip muscle activity used to compensate for the lack of ankle muscle function on the amputated side. Thus, improving joint loading asymmetry may require the development of a prosthesis that mimics the function of ankle muscles during different activities of daily living. In addition, sensory feedback could also enable more confidence in the prosthesis for weight-bearing throughout different activities. Future studies should investigate which ankle muscle functions are important for completing different tasks of daily living and whether a prosthesis with those functions can improve joint loading asymmetries in people with TTA.

Greater rates of low-back pain in people with TTA are also likely related to the compensatory trunk motion associated with asymmetric loading strategies. Previous studies have identified greater lateral trunk lean and axial rotation toward the intact leg in people with TTA as a strategy to place more weight on the intact leg for body support during walking (Yoder et al., 2015) and sit-to-stand (Actis et al., 2018b) tasks. Our results in Chapter 4 provide further support for this strategy during sit-to-stand in people with TTA compared to an age- and height- matched control group. In Chapter 4 we also found that a posterior shift of the prosthetic foot relative to the socket reduced trunk range of motion in people with TTA, but that did not result in a reduced low-back joint contact force magnitude in Chapter 5. While these results could suggest that altered trunk motion may be a small contributor to developing low-back pain in people with TTA, it is possible that the difference in trunk range of motion was not large enough to affect low-back joint loading during sit-to-stand. Furthermore, low-back pain is multifactorial and can be affected by several biomechanical, psychosocial, and personal factors (Sivapuratharasu et al., 2019) which suggests that the effect of prosthetic alignment on the strategy used to complete a task when dealing with low-back pain likely differs on an individual basis. As such, it may be necessary to focus on an individual's characteristics and movement strategies during daily life to evaluate the risk of developing low-back pain and other secondary conditions.

6.1 Limitations

There are several limitations of the studies described herein. First, we included a relatively small number of participants who all had high functional mobility and were mostly male. We specifically recruited high functioning individuals due to the demanding nature of the study protocol which lasted an average of 5 hours. The results may not generalize to individuals that have greater trouble with maintaining balance and for whom prosthetic alignment may have larger effects on their movement strategies. In addition, the male-to-female ratio of participants (9:1) does not represent the population ratio for people with TTA. While there may be differences in movement strategies between males and females, the compensatory strategies of asymmetric loading and altered trunk kinematics was similar across all participants in these studies. However, more work is needed to fully understand sex and functional level differences in movement strategies across activities of daily living for people with TTA. Another limitation was that the tasks with the prescribed alignment were always performed before other conditions for all participants. With the prescribed alignment being the first exposure to the functional tasks during the data collection session, participants likely learned and got more comfortable with the tasks as the data collection progressed. While this can result in faster task completion times, this was only observed during the sit-to-stand and stand-to-sit tasks while all other tasks had similar completion times across alignment conditions. As such, further research is needed to determine if this time effect is purely due to the effect of prosthetic alignment, or if other factors such as the short acclimation to the different alignments and/or variability in movement strategies also contributed. The use of each participants own prosthesis is also a limitation of these studies. Each individual participated in this study with their own prescribed prosthesis, which had different components and alignment compared to other participants. While this ensured that the prescribed condition represented how each individual completed all tasks during daily life, it likely contributed to the high between-subject variability seen in each aim. To mitigate this effect, all comparisons made were within-participant.

6.2 Conclusion

This dissertation demonstrates that highly functional people with TTA have decreased balance control and asymmetric hip joint loading across a range of functional tasks. The greater intact leg joint loading found during sit-to-stand may contribute to joint loading asymmetries previously reported during other functional tasks which have implications for the development of joint pain and degeneration in people with TTA. However, more research on long-term cumulative joint loading is needed to fully understand the risks associated with asymmetric loading strategies. Results from this dissertation also suggest that the compensatory strategies used with acute changes in prosthetic alignment enable highly functional people with TTA to maintain consistent balance control and joint loading. Future work should investigate whether these findings extend to people with TTA who may have greater difficulty adjusting to the sudden changes in prosthetic alignment. In addition, future research is needed to understand how long-term changes in prosthetic alignment affect the strategies used to complete functional tasks throughout daily life.

Appendices

Appendix A. Supplementary material for Chapter 2

Table A.1. Model fit statistics for the linear mixed models used for the range of whole-body angular momentum in each plane

DV	Effects	Estimate	Std. Error	Lower 95% CI	Upper 95% CI
Frontal	Intercept	0.047	0.004	0.039	0.055
	Group TTA	0.015	0.005	0.004	0.026
	Turn Outside	-0.002	0.001	-0.005	0.001
	Group TTA : Turn Outside	-0.011	0.002	-0.015	-0.007
Sagittal	Intercept	0.053	0.003	0.047	0.058
	Group TTA	0.018	0.004	0.01	0.026
	Turn Outside	-0.005	0.001	-0.007	-0.002
	Group TTA : Turn Outside	-0.015	0.002	-0.019	-0.012
Transverse	Intercept	0.015	0.001	0.013	0.017
	Group TTA	0.002	0.001	0.000	0.005
	Turn Outside	0.001	0.001	-0.001	0.002
	Group TTA : Turn Outside	-0.003	0.001	-0.005	-0.001

Table A.2. Model fit estimates for the positive contributions for the six segment groups in each plane of motion

Segment group by plane	Effects	Estimate	Std. Error	Lower 95% CI	Upper 95% CI
Head/Trunk					
Frontal	Intercept	21.001	2.836	15.102	26.9
	Group TTA	2.029	4.011	-6.313	10.372
	Turn Outside	-3.118	0.743	-4.585	-1.651
	Group TTA : Turn Outside	7.047	1.051	4.973	9.122
Sagittal	Intercept	7.287	0.632	5.981	8.593
	Group TTA	2.755	0.894	0.908	4.602
	Turn Outside	-0.265	0.359	-0.973	0.443
	Group TTA : Turn Outside	-3.825	0.507	-4.826	-2.824
Transverse	Intercept	23.962	1.138	21.595	26.33
	Group TTA	-0.667	1.61	-4.015	2.68
	Turn Outside	0.151	0.317	-0.476	0.777
	Group TTA : Turn Outside	-0.065	0.449	-0.951	0.821
Inside Arm					
Frontal	Intercept	4.874	0.497	3.846	5.901
	Group TTA	0.104	0.702	-1.349	1.558
	Turn Outside	-0.384	0.255	-0.887	0.119
	Group TTA : Turn Outside	0.669	0.361	-0.042	1.38
Sagittal	Intercept	2.532	0.191	2.135	2.93
	Group TTA	0.097	0.271	-0.465	0.659
	Turn Outside	0.141	0.071	0.002	0.281
	Group TTA : Turn Outside	0.376	0.1	0.179	0.573
Transverse	Intercept	11.623	0.364	10.871	12.375
	Group TTA	-0.231	0.515	-1.295	0.833
	Turn Outside	-0.806	0.205	-1.21	-0.402
	Group TTA : Turn Outside	1.815	0.29	1.243	2.387
Inside Leg					
Frontal	Intercept	26.902	2.442	21.845	31.958
	Group TTA	-2.928	3.453	-10.079	4.223
	Turn Outside	0.232	1.186	-2.108	2.572
	Group TTA : Turn Outside	-1.006	1.677	-4.315	2.303
Sagittal	Intercept	35.184	1.041	33.03	37.338
	Group TTA	-7.254	1.472	-10.3	-4.208
	Turn Outside	0.113	0.538	-0.948	1.174
	Group TTA : Turn Outside	11.053	0.76	9.552	12.553
Transverse	Intercept	6.562	0.652	5.216	7.907
	Group TTA	-0.878	0.923	-2.781	1.024

Segment group by plane	Effects	Estimate	Std. Error	Lower 95% CI	Upper 95% CI
	Turn Outside	-0.398	0.417	-1.221	0.425
	Group TTA : Turn Outside	0.039	0.59	-1.125	1.203
Outside Arm					
Frontal	Intercept	2.806	0.473	1.824	3.788
	Group TTA	0.805	0.669	-0.584	2.193
	Turn Outside	0.196	0.175	-0.149	0.541
	Group TTA : Turn Outside	1.343	0.247	0.855	1.831
Sagittal	Intercept	0.346	0.094	0.152	0.54
	Group TTA	0.046	0.132	-0.229	0.32
	Turn Outside	-0.102	0.038	-0.177	-0.028
	Group TTA : Turn Outside	-0.134	0.053	-0.239	-0.029
Transverse	Intercept	13.142	0.472	12.171	14.113
	Group TTA	0.528	0.668	-0.846	1.902
	Turn Outside	0.351	0.333	-0.306	1.008
	Group TTA : Turn Outside	1.619	0.471	0.69	2.549
Outside Leg					
Frontal	Intercept	43.162	2.849	37.261	49.063
	Group TTA	-0.175	4.028	-8.52	8.171
	Turn Outside	3.182	1.338	0.542	5.822
	Group TTA : Turn Outside	-8.435	1.892	-12.169	-4.702
Sagittal	Intercept	54.351	0.843	52.611	56.09
	Group TTA	4.244	1.193	1.783	6.704
	Turn Outside	0.121	0.525	-0.916	1.157
	Group TTA : Turn Outside	-7.522	0.743	-8.988	-6.055
Transverse	Intercept	37.715	1.529	34.542	40.887
	Group TTA	0.829	2.163	-3.657	5.315
	Turn Outside	0.519	0.63	-0.723	1.761
	Group TTA : Turn Outside	-4.017	0.89	-5.773	-2.26
Pelvis					
Frontal	Intercept	1.256	0.172	0.897	1.614
	Group TTA	0.165	0.244	-0.342	0.671
	Turn Outside	-0.108	0.062	-0.229	0.014
	Group TTA : Turn Outside	0.382	0.087	0.211	0.554
Sagittal	Intercept	0.301	0.054	0.188	0.413
	Group TTA	0.113	0.076	-0.046	0.272
	Turn Outside	-0.007	0.017	-0.041	0.027
	Group TTA : Turn Outside	0.052	0.024	0.004	0.1
Transverse	Intercept	6.996	0.444	6.072	7.921
	Group TTA	0.42	0.628	-0.888	1.727

Segment group by plane	Effects	Estimate	Std. Error	Lower 95% CI	Upper 95% CI
	Turn Outside	0.184	0.079	0.028	0.339
	Group TTA : Turn Outside	0.608	0.112	0.388	0.828

Table A.3. Model fit estimates for the negative contributions for the six segment groups in each plane of motion

Segment Group by plane	Effects	Estimate	Std. Error	Lower 95% CI	Upper 95% CI
Head/Trunk					
Frontal	Intercept	39.827	1.404	36.929	42.725
	Group TTA	-1.957	1.986	-6.055	2.141
	Turn Outside	-1.547	0.865	-3.254	0.16
	Group TTA : Turn Outside	4.585	1.224	2.171	6.999
Sagittal	Intercept	6.121	0.815	4.431	7.811
	Group TTA	1.656	1.153	-0.734	4.046
	Turn Outside	-0.74	0.371	-1.473	-0.008
	Group TTA : Turn Outside	-0.082	0.525	-1.117	0.954
Transverse	Intercept	1.024	0.353	0.299	1.749
	Group TTA	-0.716	0.499	-1.742	0.309
	Turn Outside	-0.943	0.268	-1.472	-0.414
	Group TTA : Turn Outside	0.917	0.379	0.168	1.665
Inside Arm					
Frontal	Intercept	2.263	0.276	1.69	2.835
	Group TTA	0.218	0.39	-0.592	1.027
	Turn Outside	-0.42	0.098	-0.612	-0.227
	Group TTA : Turn Outside	0.731	0.138	0.458	1.003
Sagittal	Intercept	0.024	0.014	-0.005	0.053
	Group TTA	0.028	0.02	-0.013	0.068
	Turn Outside	-0.017	0.009	-0.034	0.001
	Group TTA : Turn Outside	-0.005	0.012	-0.03	0.019
Transverse	Intercept	7.344	1.203	4.871	9.817
	Group TTA	-3.475	1.701	-6.973	0.022
	Turn Outside	-3.971	0.875	-5.697	-2.245
	Group TTA : Turn Outside	3.167	1.237	0.726	5.608
Inside Leg					
Frontal	Intercept	9.957	1.176	7.538	12.376
	Group TTA	-1.192	1.663	-4.614	2.229
	Turn Outside	1.477	0.835	-0.171	3.126
	Group TTA : Turn Outside	2.618	1.182	0.287	4.95
Sagittal	Intercept	50.147	0.517	49.079	51.215
	Group TTA	-5.387	0.732	-6.898	-3.876
	Turn Outside	1.802	0.307	1.196	2.408
	Group TTA : Turn Outside	9.956	0.434	9.099	10.813
Transverse	Intercept	21.914	2.856	16.037	27.79
	Group TTA	-0.235	4.039	-8.545	8.076

