THE EFFECT OF PROSTHETIC ALIGNMENT ON MUSCLE ACTIVITY FOR PEOPLE WITH A UNILATERAL TRANSTIBIAL AMPUTATION DURING SIT-TO-STAND

by

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ABSTRACT

People with a lower limb amputation have altered motion strategies during daily tasks compared to people without an amputation. These altered motion strategies can result in the development of secondary conditions, such as residual limb or low back pain, but the underlying muscle activity that causes altered motion strategies, and potentially pain development, is often unclear. For people with a lower limb amputation, prosthetic alignment is an important component that may contribute to altered motion strategies and can be easily adjusted in a clinical setting. However, the effect of altered prosthetic alignment on muscle activity is not well understood, especially during sit-to-stand, which is an important activity of daily living. Therefore, the purpose of this work was to evaluate the effect of altered prosthetic alignment on muscle activity for people with a unilateral transtibial amputation (TTA) during sit-to-stand. Characterizing this muscle activity will contribute to the understanding of altered motion strategies associated with TTA and how they may contribute secondary conditions. Further, this work will increase our knowledge of the effects of prosthetic alignment, which is important for improving alignment prescription guidelines.
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CHAPTER 1
INTRODUCTION

Compared to people without an amputation, people with a lower limb amputation have altered movement during daily tasks, and these altered motion strategies can result in the development of secondary conditions such as residual limb pain or low back pain (Ehde et al., 2001; Ephraim et al., 2005; Morgenroth et al., 2010). One important daily task is the STS motion which is performed fifty times per day on average (Bussmann et al., 2008) and is primarily driven by hip and knee extensors and ankle plantarflexors (Caruthers et al., 2016).

While specific differences in muscle coordination between people with and without a lower limb amputation during STS are unknown, the similarities in altered biomechanics in people with a lower limb amputation during both walking and STS indicate that altered muscle coordination during these tasks may also be similar. During both walking and STS, people with a transtibial amputation (TTA) have greater trunk-pelvis lateral bending, greater ranges of axial rotation, and greater average L4-L5 intervertebral compression force (Actis et al., 2018; Yoder et al., 2015). During walking, altered muscle coordination in people with TTA includes an increased reliance on hip extensors to compensate for the loss of functional ankle plantarflexors, which are muscles that are also central to STS execution (Prinsen et al., 2011; Silverman et al., 2008).

For people with a lower limb amputation, prosthetic alignment is another factor that may affect muscle coordination. Altering prosthetic alignment has effects on rates of energy consumption (Schmalz et al., 2002) as well as asymmetries in movement and loading (Andres and Stimmel, 1990) during walking. However, no study has specifically evaluated how shifts in prosthetic alignment may affect motion strategies and underlying muscle coordination during STS.

Therefore, the purpose of this research was to evaluate the effects of altered prosthetic alignment on muscle activity for people with TTA during STS. Characterizing muscle activity during STS will contribute to the understanding of altered motion strategies associated with TTA. This knowledge will provide insight into the development of secondary conditions in this population. Further, establishing the relationship between prosthetic alignment shifts and resulting changes in muscle activity is important for improving alignment prescription guidelines.
CHAPTER 2
REVIEW OF THE LITERATURE

2.1. Lower limb amputation

An estimated 1.6 million people in the U.S. were living with the loss of a limb in 2005 and that number is expected to more than double by 2050. Causes of limb loss include dysvascular disease (54%), trauma (45%), and cancer (< 2%). 65% of all limb loss is to a lower extremity, and of those, over half is considered major (excluding toe amputations) (Ziegler-Graham et al., 2008). Out of all lower limb amputations, 28% are transtibial (below the knee) and 26% are transfemoral (above the knee) (Dillingham et al., 2002).

2.1.1. Biomechanics of People with Lower Limb Amputations

People with a transtibial amputation (TTA) have altered motion and loading for a variety of daily tasks. In particular, analyses of walking have received the greatest attention in the literature. For example, during walking, people with TTA have greater trunk-pelvis lateral bending toward the residual limb, greater ranges of trunk-pelvis axial rotation, and greater average L4-L5 compression force (Yoder et al., 2015) compared to people without TTA. Those with TTA also have greater L4-L5 compressive loading and trunk-pelvis axial rotation relative to those without TTA during the sit-to-stand motion (Actis et al., 2018). During walking, the prosthetic limb is loaded less than the intact limb and the onset of loading is delayed, minimizing the time spent on the prosthetic limb (Breakey, 1976; Isakov et al., 2000; Nolan et al., 2003; Pinzur et al., 1995; Rossi et al., 1995). However, increasing walking speed does not increase GRF asymmetries between limbs in amputees, suggesting that greater intact limb dependence may not be strictly necessary (Silverman et al., 2008). People with TTA also have altered contralateral knee joint function during walking. Compared to control subjects, those with TTA have greater total work in the contralateral knee and less work in the ipsilateral knee (Beyaert et al., 2008; Silverman et al., 2008). Knee work is influenced by changes in activation of knee flexors and extensors, indicating altered muscle recruitment strategies during gait. Other indications of altered muscle activity include the rate of oxygen consumption when walking, which increases by 25% for those with TTA and by 55-65% for those with TFA compared to those without an amputation (Schmalz et al., 2002). Ultimately, the altered kinematics and
loading during gait in those with a lower limb amputation are driven by compensatory muscle activity.

Muscle activity in people with an amputation during daily tasks is characterized by asymmetry and includes strategies to compensate for the loss of functional ankle plantarflexors. During running, people with a lower limb amputation have muscle work asymmetries with greater hip muscle work in the prosthetic limb relative to the intact limb during the stance phase, and greater hip and knee muscle work in the intact limb relative to the prosthetic limb during swing phase (Czerniecki and Gitter, 1992). These work asymmetries are associated with muscle atrophy in the hip (TFA) and thigh (TTA) regions, and greater activity in hip and back muscles during walking gait (Devan et al., 2014). During walking gait, there is heightened hip extensor activity in the residual limb compared to the intact limb and those without an amputation (Isakov et al., 2001; Winter and Sienko, 1988), and increased gluteus maximus activity in the intact limb relative to those without an amputation (Torburn et al., 1990; Winter and Sienko, 1988). In addition, compared to the intact limb, the residual limb biceps femoris, vastus lateralis, and rectus femoris have increased activity, quantified by integrated EMG, during early to mid-stance for a range of walking speeds, and there is no difference in muscle activity during walking between the intact limb and people without an amputation (Fey et al., 2010). Overall, people with a lower limb amputation have greater hip extensor activity compared to people without an amputation, which is suggested to compensate for the loss of function of the ankle plantarflexors to provide propulsion during walking (Fey et al., 2010; Prinsen et al., 2011).