Segment Group by plane	Effects	Estimate	Std. Error	Lower 95% CI	Upper 95% CI
	Turn Outside	-5.048	2.003	-9	-1.096
	Group TTA : Turn Outside	14.762	2.833	9.173	20.351
Outside Arm					
Frontal	Intercept	3.971	0.481	2.974	4.968
	Group TTA	-0.36	0.68	-1.77	1.05
	Turn Outside	0.385	0.207	-0.023	0.794
	Group TTA : Turn Outside	0.898	0.293	0.321	1.476
Sagittal	Intercept	2.922	0.339	2.218	3.627
	Group TTA	0.986	0.479	-0.011	1.983
	Turn Outside	0.243	0.102	0.04	0.445
	Group TTA : Turn Outside	-0.679	0.145	-0.965	-0.393
Transverse	Intercept	5.898	1.526	2.745	9.051
	Group TTA	4.093	2.158	-0.366	8.552
	Turn Outside	0.741	0.87	-0.976	2.457
	Group TTA : Turn Outside	-5.124	1.23	-7.552	-2.697
Outside Leg					
Frontal	Intercept	42.074	1.602	38.781	45.368
	Group TTA	2.58	2.266	-2.077	7.238
	Turn Outside	-0.09	1.171	-2.4	2.221
	Group TTA : Turn Outside	-8.832	1.656	-12.099	-5.564
Sagittal	Intercept	40.384	0.83	38.664	42.103
	Group TTA	2.473	1.174	0.042	4.905
	Turn Outside	-1.321	0.384	-2.078	-0.564
	Group TTA : Turn Outside	-9.29	0.542	-10.36	-8.22
Transverse	Intercept	63.59	4.307	54.66	72.52
	Group TTA	0.181	6.091	-12.448	12.81
	Turn Outside	9.399	1.878	5.693	13.105
	Group TTA : Turn Outside	-13.656	2.656	-18.897	-8.415
Pelvis					
Frontal	Intercept	1.908	0.164	1.567	2.249
	Group TTA	0.711	0.232	0.229	1.193
	Turn Outside	0.193	0.064	0.067	0.32
	Group TTA : Turn Outside	-0.001	0.09	-0.179	0.178
Sagittal	Intercept	0.402	0.059	0.28	0.523
	Group TTA	0.244	0.083	0.072	0.416
	Turn Outside	0.033	0.029	-0.025	0.091
	Group TTA : Turn Outside	0.1	0.042	0.018	0.182
Transverse	Intercept	0.23	0.159	-0.098	0.558
	Group TTA	0.152	0.224	-0.312	0.616

Segment Group by plane	Effects	Estimate	Std. Error	Lower 95% CI	Upper 95% CI
	Turn Outside	-0.178	0.083	-0.342	-0.014
	Group TTA : Turn Outside	-0.066	0.118	-0.298	0.166

Table A.4. Statistical results and 95% confidence intervals for the range of whole-body angular momentum in each plane of motion during the continuation stride

	Prosthesis Inside vs. Prosthesis Outside		Prosthesis Outside vs. Non-Dominant Outside		Prosthesis Inside vs. Non-Dominant Inside		Non-Dominant Inside vs. Non-Dominant Outside	
	p	95% CI	p	95% CI	p	95% CI	p	95% CI
Range of $\vec{H}_{WB}$								
Frontal	**< 0.001**	[0.0097, 0.0172]	0.934	[-0.0187, 0.0112]	**0.047**	[-0.0301, -0.0014]	0.502	[-0.0016, 0.0058]
Sagittal	**< 0.001**	[0.0169, 0.0227]	0.921	[-0.0128, 0.0074]	**< 0.001**	[-0.0281, -0.0079]	**< 0.001**	[0.0016, 0.0075]
Transverse	**0.003**	[0.0007, 0.0043]	0.981	[-0.0029, 0.0041]	0.258	[-0.0059, 0.0011]	0.866	[-0.0024, 0.0012]

Table A.5. Statistical results and 95% confidence intervals for the segment group positive and negative contributions (%) to whole-body angular momentum during the continuation stride

	Prosthesis Inside vs. Prosthesis Outside		Prosthesis Outside vs. Non-Dominant Outside		Prosthesis Inside vs. Non-Dominant Inside		Non-Dominant Inside vs. Non-Dominant Outside	
	p	95% CI	p	95% CI	p	95% CI	p	95% CI
Positive Segment contribution during the continuation stride								
Head/Trunk								
Frontal	**< 0.001**	[-5.80, -2.05]	0.141	[-20.28, -2.12]	0.979	[-13.23, 9.17]	**< 0.001**	[1.24, 4.99]
Sagittal	**< 0.001**	[3.19, 5.00]	0.677	[-1.39, 3.53]	**0.024**	[-5.21, -0.30]	0.915	[-0.64, 1.17]
Transverse	0.998	[-0.886, 0.715]	0.986	[-3.76, 5.22]	0.990	[-3.83, 5.16]	0.982	[-0.95, 0.65]
Pelvis								
Frontal	**< 0.001**	[-0.43, -0.12]	0.146	[-1.23, 0.13]	0.942	[-0.84, 0.51]	0.292	[-0.05, 0.26]
Sagittal	**0.039**	[-0.09, -0.001]	0.169	[-0.38, 0.05]	0.497	[-0.33, 0.10]	0.989	[-0.04, 0.05]
Transverse	**< 0.001**	[-0.99, -0.59]	0.403	[-2.79, 0.73]	0.944	[-2.18, 1.34]	0.083	[-0.38, 0.016]
Inside Arm								
Frontal	0.710	[-0.93, 0.36]	0.739	[-2.71, 1.17]	0.999	[-2.04, 1.83]	0.438	[-0.26, 1.03]
Sagittal	**< 0.001**	[-0.32, 0.037]	0.340	[-1.23, 0.28]	0.994	[-0.85, 0.66]	0.176	[-0.32, 0.037]
Transverse	**< 0.001**	[-1.53, -0.49]	**0.025**	[-3.00, -0.17]	0.986	[-1.19, 1.65]	**< 0.001**	[0.29, 1.65]
Outside Arm								
Frontal	**< 0.001**	[-1.98, -1.10]	**0.020**	[-4.01, -0.29]	0.676	[-2.66, 1.06]	0.705	[-0.64, 0.24]
Sagittal	**< 0.001**	[-0.14, 0.33]	0.943	[-0.28, 0.46]	0.995	[-0.41, 0.32]	**0.029**	[0.01, 0.20]
Transverse	**< 0.001**	[-2.81, -1.13]	**0.017**	[-3.97, -0.33]	0.900	[-2.35, 1.29]	0.751	[-1.19, 0.50]
Inside Leg								
Frontal	0.945	[-2.22, 3.77]	0.715	[-5.61, 13.48]	0.877	[-6.62, 12.47]	0.999	[-3.22, 2.76]
Sagittal	**< 0.001**	[-12.52, -9.81]	0.073	[-7.86, 0.26]	**< 0.001**	[3.19, 11.32]	0.999	[-1.47, 1.24]
Transverse	0.862	[-0.69, 1.41]	0.847	[-1.69, 3.37]	0.825	[-1.65, 3.40]	0.812	[-0.65, 1.45]
Outside Leg								
Frontal	**< 0.001**	[1.88, 8.63]	0.174	[-2.54, 19.76]	0.999	[-10.97, 11.32]	0.073	[-6.56, 0.19]
Sagittal	**< 0.001**	[6.08, 8.73]	**0.049**	[0.01, 6.55]	**0.008**	[-7.51, -0.97]	0.999	[-1.45, 1.20]

	Prosthesis Inside vs. Prosthesis Outside		Prosthesis Outside vs. Non-Dominant Outside		Prosthesis Inside vs. Non-Dominant Inside		Non-Dominant Inside vs. Non-Dominant Outside	
	p	95% CI	p	95% CI	p	95% CI	p	95% CI
Transverse	**< 0.001**	[1.91, 5.09]	0.498	[-2.81, 9.19]	0.993	[-6.83, 5.17]	0.880	[-2.11, 1.07]
Negative Segment contribution during the continuation stride								
Head/Trunk								
Frontal	**0.002**	[-5.22, -0.86]	0.593	[-8.08, 2.82]	0.806	[-3.49, 7.40]	0.271	[-0.64, 3.73]
Sagittal	0.109	[-0.11, 1.68]	**0.568**	[-4.77, -1.62]	**0.522**	[-4.85, -1.54]	0.178	[-0.20, 1.68]
Transverse	0.999	[-0.65, 0.70]	0.991	[-1.56, 1.15]	0.516	[-0.64, 2.07]	0.002	[-0.65, 0.70]
Pelvis								
Frontal	0.012	[-0.35, -0.03]	0.028	[-1.36, -0.06]	0.028	[-1.36, -0.07]	0.012	[-0.35, -0.03]
Sagittal	< 0.001	[-0.21, -0.06]	0.002	[-0.57, -0.11]	0.034	[-0.47, -0.01]	0.698	[-0.11, 0.04]
Transverse	0.016	[0.03, 0.45]	0.992	[-0.71, 0.53]	0.940	[-0.77, 0.47]	0.129	[-0.03, 0.39]
Inside Arm								
Frontal	0.007	[-0.56, -0.06]	0.101	[-2.03, 0.14]	0.970	[-1.30, 0.87]	< 0.001	[-0.17, 0.67]
Sagittal	0.049	[0.0001, 0.04]	0.727	[-0.08, 0.03]	0.546	[-0.08, 0.03]	0.212	[-0.01, 0.04]
Transverse	0.832	[-1.40, 3.01]	0.999	[-4.32, 4.94]	0.198	[-1.15, 8.10]	< 0.001	[1.76, 6.18]
Outside Arm								
Frontal	< 0.001	[-1.81, -0.76]	0.901	[-2.42, 1.35]	0.975	[-1.53, 2.25]	0.235	[-0.91, 0.14]
Sagittal	< 0.001	[0.18, 0.69]	0.951	[-1.64, 1.03]	0.205	[-2.32, 0.35]	0.075	[-0.50, 0.02]
Transverse	< 0.001	[2.19, 6.58]	0.983	[-4.90, 6.97]	0.262	[-10.03, 1.84]	0.867	[-2.94, 1.45]
Inside Leg								
Frontal	**< 0.001**	[-6.20, -1.99]	0.871	[-5.96, 3.11]	0.928	[-3.34, 5.72]	0.281	[-3.58, 0.63]
Sagittal	**< 0.001**	[-12.53, -10.98]	**< 0.001**	[-6.58, -2.56]	**< 0.001**	[3.38, 7.40]	**< 0.001**	[-2.58, -1.03]
Transverse	**< 0.001**	[-14.77, -4.66]	**0.007**	[-25.54, -3.52]	0.999	[-10.78, 11.25]	0.050	[-0.004, 10.10]
Outside Leg								
Frontal	**< 0.001**	[5.97, 11.88]	**0.046**	[0.09, 12.42]	0.712	[-8.74, 3.58]	0.999	[-2.86, 3.04]
Sagittal	**< 0.001**	[9.64, 11.58]	**< 0.001**	[3.57, 10.07]	0.184	[-5.72, -0.78]	**0.003**	[0.35, 2.29]

	Prosthesis Inside vs. Prosthesis Outside		Prosthesis Outside vs. Non-Dominant Outside		Prosthesis Inside vs. Non-Dominant Inside		Non-Dominant Inside vs. Non-Dominant Outside	
	p	**95% CI**	**p**	**95% CI**	**p**	**95% CI**	**p**	**95% CI**
Transverse	0.096	[-0.48, 8.99]	**0.152**	[-3.41, 30.36]	0.999	[-17.07, 16.70]	**< 0.001**	[-14.14, -4.66]

Appendix B. Supplementary material for Chapter 3

Table B.1. Main effect and posthoc results for the range of $\vec{\boldsymbol{H}}_{\boldsymbol{WB}}$. Posthoc pairwise comparisons were only performed when the main effect of alignment was significant.

	Main Effect	Posthoc comparisons to the prescribed alignment		
	p	Comparison	*p*	95% CI
Walking				
Frontal	**0.005**	Anterior	0.158	[-0.0006, 0.0061]
		Posterior	1.000	[-0.0033, 0.0033]
		Medial	0.133	[-0.0005, 0.0060]
		Lateral	0.98	[-0.0024, 0.0040]
		Tall	**0.023**	[0.0003, 0.0068]
		Short	1.000	[-0.0035, 0.0029]
Transverse	0.714	-	-	-
Sagittal	0.176	-	-	-
Sit-to-stand				
Frontal	**0.017**	Anterior	0.986	[-0.0019, 0.0030]
		Posterior	0.978	[-0.0018, 0.0031]
		Medial	0.980	[-0.0018, 0.0030]
		Lateral	**0.037**	[0.0001, 0.0049]
		Tall	0.898	[-0.0034, 0.0016]
		Short	0.500	[-0.0010, 0.0039]
Transverse	0.1392	-	-	-
Sagittal	0.2169	-	-	-
Stand-to-sit				
Frontal	0.278	-	-	-

	Main Effect	Posthoc comparisons to the prescribed alignment		
	p	Comparison	*p*	95% CI
Transverse	0.091	-	-	-
Sagittal	0.153	-	-	-
Sit-to-walk				
Frontal	0.120	-	-	-
Transverse	0.353	-	-	-
Sagittal	0.663	-	-	-
Walk-to-sit				
Frontal	0.580	-	-	-
Transverse	0.412	-	-	-
Sagittal	0.420	-	-	-

Table B.2. Main effect and posthoc results for the number of adjustments to $\vec{\boldsymbol{H}}_{\boldsymbol{WB}}$. Posthoc pairwise comaprisons were only performed when the main effect of alignment was significant.

	Main Effect	Posthoc comparisons to the prescribed alignment		
	p	Comparison	*p*	95% CI
Walking				
Frontal	0.384	-	-	-
Transverse	0.063	-	-	-
Sagittal	0.159	-	-	-
Sit-to-stand				
Frontal	0.061	-	-	-
Transverse	0.689	-	-	-
Sagittal	**0.016**	Anterior	**0.048**	[-2.82, -0.007]
		Posterior	**0.034**	[-2.88, -0.073]
		Medial	0.086	[-2.62, 0.108]
		Lateral	0.980	[-1.72, 1.01]
		Tall	0.999	[-1.62, 1.19]
		Short	0.183	[-2.53, 0.281]
Stand-to-sit				
Frontal	0.095	-	-	-
Transverse	0.516	-	-	-
Sagittal	0.612	-	-	-
Sit-to-walk				
Frontal	0.952	-	-	-
Transverse	0.983	-	-	-
Sagittal	0.181	-	-	-
Walk-to-sit				
Frontal	0.618	-	-	-
Transverse	0.578	-	-	-
Sagittal	0.945	-	-	-

Table B.3. Welch's t-test results for group comparisons for range of $\vec{\boldsymbol{H}}_{\boldsymbol{WB}}$ and number of adjustments to $\vec{\boldsymbol{H}}_{\boldsymbol{WB}}$.

	Range of $\vec{\boldsymbol{H}}_{\boldsymbol{WB}}$		**Number of adjustments to $\vec{\boldsymbol{H}}_{\boldsymbol{WB}}$**	
	p	**95% CI**	***p***	**95% CI**
Walking				
Frontal	0.786	[-0.009, 0.012]	0.826	[-1.519, 1.879]
Transverse	0.640	[-0.005, 0.003]	0.753	[-1.819, 1.339]
Sagittal	< **0.001**	[0.013, 0.023]	0.559	[-0.355, 0.635]
Sit-to-stand				
Frontal	**0.002**	[0.012, 0.007]	**0.015**	[-9.936, -1.224]
Transverse	**0.001**	[0.002, 0.006]	**0.024**	[-11.68, -0.972]
Sagittal	**0.038**	[0.001, 0.016]	0.114	[-5.016, 0.606]
Stand-to-sit				
Frontal	0.086	[-0.001, 0.008]	**0.002**	[-9.544, -2.716]
Transverse	**0.025**	[0.0003, 0.005]	0.081	[-7.974, 0.514]
Sagittal	0.879	[-0.013, 0.011]	**0.011**	[-3.203, -0.486]
Sit-to-walk				
Frontal	0.519	[-0.010, 0.005]	0.134	[-3.741, 0.541]
Transverse	0.827	[-0.003, 0.003]	0.053	[-0.033, 4.633]
Sagittal	0.566	[-0.010, 0.017]	0.261	[-2.266, 0.666]
Walk-to-sit				
Frontal	0.196	[-0.004, 0.020]	0.188	[-1.506, 7.106]
Transverse	0.069	[-0.010, 0.0004]	**0.041**	[0.308, 12.29]
Sagittal	0.752	[-0.011, 0.015]	**0.005**	[1.558, 7.242]

Appendix C. Supplementary material for Chapter 4

C.1 Sit-to-stand Phase Definitions

To better understand the outcome measures in the context of the sit-to-stand motion we defined different phases as “acceleration”, “transition”, and “deceleration” based on the center of mass velocity (Roebroeck et al., 1994). Center of mass motion was defined with respect to the laboratory with the superior, anterior, and lateral directions defined positively. The acceleration phase was defined from the initiation of anterior center of mass velocity until maximum anterior velocity. The transition phase was from maximum anterior center of mass velocity to maximum superior center of mass velocity and finally the deceleration phase was from the maximum superior center of mass velocity to the point where both superior and anterior center of mass velocities were 0.

Table C.1. Average (standard deviation) sit-to-stand duration (seconds) and percent of time spent in each phase for each alignment condition and for Controls.

	Condition			
Duration	**Anterior**	**Posterior**	**Prescribed**	**Control**
Total STS Time (sec)	1.80 (0.21)	1.86 (0.20)	2.00 (0.36)	2.05 (0.39)
Acceleration Time (%)	30.51 (3.73)	31.17 (3.54)	27.26 (4.08)	27.2 (4.25)
Transition Time (%)	21.04 (3.82)	20.40 (5.21)	19.09 (4.54)	20.9 (2.11)
Deceleration Time (%)	48.46 (5.46)	48.43 (7.22)	53.65 (5.85)	51.9 (5.95)

Table C.2. Average (standard deviation) hip, knee, and ankle flexion angles at the initiation of the STS movement for the non-dominant and dominant legs of Controls and the prosthetic and intact legs of people with TTA.

	Condition						Controls	
	Anterior		Posterior		Prescribed			
Angle (º)	**Intact**	**Prosthetic**	**Intact**	**Prosthetic**	**Intact**	**Prosthetic**	**Dominant**	**Non-Dominant**
Hip	68.2 (6.3)	70.5 (7.6)	67.3 (6.0)	70.2 (7.6)	68.9 (5.3)	71.0 (7.9)	68.5 (11.3)	68.5 (11.5)
Knee	88.7 (6.8)†	81.4 (10.3)	87.3 (5.8)†	83.2 (9.6)	86.2 (8.1)†	80.7 (8.8)	91.0 (9.6)†	89.6 (8.7)
Ankle	12.3 (4.7)†	3.4 (4.2)	10.5 (4.5)†	5.8 (4.6)	10.3 (5.1)†	4.3 (4.4) *	10.8 (7.02)	10.01 (6.3)

* significant difference from Controls

† significant difference from Prosthetic/Non-dominant side

Table C.3. Results Summary - Comparisons between people with TTA and Controls

	Group Comparison (Control – TTA)			Leg Comparison (Intact/Dominant – Prosthetic/Non-Dominant)			Group × Leg Interaction					
	p	*95% CI*	*Cohen's d*	*p*	*95% CI*	*Cohen's d*	*p*	*Post-hoc Interactions*		*p*	*95% CI*	*Cohen's d*
<u>ANOVAs</u>												
GRF Impulse												
Braking	**0.006**	[-0.015, -0.003]	-0.80	**0.003**	[-0.019, -0.005]	-1.08	0.056	Intact/ Dominant	Controls - TTA	-	[-0.011, 0.007]	-0.31
								Prosthetic/ Non-Dominant	Controls - TTA	-	[-0.025, -0.007]	-1.45
Propulsion	0.072	[-0.006, 0.0003]	-0.64	0.090	[-0.0005, 0.006]	0.57	0.384	-		-	-	-
Medial	0.383	[-0.054, 0.022]	-0.44	0.141	[-0.004, 0.001]	-0.05	0.588	-		-	-	-
Vertical	0.841	[-0.161, 0.195]	0.09	**0.020**	[0.015, 0.016]	0.46	0.079	Intact/ Dominant	Controls - TTA	-	[-0.235, 0.142]	-0.24
								Prosthetic/ Non-Dominant	Controls - TTA	-	[-0.107, 0.269]	0.42
Knee Joint Moment	**0.009**	[0.064, 0.393]	0.76	**< 0.001**	[0.278, 0.434]	1.33	**< 0.001**	Intact/ Dominant	Controls - TTA	0.306	[-0.268, 0.088]	-0.55
								Prosthetic/ Non-Dominant	Controls - TTA	**< 0.001**	[0.370, 0.725]	2.73
AP Center of Pressure Position	**< 0.001**	[-0.071, -0.029]	-1.37	**< 0.001**	[-0.05, -0.027]	-0.97	**< 0.001**	Intact/ Dominant	Controls - TTA	0.323	[-0.035, 0.012]	-0.55
								Prosthetic/ Non-Dominant	Controls - TTA	**< 0.001**	[-0.111, -0.065]	-3.26
<u>Welch's t-test</u>												
Trunk Range of Motion												
Lateral	**0.003**	[-2.48, 0.594]	-1.64		-				-			
Axial	0.311	[-10.4, 3.51]	-0.49		-				-			
Flexion	**< 0.001**	[-5.38-, 2.12]	-2.39		-				-			