The altered muscle coordination and loading observed in people with a lower limb amputation are linked to back pain development in the general population. People with TTA have a greater range of muscle forces in the erector spinae and obliques in double-limb support phases during gait (Yoder et al., 2015). However, during sit-to-stand those with TTA do not have significant differences in these muscle forces compared to those without TTA, although increased L4-L5 loading is present (Actis et al., 2018). In addition, hip extensor fatigability, strength and imbalance are linked to LBP in those without a lower limb amputation (Kankaanpää et al., 1998; Nadler et al., 2001, 2000), which may influence low back pain (LBP) development because of the proximity of these muscles to low back skeletal structure and musculature. Overall, altered muscle activity and loading in those with a lower limb amputation suggests possible mechanisms for pain development.
Ultimately, people with a lower limb amputation have altered movement, loading, and muscle coordination compared to those without an amputation. Characterizing these altered motion strategies is important to better understand potential mobility limitations and risks for secondary conditions, including residual limb pain and low back pain, which can inform therapy and device interventions to improve long-term health.

2.1.2. Lower Limb Prostheses

Prosthetic comfort is important for patient physical and mental health and can be influenced by prostheses components and alignment. Ultimately, satisfactory prosthetic fit and function correspond to greater independence and mobility, improved perceived quality of life, and employment success (Webster et al., 2012). Important components of a lower limb prosthesis include the socket, pylon, and foot (Figure 2.1). Socket design selection depends on the shape and condition of the residual limb, and is important for postural stabilization during standing, walking, and other activities of daily living. Prosthetic feet are either articulated (has a hinged ankle mechanism) or non-articulated (has a rigid ankle connection) and selection of food design is heavily dependent on age, health, and activity level. The prosthetic pylon in particular is an important component in prosthetic alignment, which is defined as the position of the pylon relative to the socket.

Ideally, prostheses would perform functional tasks normally executed by the lower limb, however prosthetic deficiencies often lead to altered biomechanics in those with a lower limb amputation. Prostheses are successful in providing body support in the absence of ankle muscles and in providing trunk propulsion in the absence of the soleus muscle. However, prostheses do not aid in the initiation of leg swing in the absence of the gastrocnemius (Silverman and Neptune, 2012). Another influence on the biomechanics of those with a lower limb amputation is prosthetic foot design. A limitation of many prosthetic feet is that they do not accurately mimic ankle function because of the absence of plantarflexors, which is a significant influence on the altered biomechanics of those with a lower limb amputation. One type of prosthetic foot is energy-storage-and return (ESAR), which are designed to approximate controlled plantarflexion and dorsiflexion of unimpaired ankles. However these feet only produce 50% of plantarflexion power compared to unimpaired ankles (Ferris et al., 2012) as they don’t actively contract like the ankle plantarflexors. There are many limitations in how prostheses imitate a natural limb, which contribute to altered movement seen in those with a lower limb amputation.
2.1.2.1. Prosthetic Alignment

Prosthetic alignment is usually prescribed based on clinician experience and patient feedback and alignment adjustments can be made quickly and easily. Alignment adjustment parameters include anterior/posterior or medial/lateral translations and rotations, socket height, and foot angle. These adjustments are achieved through translations and/or rotations between the socket and pylon. Socket height is altered by increasing or decreasing the length of the pylon. Toe angle is altered by changing the angle between the foot axis and the direction of propulsion during walking (Figure 2.2).

The alignment prescription process is highly iterative and includes bench, static, and dynamic alignment steps. During bench alignment, the prosthesis is assembled without the user to establish a starting point that will provide stability in weight bearing during the first static alignment trials. While there are different systems of achieving bench alignment, the process is generally dependent on vertical reference line geometry and an acceptable alignment can be achieved using different systems. Figure 2.3 compares two acceptable transfemoral bench
alignments using the German and UC Berkeley reference positions (Radcliffe, 1994). The next
step is a static alignment that is conducted with the user in sitting and standing positions. The
static alignment process aims to make any height corrections as necessary, and to adjust socket
orientation so the patient can maintain balance while standing. The final step is a dynamic
alignment, which is conducted during walking and aims to maximize overall function and
comfort. Both static and dynamic alignment requirements are assessed based on patient feedback
and observations of symmetry by the prosthetist. Alignment requirements are specific to each
prosthetic user, but prosthetists and patients consider a wide range of alignments acceptable, with
variations up to 148 mm in translations and 17 degrees in rotations (Zahedi et al., 1986).
However, when alignment is altered and tested for use in non-level walking, the range of
acceptable alignments is reduced, suggesting that there is an “optimal” alignment that maximizes
performance for a range of physical activities (Sin et al., 2001).

Figure 2.2: Alignment parameters for a transtibial prosthesis include: anterior/posterior and
medial/lateral tilts and shifts, toe angle, and height. (Zahedi et al., 1986)
2.1.2.2. Effect of Prosthetic Alignment on Biomechanics

The effects of prosthetic alignment have been primarily investigated during walking and studies show that while those with a lower limb amputation will consider some alignments to be unacceptable due to discomfort, gait adaptation makes it possible to walk even with malaligned prostheses. For example, altering prosthetic foot angles has no effect on overall walking speed, but adaptation for excessive external rotation of the foot includes altered stance time, swing time, and step length of the prosthetic limb, and increased internal rotation of the hip joint (Fridman et al., 2003). Further, anterior/posterior shifts in alignment produce the greatest gait asymmetries during the prosthetic limb stance phase with decreased hip and knee angles at heel strike associated with anterior malalignments, and increased heel strike hip and knee angles associated with posterior malalignments (Andres and Stimmel, 1990). However, when alignment is altered within the range of what is acceptable, as determined experimentally for individual participants including maximum translations of 30 mm and rotations of 6 degrees, there is no significant difference between conditions in vertical ground reaction force, stance duration, and step length.
during walking (Chow et al., 2006). Therefore, these parameters may be a valuable indication of unacceptable alignments, but further study is needed. Ultimately, extreme changes in alignment show clear effects on gait parameters, whereas small alignment changes do not. Further research is needed to investigate the biomechanical effects of alignment changes within the acceptable range.

Static alignment procedures aim to adjust socket orientation to facilitate proper balance while standing, as observed by the prosthetist, and a malaligned prosthesis may contribute to an increased risk of falling. When altering alignment of prostheses, Kolarova et al., found that sagittal plane rotation had a larger effect on movement strategies to maintain balance than changing the length of the prostheses (Kolarova et al., 2013). Instability, often characterized by increased risk of falling, is linked to overall LBP development (Luoto et al., 1998), and prosthetic alignment can contribute to altered motions strategies to maintain balance.

The muscular compensations seen in people with a lower limb amputation affect oxygen consumption and knee joint function and further changes are present under altered alignment conditions. The rate of oxygen consumption is greater for people with a lower limb amputation compared to people without an amputation. Further, altered prosthetic alignment, specifically excessive dorsiflexion or plantarflexion, increases energy consumption rates for both those with TFA and TTA (Schmalz et al., 2002). People with a lower limb amputation also have greater total work in the contralateral knee. For uncomfortable malalignment of the prosthesis, the work in the contralateral knee further increases, while work in the ipsilateral knee stays constant (Beyaert et al., 2008). Therefore, altered prosthetic alignment may contribute to further increasing the altered muscle coordination seen in people with a lower limb amputation. However, no study has specifically investigated the effects of prosthetic alignment on muscle activation.

Prosthetic alignment also affects the interface between the socket and residual limb. Malalignments of the prosthesis cause greater socket pressure in those with TTA (Pearson et al., 1973), and there are significant changes in socket reaction moments, specifically for the coronal plane, due to rotational and translational malalignments (Boone et al., 2013). Because there are other planes that have relatively little change in socket reaction moment in response to alignment perturbations, individuals are likely able to accommodate for the alignment change with altered muscle activity in the residual limb. No study has investigated the influence of prosthetic
alignment on muscular activity, and a greater understanding of this relationship will inform therapy and device interventions to improve long-term mobility with potential to increase stability and limit pain development in people with TTA.

2.1.3. Development of Secondary Pain Conditions

People with physical disabilities such as spinal cord injury, multiple sclerosis, cerebral palsy, and limb amputations often have secondary pain conditions, however the specific sources of pain and how it may impact their disability is often not well understood (Ehde et al., 2003). Secondary pain conditions as a result of amputation include phantom pain, residual limb pain, and back pain, and amputees with chronic pain report significantly more disability than those without pain (Marshall et al., 1992). Predicting factors of chronic pain in people with an amputation include, level of amputation, presence of pre-amputation pain, and fit of the prosthesis (Houghton et al., 1994; Nikolajsen et al., 1997). Further, chronic pain is negatively correlated with employment (Schoppen et al., 2002) and nearly 95% of amputees report experiencing at least one type of amputation related pain (Ephraim et al., 2005).

2.1.3.1. Phantom and Residual Limb Pain

Phantom and residual limb pain are common secondary conditions associated with lower limb amputations with nearly 80% experiencing phantom pain (Ephraim et al., 2005) and 68% experiencing residual limb pain (Ephraim et al., 2005). Phantom pain is defined as pain that is perceived as located in the missing portion of the amputated limb. Etiology of phantom pain is unclear, however sensations of cramping phantom pain are correlated with increases in muscle tension in the residual limb (Sherman et al., 1992). Residual limb pain is pain that occurs in the residual portion of the amputated limb and while 10-13% of people with an amputation report experiencing residual limb pain in the first 2 years following amputation, the prevalence of residual limb pain can be as high as 76% for those who have had an amputation for more than 2 years (Ehde et al., 2000; Jensen et al., 1985; Parkes, 1973). Increasing prevalence of residual limb pain over time suggests that altered biomechanics in people with an amputation may eventually lead to pain.

2.1.3.2. Low Back Pain

While LBP is a problem faced by nearly 29% of U.S. adults (Schiller et al., 2012), the prevalence of LBP is substantially larger in those with a lower limb amputation. In this population, reports of LBP can be as high as 65% (Ephraim et al., 2005). For those with a lower
limb amputation with persistent, bothersome back pain, 25% reported severe pain that interfered
with daily activities (Ehde et al., 2001). However when looking at MRI scans of the lumbar spine
in those with a lower limb amputation, there were no differences observed between ‘pain’ and
‘no pain’ groups, suggesting that LBP in people with an amputation has a biomechanical, rather
than a degenerative, etiology. Further, over half of those with a lower limb amputation report
pain onset in the first two years after amputation, indicating that amputation is an important
contributing factor to LBP and may influence biomechanical causes (Kulkarni et al., 2005).

Experimental studies have shown differences in trunk motion strategies between people with
and without pain which may be contributing factors to pain development. When comparing
lumbopelvic kinematics in people with a transfemoral amputation (TFA), with and without LBP,
there are significant differences found in movement patterns, including greater lumbar rotation
toward the prosthetic limb and increased extension of the spine in the sagittal plane for people
with LBP. However, for people with TTA, there are no notable differences between pain groups
(Devan et al., 2009). In addition, there is greater transverse plane rotation in people with TFA
and LBP compared to people without LBP (Morgenroth et al., 2010), which may indicate
rotational instability. Rotational instability of the spine in the transverse plane, characterized by
excessive displacement of spinal motion segments, is implicated as a contributing factor to
development of specific LBP. In addition, cadaveric lumbar spine motion segments have shown
that greater transverse plane rotation is linked to the severity of intervertebral disc degeneration
(Fujiwara et al., 1999). Models have also shown that transverse rotation causes the greatest IVD
tensile strain compared to any other triplanar motion of the spine (Schmidt et al., 2007), which
may contribute to non-specific forms of LBP. Overall, differences between pain groups indicate
that increased lumbar rotation and rotational instability of the spine may be contributing
biomechanical factors to pain development.