Table C.4. Results Summary - Comparisons between alignments

	Alignment Comparison					Leg Comparison (Prosthetic – Intact)			Alignment × Leg Interaction					
	p	*Pairwise Post-hoc*	*p*	*95% CI*	*Cohen's d*	*p*	*95% CI*	*Cohen's d*	*p*	*Post-hoc Interactions*		*p*	*95% CI*	*Cohen's d*
<u>ANOVAs</u>														
GRF Impulse														
Braking	*0.058*	*Prescribed-Posterior*	0.170	[-0.003, 0.0004]	-0.10	**0.027**	[0.003, 0.031]	1.42	**0.020**	Anterior	Prosthetic - Intact	**0.006**	[0.008, 0.038]	1.93
		Prescribed-Anterior	0.077	[-0.003, 0.0001]	-0.10					Posterior	Prosthetic - Intact	0.181	[-0.006, 0.025]	0.80
		Posterior-Anterior	0.966	[-0.002, 0.002]	-0.02					Prescribed	Prosthetic - Intact	**0.021**	[0.003, 0.033]	1.51
Propulsion	***< 0.001***	*Prescribed-Posterior*	**< 0.001**	[0.001, 0.004]	0.62	0.104	[-0.007, 0.001]	-0.71	0.309	-	-	-	-	-
		Prescribed-Anterior	**< 0.001**	[0.002, 0.005]	0.69									
		Posterior-Anterior	0.948	[-0.001, 0.002]	0.08									
Medial	*0.104*	*Prescribed-Posterior*	0.185	[-0.006, 0.040]	0.46	0.326	[-0.002, 0.006]	0.05	0.745	-	-	-	-	-
		Prescribed-Anterior	0.101	[-0.003, 0.042]	0.55									
		Posterior-Anterior	0.984	[-0.020, 0.025]	0.10									
Vertical	*0.067*	*Prescribed-Posterior*	0.121	[-0.019, 0.199]	0.55	**0.009**	[-0.25, -0.050]	-1.06	0.408	-	-	-	-	-
		Prescribed-Anterior	0.055	[-0.002, 0.216	0.62									
		Posterior-Anterior	0.971	[-0.092, 0.126]	0.13									
Knee Joint Moment	*0.694*	*Prescribed-Posterior*	0.979	[-0.072, 0.055]	-0.02	**< 0.001**	[-0.894, -0.493]	-3.03	0.417	-	-	-	-	-
		Prescribed-Anterior	0.947	[-0.051, 0.075]	0.02									
		Posterior-Anterior	0.786	[-0.084, 0.043]	0.05									

	Alignment Comparison					Leg Comparison (Prosthetic – Intact)			Alignment × Leg Interaction				
	p	*Pairwise Post-hoc*	*p*	*95% CI*	*Cohen's d*	*p*	*95% CI*	*Cohen's d*	*p*	*Post-hoc Interactions*	*p*	*95% CI*	*Cohen's d*
AP Center of Pressure Position	*0.180*	*Prescribed-Posterior*	0.997	[-0.014, 0.012]	**-0.02**	**< 0.001**	[0.043, 0.102]	2.41	0.500	- -	-	-	-
		Prescribed-Anterior	0.255	[-0.022, 0.004]	-0.19								
		Posterior-Anterior	0.345	[-0.005, 0.021]	-0.17								
Trunk Range of Motion													
Lateral	*0.070*	*Prescribed-Posterior*	0.071	[-0.045, 1.29]	0.67		-			-			
		Prescribed-Anterior	0.741	[-0.433, 0.903]	0.22		-			-			
		Posterior-Anterior	0.367	[-1.06, 0.280]	-0.40		-			-			
Axial	***0.042***	*Prescribed-Posterior*	**0.017**	[0.329, 3.74]	1.26		-			-			
		Prescribed-Anterior	0.285	[-0.609, 2.81]	0.45		-			-			
		Posterior-Anterior	0.413	[-2.65, 0.770]	-0.44		-			-			
Flexion	*0.795*	*Prescribed-Posterior*	0.921	[-4.54, 2.96]	-0.10		-			-			
		Prescribed-Anterior	0.907	[-4.61, 2.88]	-0.13		-			-			
		Posterior-Anterior	0.999	[-3.82, 3.67]	-0.01		-			-			

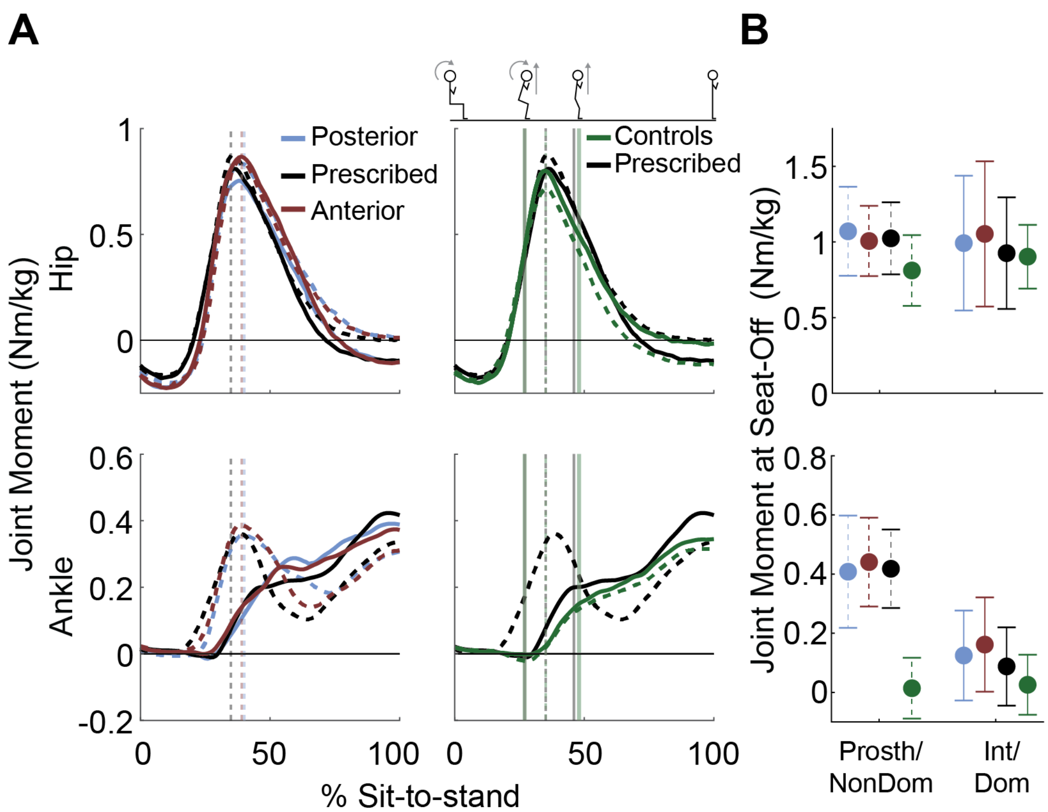

Figure C.1. A) Average internal sagittal plane hip, and ankle moments (Nm/kg) during the STS task for the posterior (blue), anterior (red), and prescribed (black) alignments and Controls (green). Solid lines represent the Intact/Dominant (Int/Dom) side while dashed lines represent the Prosthetic/Non-Dominant (Prosth/NonDom) side. The vertical solid lines indicate the beginning and end of the transition (see Appendix C-C.1) phase while the vertical dashed line indicates the instance of seat-off. **B)** Average joint moment (Nm/kg) at the time of seat-off for each side. Error bars represent ± one standard deviation.

Bibliography

Actis, J.A., Honegger, J.D., Gates, D.H., Petrella, A.J., Nolasco, L.A., Silverman, A.K., 2018a. Validation of lumbar spine loading from a musculoskeletal model including the lower limbs and lumbar spine. J Biomech 68, 107-114.

Actis, J.A., Nolasco, L.A., Gates, D.H., Silverman, A.K., 2018b. Lumbar loads and trunk kinematics in people with a transtibial amputation during sit-to-stand. J Biomech 69, 1-9.

Agrawal, V., Gailey, R., Gaunaurd, I., Gailey, R., 3rd, O'Toole, C., 2011. Weight distribution symmetry during the sit-to-stand movement of unilateral transtibial amputees. Ergonomics 54, 656-664.

Agrawal, V., O'Toole, C., Gaunaurd, I.A., Gailey, R.S., 2016. Analysis of weight distribution strategies in unilateral transtibial amputees during the stand-to-sit activity. Ergonomics 59, 121-129.

Anderson, F.C., Pandy, M.G., 1999. A Dynamic Optimization Solution for Vertical Jumping in Three Dimensions. Comput Methods Biomech Biomed Engin 2, 201-231.

Anderson, F.C., Pandy, M.G., 2001. Dynamic optimization of human walking. J Biomech Eng 123, 381-390.

Andres, R.O., Stimmel, S.K., 1990. Prosthetic alignment effects on gait symmetry: a case study. Clin Biomech (Bristol, Avon) 5, 88-96.

Arnold, A.S., Delp, S.L., 2005. Computer modeling of gait abnormalities in cerebral palsy: application to treatment planning. Theoretical Issues in Ergonomics Science 6, 305-312.

Arya, A.P., Lees, A., Nirula, H.C., Klenerman, L., 1995. A Biomechanical Comparison of the Sach, Seattle and Jaipur Feet Using Ground Reaction Forces. Prosthetics and Orthotics International 19, 37-45.

Azizan, N.A., Basaruddin, K.S., Salleh, A.F., Sulaiman, A.R., Safar, M.J.A., Rusli, W.M.R., 2018. Leg Length Discrepancy: Dynamic Balance Response during Gait. J Healthc Eng 2018, 7815451.

Bateni, H., 2002. Kinematic and Kinetic Variations of Below-Knee Amputee Gait. Journal of Prosthetics and Orthotics 14.

Beltran, E.J., Dingwell, J.B., Wilken, J.M., 2014. Margins of stability in young adults with traumatic transtibial amputation walking in destabilizing environments. J Biomech 47, 1138-1143.

Bennett, B.C., Russell, S.D., Sheth, P., Abel, M.F., 2010. Angular momentum of walking at different speeds. Hum Mov Sci 29, 114-124.

Berg, K.O., Wood-Dauphinee, S.L., Williams, J.I., Maki, B., 1992. Measuring balance in the elderly: validation of an instrument. Can J Public Health 83 Suppl 2, S7-11.

Bergmann, G., Bender, A., Dymke, J., Duda, G., Damm, P., 2016. Standardized Loads Acting in Hip Implants. PLoS One 11, e0155612.

Bergmann, G., Deuretzbacher, G., Heller, M., Graichen, F., Rohlmann, A., Strauss, J., Duda, G.N., 2001. Hip contact forces and gait patterns from routine activities. J Biomech 34, 859-871.

Beyaert, C., Grumillier, C., Martinet, N., Paysant, J., Andre, J.M., 2008. Compensatory mechanism involving the knee joint of the intact limb during gait in unilateral below-knee amputees. Gait Posture 28, 278-284.

Blumentritt, S., 1997. A new biomechanical method for determination of static prosthetic alignment. Prosthet Orthot Int 21, 107-113.

Blumentritt, S., Schmalz, T., Jarasch, R., 2001. [Significance of static prosthesis alignment for standing and walking of patients with lower limb amputation]. Orthopade 30, 161-168.

Blumentritt, S., Schmalz, T., Jarasch, R., Schneider, M., 1999. Effects of sagittal plane prosthetic alignment on standing trans-tibial amputee knee loads. Prosthet Orthot Int 23, 231-238.

Boone, D.A., Kobayashi, T., Chou, T.G., Arabian, A.K., Coleman, K.L., Orendurff, M.S., Zhang, M., 2012. Perception of socket alignment perturbations in amputees with transtibial prostheses. J Rehabil Res Dev 49, 843-853.

Bruno, A.G., Bouxsein, M.L., Anderson, D.E., 2015. Development and Validation of a Musculoskeletal Model of the Fully Articulated Thoracolumbar Spine and Rib Cage. J Biomech Eng 137, 081003.

Buckley, J.G., O'Driscoll, D., Bennett, S.J., 2002. Postural sway and active balance performance in highly active lower-limb amputees. Am J Phys Med Rehabil 81, 13-20.