2.2. Sit-to-Stand Motion

The sit-to-stand (STS) motion is a complex, whole-body transfer task that requires horizontal
acceleration of the body center of mass driven by trunk flexion, followed by vertical center of
mass acceleration driven by trunk and knee extension. There are many determinants of STS
performance, including factors related to the person, the chair, and the motion strategy used.
Therefore, the kinematics, kinetics, and muscle activity of the STS motion are variable across
factors such as age, sex, muscle strength, chair height, foot position, and type of arm movements.
For instance, different foot position strategies will affect the relationship between the center of mass and center of pressure of a person during STS and this change is accompanied by altered muscle activity (Frykberg and Häger, 2015).

While determinants of STS can affect task motion strategies and the underlying muscle activity, some aspects of muscle coordination are standard to the STS motion. STS is characterized by early activation of the tibialis anterior and abdominal muscles that prepare postural movement as well as by the sequential activation of the lumbar paraspinals, quadriceps, and hamstrings (Rodrigues-De-Paula Goulart and Valls-Solé, 1999). This sequence of muscle activation is tied to the time of lift off from the chair and is not affected by type of STS strategy, indicating that the lumbar paraspinals, quadriceps, and hamstrings are vital for STS execution.

The onset of hip extension during STS is associated with activation of the biceps femoris, tibialis anterior, and gastrocnemius, while the onset of knee extension is heavily dependent on foot position strategy (Khemlani et al., 1999). At the time of lift of from the chair, the largest increase in activity is seen in the knee extensors with a moderate increase in hip extensor activity (Roebroeck et al., 1994). Finally, during the sit-to-stand motion (STS) in healthy individuals, there is no statistical difference between dominant and non-dominant sides in the root mean square of the erector spinae and rectus femoris EMG signals (Burnett et al., 2011).

The STS motion is an important activity for daily living that is performed 50 times a day on average (Bussmann et al., 2008, 2004) and has greater spinal loading than either walking or running (Wilke et al., 1999). Therefore, characterizing the biomechanics of STS is important to understand potential mobility limitations associated with an important daily task, and is likely especially relevant for determining potential biomechanical causes of pain, particularly LBP.

### 2.3. Experimental Methods in Biomechanics

Characterizing movement under different conditions, such as changes in low back biomechanics under altered prosthetic alignment and varying pain levels, requires methods of biomechanical measurement. Non-invasive experimental methods can measure ground reaction forces, surface level muscle activity, and whole-body kinematics. Ground reaction forces are measured through in ground force plates, or an instrumented treadmill. Surface level muscle activity is measured through electromyographic (EMG) sensors on the skin that measure the excitation of muscles. Kinematics are measured using an optical motion capture system that tracks marker positions, which provide estimates of skeletal positions. However, methods of
measuring three-dimensional kinematics through marker sets are not without error and can be affected by anatomical landmark misplacement and soft tissue artifacts. Appropriate landmark placement is dependent on the instrumentor, but the instrumentation process for marker placement should be repeatable between participants, sessions, and laboratories. When investigating test-retest reliability, Fernandes et al. found that while there was varied reliability for multi-segment trunk joint angles during gait, the level of error was acceptable (< 4 degrees for all parameters), especially for the sagittal plane (Fernandes et al., 2016).

Surface EMG (sEMG) sensors provide a non-invasive way to measure activity of muscles, however there are associated limitations. As the name suggests, sEMG sensors can only measure the activity of surface muscles. Other limitations associated with sEMG include the large variation in the amplitude of signals, and differences in the electric conductivity of tissue (Oliver et al., 1996). In addition, sEMG signals from a given muscle are susceptible to ‘cross-talk’ from neighboring muscles, so it may be difficult to isolate a single muscle activation. Therefore, sensor placement is an important part of the experimental setup. For lumbar paraspinal measurements, one method is to place surface electrodes bilaterally over the greatest convexity of the erector spinae muscles at the L4-L5 level (Humphrey et al., 2005; Oliver et al., 1996). Other placement methods include placing bilateral sensors 2-4 centimeters from the midline of the spine (Ekstrom et al., 2008; Neblett et al., 2014).

Because of their location in the torso, EMG measurements from muscles in the low back are susceptible to electrocardiogram (ECG) contamination. Therefore, processing techniques to remove ECG contamination may be necessary when investigating trunk muscle activity. One method of ECG signal removal that balances ease of implementation, time investment, and overall performance is a fourth order Butterworth filter with a cutoff frequency of 30 Hz (Drake and Callaghan, 2006), however this method has only been investigated for 10-25% of voluntary contraction. Overall, trunk muscle EMG data may need additional processing compared to EMG data from other muscles.

2.4. Summary

People with a lower limb amputation have altered motion, loading, and muscle coordination during daily tasks compared to people without an amputation. Changes in whole body biomechanics in people with an amputation during walking and STS include greater ranges of axial rotation and lateral bending, greater L4-L5 compression force, and asymmetries in trunk
and lower limb muscle activity. These altered motion strategies are often a response to the loss of functional ankle control and the limitations of prosthetic devices to mimic natural limbs. Prosthetic alignment is one factor that contributes to altered movement strategies and loading in people with a lower limb amputation and while people with a lower limb amputation have altered muscle activity during gait, particularly for hip and back muscles, no study has characterized muscle activity relative to changes in prosthetic alignment.

Most people with an amputation will experience amputation related pain (phantom limb, residual limb, or low back pain) and the altered motion strategies observed in people with an amputation likely contribute to pain development. However, while altered and asymmetric motion strategies and muscle coordination are linked to pain development, the specific relationships are unclear. A better understanding of altered motion strategies used by people with an amputation, and the effect of prosthetic alignment will help to inform therapy and device interventions to improve quality of life. Therefore, the purpose of this research is to quantify muscle activity in people with a unilateral transtibial amputation during the sit-to-stand movement for varying alignment conditions.