Bussmann, J.B., Grootscholten, E.A., Stam, H.J., 2004. Daily physical activity and heart rate response in people with a unilateral transtibial amputation for vascular disease. Arch Phys Med Rehabil 85, 240-244.

Bussmann, J.B., Schrauwen, H.J., Stam, H.J., 2008. Daily physical activity and heart rate response in people with a unilateral traumatic transtibial amputation. Arch Phys Med Rehabil 89, 430-434.

Butowicz, C.M., Acasio, J.C., Dearth, C.L., Hendershot, B.D., 2018. Trunk muscle activation patterns during walking among persons with lower limb loss: Influences of walking speed. J Electromyogr Kinesiol 40, 48-55.

Caruthers, E.J., Thompson, J.A., Chaudhari, A.M., Schmitt, L.C., Best, T.M., Saul, K.R., Siston, R.A., 2016. Muscle Forces and Their Contributions to Vertical and Horizontal Acceleration of the Center of Mass During Sit-to-Stand Transfer in Young, Healthy Adults. J Appl Biomech 32, 487-503.

Catelli, D.S., Wesseling, M., Jonkers, I., Lamontagne, M., 2019. A musculoskeletal model customized for squatting task. Comput Methods Biomech Biomed Engin 22, 21-24.

Chihuri, S., Wong, C.K., 2018. Factors associated with the likelihood of fall-related injury among people with lower limb loss. Inj Epidemiol 5, 42.

Chow, D.H., Holmes, A.D., Lee, C.K., Sin, S.W., 2006. The effect of prosthesis alignment on the symmetry of gait in subjects with unilateral transtibial amputation. Prosthet Orthot Int 30, 114-128.

Christophy, M., Faruk Senan, N.A., Lotz, J.C., O'Reilly, O.M., 2012. A musculoskeletal model for the lumbar spine. Biomech Model Mechanobiol 11, 19-34.

Clemens, S.M., Gailey, R.S., Bennett, C.L., Pasquina, P.F., Kirk-Sanchez, N.J., Gaunaurd, I.A., 2018a. The Component Timed-Up-and-Go test: the utility and psychometric properties of using a mobile application to determine prosthetic mobility in people with lower limb amputations. Clin Rehabil 32, 388-397.

Clemens, S.M., Klute, G.K., Kirk-Sanchez, N.J., Raya, M.A., Kim, K.J., Gaunaurd, I.A., Gailey, R.S., 2018b. Temporal-Spatial Values During a 180 degrees Step Turn in People with Unilateral Lower Limb Amputation. Gait Posture 63, 276-281.

Collins, T.D., Ghoussayni, S.N., Ewins, D.J., Kent, J.A., 2009. A six degrees-of-freedom marker set for gait analysis: repeatability and comparison with a modified Helen Hayes set. Gait Posture 30, 173-180.

Correa, T.A., Crossley, K.M., Kim, H.J., Pandy, M.G., 2010. Contributions of individual muscles to hip joint contact force in normal walking. J Biomech 43, 1618-1622.

Courtine, G., Schieppati, M., 2003a. Human walking along a curved path. I. Body trajectory, segment orientation and the effect of vision. Eur J Neurosci 18, 177-190.

Courtine, G., Schieppati, M., 2003b. Human walking along a curved path. II. Gait features and EMG patterns. Eur J Neurosci 18, 191-205.

Couture, M., Caron, C.D., Desrosiers, J., 2010. Leisure activities following a lower limb amputation. Disabil Rehabil 32, 57-64.

Croisier, J.L., de Noordhout, B.M., Maquet, D., Camus, G., Hac, S., Feron, F., De Lamotte, O., Crielaard, J.M., 2001. Isokinetic evaluation of hip strength muscle groups in unilateral lower limb amputees. Isokinet Exerc Sci 9, 163-169.

Crowninshield, R.D., 1978. Use of Optimization Techniques to Predict Muscle Forces. J Biomech Eng-T Asme 100, 88-92.

Crowninshield, R.D., Brand, R.A., 1981. A physiologically based criterion of muscle force prediction in locomotion. J Biomech 14, 793-801.

Curtze, C., Hof, A.L., Postema, K., Otten, B., 2011. Over rough and smooth: amputee gait on an irregular surface. Gait Posture 33, 292-296.

Curtze, C., Hof, A.L., Postema, K., Otten, B., 2012. The relative contributions of the prosthetic and sound limb to balance control in unilateral transtibial amputees. Gait Posture 36, 276-281.

D'Andrea, S., Wilhelm, N., Silverman, A.K., Grabowski, A.M., 2014. Does use of a powered ankle-foot prosthesis restore whole-body angular momentum during walking at different speeds? Clin Orthop Relat Res 472, 3044-3054.

Damayanti Sethy, M., Kujur, E.S., Sau, K., 2009. Effect of balance exercise on balance control in unilateral lower limb amputees. Indian Journal of Occupational Therapy 41, 63-68.

Deans, S.A., McFadyen, A.K., Rowe, P.J., 2008. Physical activity and quality of life: A study of a lower-limb amputee population. Prosthet Orthot Int 32, 186-200.

Delp, S.L., Loan, J.P., Hoy, M.G., Zajac, F.E., Topp, E.L., Rosen, J.M., 1990. An interactive graphics-based model of the lower extremity to study orthopaedic surgical procedures. IEEE Trans Biomed Eng 37, 757-767.

Dempster, W.T., 1955. Space requirements of the seated operator: geometrical, kinematic, and mechanical aspects of the body, with special reference to the limbs. Aero Medical Laboratory Wright-Patterson Air Force Base.

Devan, H., Carman, A., Hendrick, P., Hale, L., Ribeiro, D.C., 2015. Spinal, pelvic, and hip movement asymmetries in people with lower-limb amputation: Systematic review. J Rehabil Res Dev 52, 1-19.

Drake, J.D., Callaghan, J.P., 2006. Elimination of electrocardiogram contamination from electromyogram signals: An evaluation of currently used removal techniques. J Electromyogr Kinesiol 16, 175-187.

Duncan, P.W., Weiner, D.K., Chandler, J., Studenski, S., 1990. Functional reach: a new clinical measure of balance. J Gerontol 45, M192-197.

Ehde, D.M., Smith, D.G., Czerniecki, J.M., Campbell, K.M., Malchow, D.M., Robinson, L.R., 2001. Back pain as a secondary disability in persons with lower limb amputations. Arch Phys Med Rehabil 82, 731-734.

Fang, L., Jia, X., Wang, R., 2007. Modeling and simulation of muscle forces of trans-tibial amputee to study effect of prosthetic alignment. Clin Biomech (Bristol, Avon) 22, 1125-1131.

Fang, L.D., Jia, X.H., Wang, R., Suo, S., 2009. Simulation of the ligament forces affected by prosthetic alignment in a trans-tibial amputee case study. Med Eng Phys 31, 793-798.

Ferris, A.E., Christiansen, C.L., Heise, G.D., Hahn, D., Smith, J.D., 2017a. Ertl and Non-Ertl amputees exhibit functional biomechanical differences during the sit-to-stand task. Clin Biomech (Bristol, Avon) 44, 1-6.

Ferris, A.E., Smith, J.D., Heise, G.D., Hinrichs, R.N., Martin, P.E., 2017b. A general model for estimating lower extremity inertial properties of individuals with transtibial amputation. J Biomech 54, 44-48.

Fey, N.P., Klute, G.K., Neptune, R.R., 2012. Optimization of prosthetic foot stiffness to reduce metabolic cost and intact knee loading during below-knee amputee walking: a theoretical study. J Biomech Eng 134, 111005.

Fey, N.P., Klute, G.K., Neptune, R.R., 2013. Altering prosthetic foot stiffness influences foot and muscle function during below-knee amputee walking: a modeling and simulation analysis. J Biomech 46, 637-644.

Fey, N.P., Silverman, A.K., Neptune, R.R., 2010. The influence of increasing steady-state walking speed on muscle activity in below-knee amputees. J Electromyogr Kinesiol 20, 155-161.

Fiedler, G., Slavens, B.A., O'Connor, K.M., Smith, R.O., Hafner, B.J., 2016. Effects of physical exertion on trans-tibial prosthesis users' ability to accommodate alignment perturbations. Prosthet Orthot Int 40, 75-82.

Friberg, O., 1984. Biomechanical significance of the correct length of lower limb prostheses: a clinical and radiological study. Prosthet Orthot Int 8, 124-129.

Fridman, A., Ona, I., Isakov, E., 2003. The influence of prosthetic foot alignment on trans-tibial amputee gait. Prosthet Orthot Int 27, 17-22.

Gailey, R., Allen, K., Castles, J., Kucharik, J., Roeder, M., 2008. Review of secondary physical conditions associated with lower-limb amputation and long-term prosthesis use. J Rehabil Res Dev 45, 15-29.

Gallagher, P., Maclachlan, M., 2004. The Trinity Amputation and Prosthesis Experience Scales and quality of life in people with lower-limb amputation. Arch Phys Med Rehabil 85, 730-736.

Gardas, S., Shah, H., 2020. Influence of leg length discrepancy on balance and gait in post-stroke patients: a correlational study. Bulletin of Faculty of Physical Therapy 25.

Gates, D.H., Dingwell, J.B., Scott, S.J., Sinitski, E.H., Wilken, J.M., 2012. Gait characteristics of individuals with transtibial amputations walking on a destabilizing rock surface. Gait Posture 36, 33-39.

Gates, D.H., Scott, S.J., Wilken, J.M., Dingwell, J.B., 2013. Frontal plane dynamic margins of stability in individuals with and without transtibial amputation walking on a loose rock surface. Gait Posture 38, 570-575.

Geil, M.D., 2002. Variability among Practitioners in Dynamic Observational Alignment of a Transfemoral Prosthesis. Journal of Prosthetics and Orthotics 14, 159-164.

Geil, M.D., Lay, A., 2004. Plantar foot pressure responses to changes during dynamic trans-tibial prosthetic alignment in a clinical setting. Prosthet Orthot Int 28, 105-114.

Gitter, A., Czerniecki, J.M., DeGroot, D.M., 1991. Biomechanical analysis of the influence of prosthetic feet on below-knee amputee walking. Am J Phys Med Rehabil 70, 142-148.

Glaister, B.C., Bernatz, G.C., Klute, G.K., Orendurff, M.S., 2007. Video task analysis of turning during activities of daily living. Gait Posture 25, 289-294.

Glaister, B.C., Orendurff, M.S., Schoen, J.A., Bernatz, G.C., Klute, G.K., 2008. Ground reaction forces and impulses during a transient turning maneuver. J Biomech 41, 3090-3093.

Golyski, P.R., Hendershot, B.D., 2018. Trunk and pelvic dynamics during transient turns among individuals with unilateral traumatic lower limb amputation. Hum Mov Sci 58, 41-54.

Grabke, E.P., Masani, K., Andrysek, J., 2019. Lower Limb Assistive Device Design Optimization Using Musculoskeletal Modeling: A Review. Journal of Medical Devices 13.

Grumillier, C., Martinet, N., Paysant, J., Andre, J.M., Beyaert, C., 2008. Compensatory mechanism involving the hip joint of the intact limb during gait in unilateral trans-tibial amputees. J Biomech 41, 2926-2931.

Hak, L., van Dieen, J.H., van der Wurff, P., Prins, M.R., Mert, A., Beek, P.J., Houdijk, H., 2013. Walking in an unstable environment: strategies used by transtibial amputees to prevent falling during gait. Arch Phys Med Rehabil 94, 2186-2193.

Handford, M.L., Srinivasan, M., 2016. Robotic lower limb prosthesis design through simultaneous computer optimizations of human and prosthesis costs. Sci Rep 6, 19983.