Hypotheses for this work include:

- People with TTA will have altered muscle coordination during STS compared to people without TTA, with greater hip extensor and abductor activity as previously seen during walking.
- For alignment conditions that deviate from the prescribed alignment, there will be greater activity in muscles central to STS execution, such as the knee extensors, compared to the prescribed alignment and people without TTA.
CHAPTER 3
EFFECT OF MEDIAL/LATERAL PROSTHETIC ALIGNMENT SHIFTS ON MUSCLE ACTIVITY FOR PEOPLE WITH A UNILATERAL TRANSTIBIAL AMPUTATION DURING SIT-TO-STAND
A paper to be submitted to the Journal of Electromyography and Kinesiology

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3.1. Introduction

People with a lower limb amputation have altered motion strategies during daily tasks, including the sit-to-stand (STS) motion. STS is an important activity for daily living that both people with and without a lower limb amputation perform 50 times a day on average (Bussmann et al., 2008, 2004). STS is a complex, whole-body transfer task that requires horizontal acceleration of the body center of mass driven by trunk flexion, followed by vertical center of mass acceleration driven by trunk and knee extension. Main contributors to vertical acceleration of the body center of mass include the soleus, gluteus maximus, and vastus lateralis. The gluteus maximus, soleus, and biceps femoris drive forward horizontal acceleration while the rectus femoris, and vastus lateralis oppose it (Caruthers et al., 2016). There are many determinants of STS performance, including factors related to the person, the chair, and the motion strategy used. Therefore, the kinematics, kinetics, and muscle activity of the STS motion are variable across factors such as age, sex, muscle strength, chair height, foot position, and type of arm movements (Frykberg and Häger, 2015). Generally, STS is characterized by early activation of the tibialis anterior and abdominal muscles that prepare postural movement as well as by the sequential

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activation of the lumbar paraspinals, quadriceps, and hamstrings (Rodrigues-De-Paula Goulart and Valls-Solé, 1999). This sequence of muscle activation is tied to the time of lift off from the chair and is not affected by type of STS strategy, indicating that the lumbar paraspinals, quadriceps, and hamstrings are vital for STS execution.

There are significant differences in gait between people with and without a transtibial amputation (TTA) due to compensatory mechanisms observed in people with TTA including an increased reliance on the hip extensors to compensate for the functional loss of the ankle plantarflexors (Prinsen et al., 2011; Silverman et al., 2008). Studies using electromyographic (EMG) data have identified asymmetry in people with TTA during walking. The altered walking mechanics in people with TTA are characterized by asymmetric muscle activity in the biceps femoris, vastus lateralis, and rectus femoris (Fey et al., 2010) and greater gluteus maximus activity in the intact limb (Torburn et al., 1990; Winter and Sienko, 1988). In addition, a modeling study has shown that people with TTA have greater ranges of muscle forces in the paraspinals and obliques (Yoder et al., 2015) compared to people without an amputation during walking.

Less is known about the muscle activity in people with TTA during STS, however greater trunk-pelvis lateral bending, greater ranges of axial rotation, and greater average L4-L5 intervertebral compression force are observed in people with TTA compared to people without an amputation during both walking and STS (Actis et al., 2018; Yoder et al., 2015). In addition, similar to during walking, the functional loss of ankle muscles will likely affect muscle activity in people with an amputation during STS, requiring greater activity in other muscles to achieve the task. Further, muscle groups that have altered coordination during walking in people with TTA, including the gluteus maximus, quadriceps, hamstrings, and ankle plantarflexors, are also critical in achieving the STS task (Caruthers et al., 2016). The similar changes in motion and loading in people with TTA during both walking and STS indicate that people with TTA likely also have altered muscle activity during STS compared to people without an amputation.

For people with a lower limb amputation, prosthetic alignment is a key component of prosthetic fit and function and adjustments in alignment can be made through translations or rotations between the socket and pylon. Altering prosthetic alignment has effects on asymmetry in movement and loading (Andres and Stimmel, 1990) and rates of energy consumption (Schmalz et al., 2002), during walking. However, no study has evaluated how shifts in prosthetic
alignment may affect motion strategies and the underlying muscle coordination during daily tasks. A greater understanding of the relationship between prosthetic alignment and muscle coordination during STS will inform therapy and device interventions to improve long-term mobility with potential to increase stability and limit pain development in people with TTA. Therefore, the purpose of this study was to investigate the effects of prosthetic alignment on muscle activity during STS for people with unilateral TTA. EMG signals were used to calculate integrated EMG and compare total muscle activity for people with and without TTA and for people with TTA for prescribed, medial, and lateral prosthetic alignment conditions. As a means to interpret muscle activity results, we also analyzed ground reaction forces (GRFs), trunk-pelvis flexion/extension angle, and medial/lateral center of mass (COM). We hypothesized that people with TTA would have altered muscle coordination during STS compared to people without TTA, with greater hip extensor and abductor activity as previously observed during walking. In addition, because of increased rates of energy consumption in response to alignment changes during walking, we hypothesized that the muscle coordination for people with TTA would change under altered alignment conditions. Specifically, we hypothesized that there would be greater activity in muscles central to STS execution, specifically the knee extensors, for alignments that deviated from the prescribed alignment.

### 3.2. Methods

#### 3.2.1. Participants

Eight participants with unilateral TTA (7M/1F, 48 ± 17 years, 1.77 ± 0.08 m, 93.65 ± 16.39 kg) and eight height-, weight-, sex-, and age-matched non-amputee participants (7M/1F, 43 ± 16 years, 1.79 ± 0.08 m, 84.99 ± 15.93 kg) completed the experimental protocol. We recruited people with unilateral TTA through the University of Michigan Prosthetic and Orthotics Center. Participants had to be able to walk independently for 10 minutes at a time for at least two months prior to data collection. Potential participants were excluded if they had pathology or injury of the intact limb, were taking medications that affected their ability to walk, had neurologic or cardiovascular disease, had significant vision problems, suffered from an impaired mental capacity that negatively impacted verbal communication with the clinicians and research team, or if the length of the residual limb prevented alignment adjustments. The protocol was approved by the Institutional Review Board and participants provided written informed consent. Limb dominance for participants without TTA was determined as the leg used to kick a ball and
dominant and non-dominant side muscles were compared to intact and amputated side muscles in people with TTA.

3.2.2. Experimental Protocol

Each participant completed a set of five STS trials. Participants with TTA performed sets of trials with different prosthetic alignment conditions including their prescribed alignment and 10-mm medial and lateral translations of the prosthetic pylon (Figure 3.1). A certified prosthetist performed alignment shifts. Trials began with participants seated on a backless chair, with their hips and knees in 90 degrees of flexion, and feet placed fully on the ground approximately hips width apart. Each foot and the chair were in contact with a separate force plate.

![Figure 3.1: Prosthesis for people with a transtibial (below-knee) amputation including socket, pylon, and foot components. Translational alignment adjustments are made by shifting the position of the pylon relative to the socket. The medial and lateral translation adjustments made in this study are indicated in red and blue, respectively.](image)

Kinematic data were collected using a 20-camera optical motion capture system (Motion Analysis Corp., Santa Rosa, CA, 120 Hz) simultaneously with GRF from in-ground force plates (AMTI Inc., Watertown, MA, 1200 Hz). Participants were instrumented with a set of 45 dynamic kinematic markers including markers placed bilaterally on the temple, back of head, acromion, iliac crest, anterior superior iliac spine, posterior superior iliac spine, 4-marker thigh cluster, 4-marker shank cluster, 5th metatarsal head, 1st metatarsal head, and heel. Markers were also placed on the sternum, clavicle, right back, C7, and T10-T8 triad (where 2 markers are placed at T8) (Figure 3.2a).
After shaving and cleaning the skin, 16 surface EMG sensors (Delsys, Inc., Boston, MA; contact material, 99.9% Ag; electrode dimensions, 5 by 1 mm; inter-electrode distance, 10 mm) were placed bilaterally on the thoracic paraspinals (TP), lumbar paraspinals (LP), gluteus medius (GM), rectus femoris (RF), vastus lateralis (VL), biceps femoris long head (BFlh), medial gastrocnemius (MG), and tibialis anterior (TA) (Figure 3.2b). Sensors were placed on the muscle belly halfway between origin and insertion and oriented parallel to muscle fiber direction (Konrad, 2006; Perotto, 2011) and data were collected at 1200 Hz. EMG data were not collected for the remaining portions of the amputated side MG and TA in participants with TTA to ensure socket comfort.

![Figure 3.2: (a) Approximate placement of dynamic kinematic markers. Bilateral placement on the temple, back of head, acromion, iliac crest, anterior superior iliac spine (ASIS), posterior superior iliac spine (PSIS), thigh cluster, shank cluster, 5th metatarsal head, 1st metatarsal head, heel. Placement also on C7, clavicle, sternum, right back, and T10-T8 triad. (b) Approximate locations of EMG sensors. Bilateral placement on the thoracic paraspinals, lumbar paraspinals, gluteus medius, rectus femoris, vastus lateralis, biceps femoris long head, medial gastrocnemius, and tibialis anterior.]

### 3.2.3. Data Processing and Analysis

Kinematic and GRF data were processed with a bidirectional 4th-order low-pass Butterworth filter with cutoff frequencies of 6 Hz and 10 Hz respectively. EMG data were demeaned, band-pass filtered between 30 and 500 Hz, rectified, and low-pass filtered (bidirectional 4th-order Butterworth) with a cutoff frequency of 10 Hz (Hanawa et al., 2017; Shiavi et al., 1998). STS
events were defined in Visual3D (C-Motion, Inc., Germantown, MD). The start and end of the STS motion were defined as the initiation of trunk forward velocity and termination of forward pelvis velocity respectively. EMG signals were normalized to this duration (0-100%) and their peak magnitude over this period. EMG signals for participants without TTA were normalized to the peak magnitude of the processed signal during STS, while the EMG signals for participants with TTA were normalized to the peak value of the processed signal for the prescribed alignment condition.

To compare muscle activity over percent of STS motion for all muscles, we calculated integrated EMG (iEMG) for each muscle as the integral of the processed signal. The processed signal was normalized to magnitude and time, meaning a muscle that was at peak activity (1 in normalized magnitude) for the entire duration of STS (0-100%) would have an iEMG value of 100 (normalized EMG • % STS). GRFs were normalized to body weight for each participant. iEMG values and peak GRFs following lift-off from the chair were averaged across trials and participants for people without TTA and for people with TTA for each alignment condition. Trunk-pelvis flexion extension angle and medial/lateral COM were calculated from a six degree of freedom kinematic model in Visual3D (C-Motion, Inc., Germantown, MD, USA) with Cardan sequence ZXY. The musculoskeletal model consisted of eight segments including the torso, pelvis, thighs, shanks, and feet. Medial/lateral COM was calculated as the weighted average of segment COMs based on segment masses relative to total body weight (Dempster, 1955). Trunk-pelvis flexion/extension angle and medial/lateral COM were averaged across participants and trials for people with and without TTA. Average trunk-pelvis angle and COM for people with TTA represent the average for all alignment conditions.

3.2.4. Statistical Analysis

We used separate one-factor (alignment), repeated measures ANOVAs to compare metrics across alignment conditions for people with TTA (α = 0.05). The ‘alignment’ factor had three levels, corresponding to the three alignment conditions: prescribed, medial and lateral. When significant differences were found, pairwise comparisons with a Tukey adjustment for multiple comparisons were used to determine the conditions that were significantly different from one another (overall α = 0.05). Differences between participants with TTA in their prescribed alignment condition and participants without TTA were assessed using unpaired t-tests (α = 0.05).
3.3. Results

EMG signals are sensitive and thus susceptible to artifact related to movement, condition of the skin, and other external sources (Konrad, 2006). EMG signal artifact was observed in bilateral TP and LP for both participants with and without TTA, and amputated side BF for people with TTA, and were excluded from the analysis. Artifacts in TP and LP signals were likely due to the curvature of the back, which affected skin contact, as well as sweat accumulation during the protocol. Artifact in amputated side BF was due to interference with the chair and prosthetic socket/liner in some cases. Amputated side VL was also removed from the analysis due to surface EMG sensor malfunction.