Hannah, R.E., Morrison, J.B., Chapman, A.E., 1984. Prostheses Alignment - Effect on Gait of Persons with Below-Knee Amputations. Archives of Physical Medicine and Rehabilitation 65, 159-162.

Hansen, A., 2008. Effects of alignment on the roll-over shapes of prosthetic feet. Prosthet Orthot Int 32, 390-402.

Hanvan, E., 1964. A mathematical model of the human body. Aero Medical Laboratory Wright-Patterson Air Force Base.

Harness, N., Pinzur, M.S., 2001. Health related quality of life in patients with dysvascular transtibial amputation. Clin Orthop Relat Res, 204-207.

Hashimoto, H., Kobayashi, T., Gao, F., Kataoka, M., Okuda, K., 2018a. The effect of coronal prosthetic alignment changes on socket reaction moments, spatiotemporal parameters, and perception of alignment during gait in individuals with transtibial amputation. J Rehabil Assist Technol Eng 5, 2055668318795402.

Hashimoto, H., Kobayashi, T., Gao, F., Kataoka, M., Orendurff, M.S., Okuda, K., 2018b. The effect of transverse prosthetic alignment changes on socket reaction moments during gait in individuals with transtibial amputation. Gait Posture 65, 8-14.

Hendershot, B.D., Wolf, E.J., 2014. Three-dimensional joint reaction forces and moments at the low back during over-ground walking in persons with unilateral lower-extremity amputation. Clin Biomech (Bristol, Avon) 29, 235-242.

Herr, H., Popovic, M., 2008. Angular momentum in human walking. J Exp Biol 211, 467-481.

Hewson, A., Dent, S., Sawers, A., 2020. Strength deficits in lower limb prosthesis users: A scoping review. Prosthet Orthot Int 44, 323-340.

Highsmith, M.J., Goff, L.M., Lewandowski, A.L., Farrokhi, S., Hendershot, B.D., Hill, O.T., Rabago, C.A., Russell-Esposito, E., Orriola, J.J., Mayer, J.M., 2019. Low back pain in persons with lower extremity amputation: a systematic review of the literature. Spine J 19, 552-563.

Hof, A.L., 2008. The 'extrapolated center of mass' concept suggests a simple control of balance in walking. Hum Mov Sci 27, 112-125.

Hof, A.L., Gazendam, M.G., Sinke, W.E., 2005. The condition for dynamic stability. J Biomech 38, 1-8.

Hu, X., Blemker, S.S., 2015. Musculoskeletal simulation can help explain selective muscle degeneration in Duchenne muscular dystrophy. Muscle Nerve 52, 174-182.

Hunter, S.W., Batchelor, F., Hill, K.D., Hill, A.M., Mackintosh, S., Payne, M., 2017. Risk Factors for Falls in People With a Lower Limb Amputation: A Systematic Review. PM R 9, 170-180 e171.

Hurwitz, D.E., Sumner, D.R., Block, J.A., 2001. Bone density, dynamic joint loading and joint degeneration. A review. Cells Tissues Organs 169, 201-209.

Inai, T., Takabayashi, T., Edama, M., Kubo, M., 2018. Effect of hip joint angle at seat-off on hip joint contact force during sit-to-stand movement: a computer simulation study. Biomed Eng Online 17, 177.

Isakov, E., Keren, O., Benjuya, N., 2000. Trans-tibial amputee gait: time-distance parameters and EMG activity. Prosthet Orthot Int 24, 216-220.

Isakov, E., Mizrahi, J., Ring, H., Susak, Z., Hakim, N., 1992. Standing sway and weight-bearing distribution in people with below-knee amputations. Arch Phys Med Rehabil 73, 174-178.

Isakov, E., Mizrahi, J., Susak, Z., Ona, I., Hakim, N., 1994. Influence of prosthesis alignment on the standing balance of below-knee amputees. Clin Biomech (Bristol, Avon) 9, 258-262.

Jansen, K., De Groote, F., Aerts, W., De Schutter, J., Duysens, J., Jonkers, I., 2014. Altering length and velocity feedback during a neuro-musculoskeletal simulation of normal gait contributes to hemiparetic gait characteristics. J Neuroeng Rehabil 11, 78.

Janssen, W.G., Bussmann, H.B., Stam, H.J., 2002. Determinants of the sit-to-stand movement: a review. Phys Ther 82, 866-879.

Jayakaran, P., Johnson, G.M., Sullivan, S.J., Nitz, J.C., 2012. Instrumented measurement of balance and postural control in individuals with lower limb amputation: a critical review. Int J Rehabil Res 35, 187-196.

Jeon, W., Jensen, J.L., Griffin, L., 2019. Muscle activity and balance control during sit-to-stand across symmetric and asymmetric initial foot positions in healthy adults. Gait Posture 71, 138-144.

Jia, W., Lee, 2009. Effects of Shoe Heel Height on Loading and Muscle Activity for Trans-Tibial Amputees During Standing.

Jones, G.D., James, D.C., Thacker, M., Jones, E.J., Green, D.A., 2016. Sit-to-walk and sit-to-stand-and-walk task dynamics are maintained during rising at an elevated seat-height independent of lead-limb in healthy individuals. Gait Posture 48, 226-229.

Jonkergouw, N., Prins, M.R., Buis, A.W., van der Wurff, P., 2016. The Effect of Alignment Changes on Unilateral Transtibial Amputee's Gait: A Systematic Review. PLoS One 11, e0167466.

Jonkergouw, N., Prins, M.R., van der Wurff, P., Gijsbers, J., Houdijk, H., Buis, A.W.P., 2019. Dynamic alignment using external socket reaction moments in trans-tibial amputees. Gait Posture 68, 122-129.

Karimi, M.T., Salami, F., Esrafilian, A., Heitzmann, D.W.W., Alimusaj, M., Putz, C., Wolf, S.I., 2017. Sound side joint contact forces in below knee amputee gait with an ESAR prosthetic foot. Gait Posture 58, 246-251.

Kavounoudias, A., Tremblay, C., Gravel, D., Iancu, A., Forget, R., 2005. Bilateral changes in somatosensory sensibility after unilateral below-knee amputation. Arch Phys Med Rehabil 86, 633-640.

Kerr, A., Durward, B., Kerr, K.M., 2004. Defining phases for the sit-to-walk movement. Clin Biomech (Bristol, Avon) 19, 385-390.

Kim, J., Major, M.J., Hafner, B., Sawers, A., 2019. Frequency and Circumstances of Falls Reported by Ambulatory Unilateral Lower Limb Prosthesis Users: A Secondary Analysis. PM R 11, 344-353.

Klute, G.K., Berge, J.S., Orendurff, M.S., Williams, R.M., Czerniecki, J.M., 2006. Prosthetic intervention effects on activity of lower-extremity amputees. Arch Phys Med Rehabil 87, 717-722.

Kobayashi, T., Arabian, A.K., Orendurff, M.S., Rosenbaum-Chou, T.G., Boone, D.A., 2014a. Effect of alignment changes on socket reaction moments while walking in transtibial prostheses with energy storage and return feet. Clin Biomech (Bristol, Avon) 29, 47-56.

Kobayashi, T., Orendurff, M.S., Arabian, A.K., Rosenbaum-Chou, T.G., Boone, D.A., 2014b. Effect of prosthetic alignment changes on socket reaction moment impulse during walking in transtibial amputees. J Biomech 47, 1315-1323.

Kobayashi, T., Orendurff, M.S., Boone, D.A., 2013a. Effect of alignment changes on socket reaction moments during gait in transfemoral and knee-disarticulation prostheses: case series. J Biomech 46, 2539-2545.

Kobayashi, T., Orendurff, M.S., Zhang, M., Boone, D.A., 2012. Effect of transtibial prosthesis alignment changes on out-of-plane socket reaction moments during walking in amputees. J Biomech 45, 2603-2609.

Kobayashi, T., Orendurff, M.S., Zhang, M., Boone, D.A., 2013b. Effect of alignment changes on sagittal and coronal socket reaction moment interactions in transtibial prostheses. J Biomech 46, 1343-1350.

Kobayashi, T., Orendurff, M.S., Zhang, M., Boone, D.A., 2016. Socket reaction moments in transtibial prostheses during walking at clinically perceived optimal alignment. Prosthet Orthot Int 40, 503-508.

Koelewijn, A.D., van den Bogert, A.J., 2016. Joint contact forces can be reduced by improving joint moment symmetry in below-knee amputee gait simulations. Gait Posture 49, 219-225.

Kolarova, B., Janura, M., Svoboda, Z., Elfmark, M., 2013. Limits of stability in persons with transtibial amputation with respect to prosthetic alignment alterations. Arch Phys Med Rehabil 94, 2234-2240.

Krajbich, J.I.P., Michael S. Potter, Benjamin K. Stevens, 2016. Atlas of Amputations and Limb Deficiencies.

Ku, P.X., Abu Osman, N.A., Wan Abas, W.A., 2014. Balance control in lower extremity amputees during quiet standing: a systematic review. Gait Posture 39, 672-682.

Kulkarni, J., Gaine, W.J., Buckley, J.G., Rankine, J.J., Adams, J., 2005. Chronic low back pain in traumatic lower limb amputees. Clin Rehabil 19, 81-86.

Kulkarni, J., Wright, S., Toole, C., Morris, J., Hirons, R., 1996. Falls in Patients with Lower Limb Amputations: Prevalence and Contributing Factors. Physiotherapy 82, 130-136.

Lai, A.K.M., Arnold, A.S., Wakeling, J.M., 2017. Why are Antagonist Muscles Co-activated in My Simulation? A Musculoskeletal Model for Analysing Human Locomotor Tasks. Ann Biomed Eng 45, 2762-2774.

LaPre, A.K., Price, M.A., Wedge, R.D., Umberger, B.R., Sup, F.C.t., 2018. Approach for gait analysis in persons with limb loss including residuum and prosthesis socket dynamics. Int J Numer Method Biomed Eng 34, e2936.

LaPre, A.K., Umberger, B.R., Sup, F., 2014. Simulation of a powered ankle prosthesis with dynamic joint alignment. Annu Int Conf IEEE Eng Med Biol Soc 2014, 1618-1621.

Lewallen, L.K., Srivastava, S., Kautz, S.A., Neptune, R.R., 2021. Assessment of turning performance and muscle coordination in individuals post-stroke. J Biomech 114, 110113.

Liu, M.Q., Anderson, F.C., Schwartz, M.H., Delp, S.L., 2008. Muscle contributions to support and progression over a range of walking speeds. J Biomech 41, 3243-3252.

Lu, T.W., O'Connor, J.J., 1999. Bone position estimation from skin marker co-ordinates using global optimisation with joint constraints. Journal of Biomechanics 32, 129-134.

Luengas, L.A., Sanchez, G., Novoa, K., 2017. Prosthetic Alignment and Biomechanical Parameters in Transtibial Amputees due Landmines, VII Latin American Congress on Biomedical Engineering CLAIB 2016, Bucaramanga, Santander, Colombia, October 26th -28th, 2016, pp. 765-768.

Major, M.J., Fatone, S., Roth, E.J., 2013. Validity and reliability of the Berg Balance Scale for community-dwelling persons with lower-limb amputation. Arch Phys Med Rehabil 94, 2194-2202.

Mandel, A., MscOt, Paul, K., MscOt, Paner, R., Devlin, M., Dilkas, S., Pauley, T., 2016. Balance confidence and activity of community-dwelling patients with transtibial amputation. J Rehabil Res Dev 53, 551-560.

Mansouri, M., Clark, A.E., Seth, A., Reinbolt, J.A., 2016. Rectus femoris transfer surgery affects balance recovery in children with cerebral palsy: A computer simulation study. Gait Posture 43, 24-30.

McAndrew Young, P.M., Wilken, J.M., Dingwell, J.B., 2012. Dynamic margins of stability during human walking in destabilizing environments. J Biomech 45, 1053-1059.

Miller, W.C., Deathe, A.B., Speechley, M., Koval, J., 2001a. The influence of falling, fear of falling, and balance confidence on prosthetic mobility and social activity among individuals with a lower extremity amputation. Arch Phys Med Rehabil 82, 1238-1244.

Miller, W.C., Speechley, M., Deathe, A.B., 2002. Balance confidence among people with lower-limb amputations. Physical Therapy 82, 856-865.

Miller, W.C., Speechley, M., Deathe, B., 2001b. The prevalence and risk factors of falling and fear of falling among lower extremity amputees. Arch Phys Med Rehabil 82, 1031-1037.

Mizrahi, J., Susak, Z., Seliktar, R., Najenson, T., 1986. Alignment procedure for the optimal fitting of lower limb prostheses. J Biomed Eng 8, 229-234.

Molina-Rueda, F., Alguacil-Diego, I.M., Cuesta-Gomez, A., Iglesias-Gimenez, J., Martin-Vivaldi, A., Miangolarra-Page, J.C., 2014. Thorax, pelvis and hip pattern in the frontal plane during walking in unilateral transtibial amputees: biomechanical analysis. Braz J Phys Ther 18, 252-258.

Molina Rueda, F., Alguacil Diego, I.M., Molero Sanchez, A., Carratala Tejada, M., Rivas Montero, F.M., Miangolarra Page, J.C., 2013. Knee and hip internal moments and upper-body kinematics in the frontal plane in unilateral transtibial amputees. Gait Posture 37, 436-439.

Morgenroth, D.C., Segal, A.D., Zelik, K.E., Czerniecki, J.M., Klute, G.K., Adamczyk, P.G., Orendurff, M.S., Hahn, M.E., Collins, S.H., Kuo, A.D., 2011. The effect of prosthetic foot push-off on mechanical loading associated with knee osteoarthritis in lower extremity amputees. Gait Posture 34, 502-507.

Nadollek, H., Brauer, S., Isles, R., 2002. Outcomes after trans-tibial amputation: the relationship between quiet stance ability, strength of hip abductor muscles and gait. Physiother Res Int 7, 203-214.

Neptune, R.R., Kautz, S.A., Zajac, F.E., 2001. Contributions of the individual ankle plantar flexors to support, forward progression and swing initiation during walking. J Biomech 34, 1387-1398.

Neptune, R.R., McGowan, C.P., 2011. Muscle contributions to whole-body sagittal plane angular momentum during walking. J Biomech 44, 6-12.

Neptune, R.R., McGowan, C.P., 2016. Muscle contributions to frontal plane angular momentum during walking. J Biomech 49, 2975-2981.

Neptune, R.R., Zajac, F.E., Kautz, S.A., 2004. Muscle force redistributes segmental power for body progression during walking. Gait Posture 19, 194-205.

Nguyen, V.Q., Umberger, B.R., Sup, F.C., 2019. Predictive Simulation of Human Walking Augmented by a Powered Ankle Exoskeleton. IEEE Int Conf Rehabil Robot 2019, 53-58.

Nolan, L., 2009. Lower limb strength in sports-active transtibial amputees. Prosthet Orthot Int 33, 230-241.

Nolan, L., 2012. A training programme to improve hip strength in persons with lower limb amputation. J Rehabil Med 44, 241-248.

Nolan, L., Lees, A., 2000. The functional demands on the intact limb during walking for active trans-femoral and trans-tibial amputees. Prosthet Orthot Int 24, 117-125.

Nolan, L., Wit, A., Dudzinski, K., Lees, A., Lake, M., Wychowanski, M., 2003. Adjustments in gait symmetry with walking speed in trans-femoral and trans-tibial amputees. Gait Posture 17, 142-151.

Nolasco, L.A., Morgenroth, D.C., Silverman, A.K., Gates, D.H., 2020. Effects of anterior-posterior shifts in prosthetic alignment on the sit-to-stand movement in people with a unilateral transtibial amputation. J Biomech 109, 109926.

Nolasco, L.A., Silverman, A.K., Gates, D.H., 2019. Whole-body and segment angular momentum during 90-degree turns. Gait Posture 70, 12-19.

Norvell, D.C., Czerniecki, J.M., Reiber, G.E., Maynard, C., Pecoraro, J.A., Weiss, N.S., 2005. The prevalence of knee pain and symptomatic knee osteoarthritis among veteran traumatic amputees and nonamputees. Arch Phys Med Rehabil 86, 487-493.

Nott, C.R., Neptune, R.R., Kautz, S.A., 2014. Relationships between frontal-plane angular momentum and clinical balance measures during post-stroke hemiparetic walking. Gait Posture 39, 129-134.

Olensek, A., Zadravec, M., Burger, H., Matjacic, Z., 2021. Dynamic balancing responses in unilateral transtibial amputees following outward-directed perturbations during slow treadmill walking differ considerably for amputated and non-amputated side. J Neuroeng Rehabil 18, 123.

Orendurff, M.S., Segal, A.D., Berge, J.S., Flick, K.C., Spanier, D., Klute, G.K., 2006. The kinematics and kinetics of turning: limb asymmetries associated with walking a circular path. Gait Posture 23, 106-111.

Ozyurek, S., Demirbuken, I., Angin, S., 2014. Altered movement strategies in sit-to-stand task in persons with transtibial amputation. Prosthet Orthot Int 38, 303-309.

Pai, Y.C., Rogers, M.W., 1991. Segmental contributions to total body momentum in sit-to-stand. Med Sci Sports Exerc 23, 225-230.

Paráková, M., Janura, 2009. The influence of prostheses and prosthetic foot alignment on postural behavior in transtibial amputees.

Pedrinelli, A., Saito, M., Coelho, R.F., Fontes, R.B., Guarniero, R., 2002. Comparative study of the strength of the flexor and extensor muscles of the knee through isokinetic evaluation in normal subjects and patients subjected to trans-tibial amputation. Prosthet Orthot Int 26, 195-205.

Pew, C., Segal, A.D., Neptune, R.R., Klute, G.K., 2019. Ideal operating conditions for a variable stiffness transverse plane adapter for individuals with lower-limb amputation. J Biomech 96, 109330.

Pickle, N.T., Grabowski, A.M., Jeffers, J.R., Silverman, A.K., 2017a. The Functional Roles of Muscles, Passive Prostheses, and Powered Prostheses During Sloped Walking in People With a Transtibial Amputation. J Biomech Eng 139.

Pickle, N.T., Silverman, A.K., Wilken, J.M., Fey, N.P., 2017b. Segmental Contributions to Sagittal-Plane Whole-body Angular Momentum When Using Powered Compared to Passive Ankle-foot Prostheses on Ramps. Int C Rehab Robot, 1609-1614.

Pickle, N.T., Silverman, A.K., Wilken, J.M., Fey, N.P., 2019. Statistical analysis of timeseries data reveals changes in 3D segmental coordination of balance in response to prosthetic ankle power on ramps. Sci Rep 9, 1272.

Pickle, N.T., Wilken, J.M., Aldridge, J.M., Neptune, R.R., Silverman, A.K., 2014. Whole-body angular momentum during stair walking using passive and powered lower-limb prostheses. J Biomech 47, 3380-3389.

Pickle, N.T., Wilken, J.M., Aldridge Whitehead, J.M., Silverman, A.K., 2016. Whole-body angular momentum during sloped walking using passive and powered lower-limb prostheses. J Biomech 49, 3397-3406.

Pinzur, M.S., Cox, W., Kaiser, J., Morris, T., Patwardhan, A., Vrbos, L., 1995. The effect of prosthetic alignment on relative limb loading in persons with trans-tibial amputation: a preliminary report. J Rehabil Res Dev 32, 373-377.

Radhakrishnan, S., Kohler, F., Gutenbrunner, C., Jayaraman, A., Li, J., Pieber, K., Schiappacasse, C., 2017. The use of the International Classification of Functioning, Disability and Health to classify the factors influencing mobility reported by persons with an amputation: An international study. Prosthetics and Orthotics International 41, 412-419.

Remes, L., Isoaho, R., Vahlberg, T., Viitanen, M., Rautava, P., 2009. Predictors for institutionalization and prosthetic ambulation after major lower extremity amputation during an eight-year follow-up. Aging Clin Exp Res 21, 129-135.

Rodosky, M.W., Andriacchi, T.P., Andersson, G.B., 1989. The influence of chair height on lower limb mechanics during rising. J Orthop Res 7, 266-271.

Roebroeck, M.E., Doorenbosch, C.A., Harlaar, J., Jacobs, R., Lankhorst, G.J., 1994. Biomechanics and muscular activity during sit-to-stand transfer. Clin Biomech (Bristol, Avon) 9, 235-244.

Rossi, D., Skinner, 1995. Gait initiation of persons with below-knee amputation the characterization and comparison of force profil.

Royer, T.D., Wasilewski, C.A., 2006. Hip and knee frontal plane moments in persons with unilateral, trans-tibial amputation. Gait Posture 23, 303-306.

Rusaw, D.F., 2019. Adaptations from the prosthetic and intact limb during standing on a sway-referenced support surface for transtibial prosthesis users. Disabil Rehabil Assist Technol 14, 682-691.

Russell Esposito, E., Miller, R.H., 2018. Maintenance of muscle strength retains a normal metabolic cost in simulated walking after transtibial limb loss. PLoS One 13, e0191310.

Sadeghi, H., Allard, P., Duhaime, P.M., 2001. Muscle power compensatory mechanisms in below-knee amputee gait. Am J Phys Med Rehabil 80, 25-32.

Sanders, J.E., Bell, D.M., Okumura, R.M., Dralle, A.J., 1998. Effects of alignment changes on stance phase pressures and shear stresses on transtibial amputees: measurements from 13 transducer sites. IEEE Trans Rehabil Eng 6, 21-31.

Sanders, J.E., Daly, C.H., 1999. Interface pressures and shear stresses: sagittal plane angular alignment effects in three trans-tibial amputee case studies. Prosthet Orthot Int 23, 21-29.

Sanderson, D.J., Martin, P.E., 1997. Lower extremity kinematic and kinetic adaptations in unilateral below-knee amputees during walking. Gait & Posture 6, 126-136.

Schmalz, T., Blumentritt, S., Jarasch, R., 2002. Energy expenditure and biomechanical characteristics of lower limb amputee gait: the influence of prosthetic alignment and different prosthetic components. Gait Posture 16, 255-263.

Schmalz, T., Blumentritt, S., Reimers, C.D., 2001. Selective thigh muscle atrophy in trans-tibial amputees: an ultrasonographic study. Arch Orthop Trauma Surg 121, 307-312.

Schoppen, T., Boonstra, A., Groothoff, J.W., de Vries, J., Goeken, L.N., Eisma, W.H., 2003. Physical, mental, and social predictors of functional outcome in unilateral lower-limb amputees. Arch Phys Med Rehabil 84, 803-811.

Schoppen, T., Boonstra, A., Groothoff, J.W., de Vries, J., Goeken, L.N.H., Eisma, W.H., 1999. The timed "up and go" test: Reliability and validity in persons with unilateral lower limb amputation. Archives of Physical Medicine and Rehabilitation 80, 825-828.

Sedgman, R., Goldie, P., Iansek, R., Year Development of a measure of turning during walking. In Advancing rehabilitation: Proceedings of the inaugural conference of the faculty of health sciences. La Trobe University.

Seelen, H.A.M., Anemaat, S., Janssen, H.M.H., Deckers, J.H.M., 2003. Effects of prosthesis alignment on pressure distribution at the stump/socket interface in transtibial amputees during unsupported stance and gait. Clinical Rehabilitation 17, 787-796.