There was no significant difference in time to complete one STS motion between people without TTA (mean = 2.09, SD = 0.41 s) and people with TTA in the prescribed alignment condition (mean = 2.18, SD = 0.47 s) (p = 0.345), however there was an alignment effect on STS motion time (p < 0.001). For participants with TTA, STS was performed in a shorter period of time during the medial (mean = 1.86, SD = 0.31 s) and lateral (mean = 2.00, SD = 0.31 s) alignment conditions compared to the prescribed condition (p < 0.001 and p = 0.018, respectively).

3.3.1. Comparison of Participants with and without TTA

Participants with TTA had greater amputated side RF iEMG in the prescribed condition compared to participants without TTA in non-dominant side RF (Table 3.2, Figures 3.3 and 3.4, p < 0.001). Differences between people with and without TTA for bilateral GM approached significance (0.050 < p ≤ 0.100) with smaller intact side (p = 0.064) and greater amputated side (p = 0.076) iEMG in people with TTA compared to dominant and non-dominant sides in people without TTA, respectively. The difference between iEMG in dominant/intact side BF1h also approached significance (p = 0.097) with greater activity for people with TTA. While differences in iEMG for hip extensors and abductors between people with and without TTA approached significance, people with TTA did not have greater iEMG for all muscles and sides. Therefore, the hypothesis that there would be greater hip extensor and abductor activity for people with TTA compared to those without was not supported.

Participants with and without TTA both had trunk flexion and extension phases during the STS motion and the instant of lift-off from the chair followed the transition from flexion to extension (Figure 3.5). Following lift-off from the chair, people with TTA had greater intact side
peak vertical GRF and a smaller amputated side peak vertical GRF in addition to greater bilateral peak posterior GRF compared to those without TTA (p < 0.001) (Figure 3.6). In addition to flexion/extension phases, participants with TTA also demonstrated two distinct phases of medial/lateral body COM motion with the COM moving toward the intact side for the first half STS followed by the COM moving toward the amputated side for the last half of STS. Distinct medial/lateral COM phases were not observed in participants without TTA (Figure 3.5).

![](image)

Figure 3.3: Muscle average integrated EMG (iEMG) (standard deviation) over % of sit-to-stand for participants without TTA (dominant and non-dominant sides) and with TTA (intact and amputated sides) for the prescribed, medial, and lateral alignment conditions. GM = gluteus medius, RF = rectus femoris, VL = vastus lateralis, BFh = biceps femoris long head, MG = medial gastrocnemius, and TA = tibialis anterior. ‘*’ indicates significant differences (p ≤ 0.050) compared to Prescribed and ‘▴’ indicates differences that approached significance (0.050 < p ≤ 0.100). TTA participants did not have sensors placed for amputated side MG and TA and amputated side VL and BF data were not included in the analysis due to signal artifact.
Figure 3.4: Average normalized EMG activity for participants without TTA (dominant and non-dominant sides) and with TTA for the prescribed (solid black), medial (red dashed), and lateral (blue dotted) alignment conditions (intact and amputated sides) over % of sit-to-stand (STS). GM = gluteus medius, RF = rectus femoris, VL = vastus lateralis, BFlh = biceps femoris long head, MG = medial gastrocnemius, and TA = tibialis anterior. TTA participants did not have sensors placed for amputated side MG and TA and amputated side VL and BF data were not included in the analysis due to signal artifact.
Figure 3.5: Average (standard deviation) trunk-pelvis flexion/extension angle with flexion and extension phases of STS for all participants (top). Average (standard deviation) medial/lateral body center of mass (COM) with directional COM phases for participants with TTA (bottom). Flexion and medial are positive.
Figure 3.6: Ground reaction forces (GRF) for participants without TTA (dominant and non-dominant sides) and with TTA (intact and amputated sides) for prescribed, medial, and lateral alignment conditions over % sit-to-stand (STS). Vertical, anterior/posterior (A/P), and medial/lateral (M/L) GRF were normalized to body weight (BW). Anterior, superior, and medial are positive.
3.3.2. Effect of Altered Alignment

There was a significant alignment effect for GM, RF, VL, BFh, and MG (Table 3.1). Significantly greater muscle activity in response to alignment shifts were only observed in the medial alignment condition for the intact and amputated side RF and intact side VL (post hoc pairwise comparisons, $p < 0.001$) (Table 3.2, Figure 3.3). Significantly smaller muscle activity was observed for intact side BFh in the medial and lateral conditions, the intact side MG in the medial condition, and the amputated side GM in the lateral condition ($p < 0.001$). Differences in muscle activity in response to alignment changes were dependent on muscle and direction of alignment change (Figures 3.3 and 3.4). Therefore, the hypothesis that there would be greater muscle activity, specifically in knee extensors, for alignments that deviate from the prescribed alignment was not supported across all muscles. However, this hypotheses was supported in the medial condition for the RF bilaterally and the intact side VL.

Table 3.1: Main effect of alignment for integrated EMG (iEMG) values and peak ground reaction forces (GRF) following lift-off from the chair. '-' indicates a significant effect was not observed. n/a indicates that amputated side muscle activity was either not recorded or not included in the analysis due to artifact.