Segal, A.D., Orendurff, M.S., Czerniecki, J.M., Schoen, J., Klute, G.K., 2011. Comparison of transtibial amputee and non-amputee biomechanics during a common turning task. Gait Posture 33, 41-47.

Segal, A.D., Orendurff, M.S., Czerniecki, J.M., Shofer, J.B., Klute, G.K., 2010. Local dynamic stability of amputees wearing a torsion adapter compared to a rigid adapter during straight-line and turning gait. J Biomech 43, 2798-2803.

Segal, A.D., Shofer, J.B., Klute, G.K., 2015. Lower-limb amputee ankle and hip kinetic response to an imposed error in mediolateral foot placement. J Biomech 48, 3982-3988.

Sepp, L., Baum, B.S., Nelson-Wong, E., Silverman, A., 2020. Hip Joint Contact Forces During Running with a Transtibial Amputation. J Biomech Eng.

Seyedali, M., Czerniecki, J.M., Morgenroth, D.C., Hahn, M.E., 2012. Co-contraction patterns of trans-tibial amputee ankle and knee musculature during gait. J Neuroeng Rehabil 9, 29.

Sharifmoradi, K., Karimi, M.T., Ardekani, M.K., Tahmasebi, A., 2017. The Interaction of Knee, Hip and L5-S1 Joint Contact Forces and Spatiotemporal Variables Between Sound and Prosthetic Leg in Patients with Unilateral Below-Knee Amputation During Walking. Journal of Rehabilitation Sciences and Research 4, 53-59.

Shell, C.E., Segal, A.D., Klute, G.K., Neptune, R.R., 2017. The effects of prosthetic foot stiffness on transtibial amputee walking mechanics and balance control during turning. Clin Biomech (Bristol, Avon) 49, 56-63.

Silverman, A.K., Fey, N.P., Portillo, A., Walden, J.G., Bosker, G., Neptune, R.R., 2008. Compensatory mechanisms in below-knee amputee gait in response to increasing steady-state walking speeds. Gait Posture 28, 602-609.

Silverman, A.K., Neptune, R.R., 2011. Differences in whole-body angular momentum between below-knee amputees and non-amputees across walking speeds. J Biomech 44, 379-385.

Silverman, A.K., Neptune, R.R., 2012. Muscle and prosthesis contributions to amputee walking mechanics: a modeling study. J Biomech 45, 2271-2278.

Silverman, A.K., Neptune, R.R., 2014. Three-dimensional knee joint contact forces during walking in unilateral transtibial amputees. J Biomech 47, 2556-2562.

Sin, S.W., Chow, D.H., Cheng, J.C., 2001. Significance of non-level walking on transtibial prosthesis fitting with particular reference to the effects of anterior-posterior alignment. J Rehabil Res Dev 38, 1-6.

Sinha, R., Van Den Heuvel, W.J., 2011. A systematic literature review of quality of life in lower limb amputees. Disabil Rehabil 33, 883-899.

Sinitski, E.H., Lemaire, E.D., Baddour, N., Besemann, M., Dudek, N., Hebert, J.S., 2019. Maintaining stable transtibial amputee gait on level and simulated uneven conditions in a virtual environment. Disabil Rehabil Assist Technol, 1-9.

Sivapuratharasu, B., Bull, A.M.J., McGregor, A.H., 2019. Understanding Low Back Pain in Traumatic Lower Limb Amputees: A Systematic Review. Archives of Rehabilitation Research and Clinical Translation 1.

Slajpah, S., Kamnik, R., Burger, H., Bajd, T., Munih, M., 2013. Asymmetry in sit-to-stand movement in patients following transtibial amputation and healthy individuals. Int J Rehabil Res 36, 275-283.

Soo, C.H., Donelan, J.M., 2010. Mechanics and energetics of step-to-step transitions isolated from human walking. J Exp Biol 213, 4265-4271.

Stansfield, B.W., Nicol, A.C., Paul, J.P., Kelly, I.G., Graichen, F., Bergmann, G., 2003. Direct comparison of calculated hip joint contact forces with those measured using instrumented implants. An evaluation of a three-dimensional mathematical model of the lower limb. Journal of Biomechanics 36, 929-936.

Stineman, M.G., Kurichi, J.E., Kwong, P.L., Maislin, G., Reker, D.M., Vogel, W.B., Prvu-Bettger, J.A., Bidelspach, D.E., Bates, B.E., 2009. Survival analysis in amputees based on physical independence grade achievement. Arch Surg 144, 543-551; discussion 552.

Strike, S.C., Taylor, M.J., 2009. The temporal-spatial and ground reaction impulses of turning gait: is turning symmetrical? Gait Posture 29, 597-602.

Struyf, P.A., van Heugten, C.M., Hitters, M.W., Smeets, R.J., 2009. The prevalence of osteoarthritis of the intact hip and knee among traumatic leg amputees. Arch Phys Med Rehabil 90, 440-446.

Taylor, M.J., Dabnichki, P., Strike, S.C., 2005. A three-dimensional biomechanical comparison between turning strategies during the stance phase of walking. Hum Mov Sci 24, 558-573.

Umberger, B.R., 2010. Stance and swing phase costs in human walking. J R Soc Interface 7, 1329-1340.

Umberger, B.R., Gerritsen, K.G., Martin, P.E., 2006. Muscle fiber type effects on energetically optimal cadences in cycling. J Biomech 39, 1472-1479.

Van Velzen, J.M., Houdijk, H., Polomski, W., Van Bennekom, C.A., 2005. Usability of gait analysis in the alignment of trans-tibial prostheses: a clinical study. Prosthet Orthot Int 29, 255-267.

Vanicek, N., Strike, S., McNaughton, L., Polman, R., 2009a. Gait patterns in transtibial amputee fallers vs. non-fallers: biomechanical differences during level walking. Gait Posture 29, 415-420.

Vanicek, N., Strike, S., McNaughton, L., Polman, R., 2009b. Postural responses to dynamic perturbations in amputee fallers versus nonfallers: a comparative study with able-bodied subjects. Arch Phys Med Rehabil 90, 1018-1025.

Ventura, J.D., Klute, G.K., Neptune, R.R., 2015. Individual muscle contributions to circular turning mechanics. J Biomech 48, 1067-1074.

Ventura, J.D., Segal, A.D., Klute, G.K., Neptune, R.R., 2011. Compensatory mechanisms of transtibial amputees during circular turning. Gait Posture 34, 307-312.

Vistamehr, A., Kautz, S.A., Bowden, M.G., Neptune, R.R., 2016. Correlations between measures of dynamic balance in individuals with post-stroke hemiparesis. J Biomech 49, 396-400.

Wächter, A., Biegler, L.T., 2005. On the implementation of an interior-point filter line-search algorithm for large-scale nonlinear programming. Mathematical Programming 106, 25-57.

Wagner, K.E., Nolasco, L.A., Morgenroth, D.C., Gates, D.H., Silverman, A.K., 2020. The effect of lower-limb prosthetic alignment on muscle activity during sit-to-stand. J Electromyogr Kinesiol 51, 102398.

Wasser, J.G., Vincent, K.R., Herman, D.C., Vincent, H.K., 2020. Potential lower extremity amputation-induced mechanisms of chronic low back pain: role for focused resistance exercise. Disabil Rehabil 42, 3713-3721.

Watson, F., Fino, P.C., Thornton, M., Heracleous, C., Loureiro, R., Leong, J.J.H., 2021. Use of the margin of stability to quantify stability in pathologic gait - a qualitative systematic review. BMC Musculoskelet Disord 22, 597.

Weiss, A., Mirelman, A., Giladi, N., Barnes, L.L., Bennett, D.A., Buchman, A.S., Hausdorff, J.M., 2016. Transition Between the Timed up and Go Turn to Sit Subtasks: Is Timing Everything? J Am Med Dir Assoc 17, 864 e869-864 e815.

White, S.C., Gilchrist, L.A., Wilk, B.E., 2004. Asymmetric limb loading with true or simulated leg-length differences. Clin Orthop Relat Res, 287-292.

WHO, 2001. International Classification of Functioning, Disability and Health (ICF).

Wilken, J.M., Marin, R., 2009. Gait analysis and training of people wiht limb loss, Care of the Combat Amputee.

Wilken, J.M., Rodriguez, K.M., Brawner, M., Darter, B.J., 2012. Reliability and Minimal Detectible Change values for gait kinematics and kinetics in healthy adults. Gait Posture 35, 301-307.

Winter, D.A., 1983. Energy Generation and Absorption at the Ankle and Knee during Fast, Natural, and Slow Cadences. Clin Orthop Relat R, 147-154.

Winter, D.A., 2009. Biomechanics and motor control of human movement. John Wiley & Sons.

Winter, D.A., Sienko, S.E., 1988. Biomechanics of below-knee amputee gait. J Biomech 21, 361-367.

Xiaohong, J., Xiaobing, L., Peng, D., Ming, Z., 2005. The Influence of Dynamic Trans-tibial Prosthetic Alignment on Standing Plantar Foot Pressure. Conf Proc IEEE Eng Med Biol Soc 2005, 6916-6918.

Yamaguchi, G.T., Zajac, F.E., 1989. A planar model of the knee joint to characterize the knee extensor mechanism. J Biomech 22, 1-10.

Yamaguchi, T., Yano, M., Onodera, H., Hokkirigawa, K., 2012. Effect of turning angle on falls caused by induced slips during turning. J Biomech 45, 2624-2629.

Yeung, L.F., Leung, A.K., Zhang, M., Lee, W.C., 2013. Effects of heel lifting on transtibial amputee gait before and after treadmill walking: a case study. Prosthet Orthot Int 37, 317-323.

Yoder, A.J., Petrella, A.J., Silverman, A.K., 2015. Trunk-pelvis motion, joint loads, and muscle forces during walking with a transtibial amputation. Gait Posture 41, 757-762.

Yoder, A.J., Silder, A., Farrokhi, S., Dearth, C.L., Hendershot, B.D., 2019. Lower Extremity Joint Contributions to Trunk Control During Walking in Persons with Transtibial Amputation. Sci Rep 9, 12267.

Yu, C.H., Hung, Y.C., Lin, Y.H., Chen, G.X., Wei, S.H., Huang, C.H., Chen, C.S., 2014. A 3D mathematical model to predict spinal joint and hip joint force for trans-tibial amputees with different SACH foot pylon adjustments. Gait Posture 40, 545-548.

Yu, J.C., Lam, K., Nettel-Aguirre, A., Donald, M., Dukelow, S., 2010. Incidence and risk factors of falling in the postoperative lower limb amputee while on the surgical ward. PM R 2, 926-934.

Zahedi, S., 1986. Alignment of l owe -limb prostheses.

Zajac, F.E., 1989. Muscle and tendon: properties, models, scaling, and application to biomechanics and motor control. Critical reviews in biomedical engineering 17, 359-411.

Zajac, F.E., Neptune, R.R., Kautz, S.A., 2002. Biomechanics and muscle coordination of human walking. Part I: introduction to concepts, power transfer, dynamics and simulations. Gait Posture 16, 215-232.

Zajac, F.E., Neptune, R.R., Kautz, S.A., 2003. Biomechanics and muscle coordination of human walking: part II: lessons from dynamical simulations and clinical implications. Gait Posture 17, 1-17.

Ziegler-Graham, K., MacKenzie, E.J., Ephraim, P.L., Travison, T.G., Brookmeyer, R., 2008. Estimating the prevalence of limb loss in the United States: 2005 to 2050. Arch Phys Med Rehabil 89, 422-429.

Zmitrewicz, R.J., Neptune, R.R., Sasaki, K., 2007. Mechanical energetic contributions from individual muscles and elastic prosthetic feet during symmetric unilateral transtibial amputee walking: a theoretical study. J Biomech 40, 1824-1831.