<table>
<thead>
<tr>
<th>ANOVA Main Alignment Effect for:</th>
<th>Intact Side</th>
<th>Amputated Side</th>
</tr>
</thead>
<tbody>
<tr>
<td>iEMG</td>
<td></td>
<td></td>
</tr>
<tr>
<td>GM</td>
<td></td>
<td>&lt; 0.001</td>
</tr>
<tr>
<td>RF</td>
<td>&lt; 0.001</td>
<td>&lt; 0.001</td>
</tr>
<tr>
<td>VL</td>
<td>&lt; 0.001</td>
<td>n/a</td>
</tr>
<tr>
<td>BFh</td>
<td>&lt; 0.001</td>
<td>n/a</td>
</tr>
<tr>
<td>MG</td>
<td>0.001</td>
<td>n/a</td>
</tr>
<tr>
<td>TA</td>
<td>-</td>
<td>n/a</td>
</tr>
<tr>
<td>Peak GRF</td>
<td></td>
<td></td>
</tr>
<tr>
<td>A/P</td>
<td>&lt; 0.001</td>
<td>&lt; 0.001</td>
</tr>
<tr>
<td>Vertical</td>
<td>&lt; 0.001</td>
<td>0.026</td>
</tr>
<tr>
<td>M/L</td>
<td>&lt; 0.001</td>
<td>0.008</td>
</tr>
</tbody>
</table>
Table 3.2: Mean (standard deviation) integrated EMG (iEMG) for participants without TTA (no TTA) and with TTA for prescribed, medial, and lateral alignment conditions. Significant differences compared to Prescribed are indicated by ‘**’ (p ≤ 0.050) and differences that approached significance are indicated by ‘*’ (0.050 < p ≤ 0.100). D/I = Dominant/Intact side, nD/A = non-dominant/amputated side.

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Side</th>
<th>iEMG (Normalized EMG • % STS) Mean (SD)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>no TTA</td>
</tr>
<tr>
<td>GM</td>
<td>D/I</td>
<td>33.46 (7.83) ▲</td>
</tr>
<tr>
<td></td>
<td>nD/A</td>
<td>30.89 (9.65) ▲</td>
</tr>
<tr>
<td>RF</td>
<td>D/I</td>
<td>16.34 (6.61)</td>
</tr>
<tr>
<td></td>
<td>nD/A</td>
<td>15.74 (6.60) *</td>
</tr>
<tr>
<td>VL</td>
<td>D/I</td>
<td>20.25 (6.23)</td>
</tr>
<tr>
<td>BFh</td>
<td>D/I</td>
<td>24.68 (5.92) ▲</td>
</tr>
<tr>
<td>MG</td>
<td>D/I</td>
<td>23.94 (8.20)</td>
</tr>
<tr>
<td>TA</td>
<td>D/I</td>
<td>16.31 (4.81)</td>
</tr>
</tbody>
</table>

Table 3.3: Mean (standard deviation) peak GRF following lift-off from the chair for participants without TTA (no TTA) and with TTA for prescribed, medial, and lateral alignment conditions. Significant differences compared to Prescribed are indicated by ‘*’ (p ≤ 0.050). D/I = Dominant/Intact side, nD/A = non-dominant/amputated side.

<table>
<thead>
<tr>
<th>Side</th>
<th>no TTA</th>
<th>Prescribed</th>
<th>Medial</th>
<th>Lateral</th>
</tr>
</thead>
<tbody>
<tr>
<td>A/P</td>
<td>D/I</td>
<td>0.060 (0.012) *</td>
<td>0.076 (0.033)</td>
<td>0.087 (0.036) *</td>
</tr>
<tr>
<td></td>
<td>nD/A</td>
<td>0.064 (0.013) *</td>
<td>0.120 (0.034)</td>
<td>0.129 (0.036) *</td>
</tr>
<tr>
<td>Vertical</td>
<td>D/I</td>
<td>0.637 (0.068) *</td>
<td>0.746 (0.072)</td>
<td>0.778 (0.101) *</td>
</tr>
<tr>
<td></td>
<td>nD/A</td>
<td>0.633 (0.047) *</td>
<td>0.569 (0.059)</td>
<td>0.582 (0.083)</td>
</tr>
<tr>
<td>M/L</td>
<td>D/I</td>
<td>0.073 (0.023)</td>
<td>0.073 (0.021)</td>
<td>0.066 (0.020) *</td>
</tr>
<tr>
<td></td>
<td>nD/A</td>
<td>0.077 (0.023)</td>
<td>0.070 (0.015)</td>
<td>0.063 (0.014) *</td>
</tr>
</tbody>
</table>

There was a significant alignment effect for bilateral peak posterior, vertical, and medial GRF following lift-off from the chair (Table 3.1). Compared to the prescribed condition, people with TTA had greater peak posterior GRF following lift-off in the medial condition for both the intact and amputated sides (p < 0.001) and in the lateral condition for the intact side (p = 0.003) (Table 3.3, Figure 3.6). People with TTA also had greater intact side peak vertical GRF.
following lift-off for both the medial and lateral conditions compared to prescribed (p < 0.001). There was a significant alignment effect for the amputated side, but pairwise comparisons showed that no significant differences occurred between altered alignment conditions and the prescribed condition. There was smaller bilateral peak medial GRF following lift-off from the chair for the medial condition compared to the prescribed condition (p = 0.009, intact; p = 0.013, amputated).

3.4. Discussion

The purpose of this study was to characterize differences in muscle activity between people with and without unilateral TTA during STS and to investigate the effects of medial/lateral prosthetic alignment shifts on muscle activity in people with TTA. Further, differences in muscle activity were interpreted in light of trunk-pelvis flexion/extension angle, medial/lateral COM position, and GRFs during STS to provide insight into movement coordination. Differences in lower limb muscle activity in people with TTA compared to those without TTA may be a result of the medial/lateral COM shift observed in people with TTA during STS (Figure 3.3). STS is a complex, whole-body transfer task that requires horizontal and vertical acceleration of the body COM, which is primarily driven by a period of trunk flexion, followed by trunk and knee extension. Participants with and without TTA both had distinct flexion and extension phases which are characteristics of the STS motion and previously well-defined (Kralj et al., 1985; Millington et al., 1992; Schenkman et al., 1990), however the frontal plane movement of the medial/lateral body COM observed in people with TTA was not present for people without TTA. This result highlights the importance of analyzing STS in three dimensions, particularly in impaired populations.

The movement of the COM toward the amputated side for people with TTA was accompanied by increasing amputated side and decreasing intact side vertical GRF (Figure 3.4). This asymmetry in GRFs for people with TTA has been previously observed during STS (Actis et al., 2018; Agrawal et al., 2011; Ferris et al., 2017; Özyürek et al., 2014) and the change in limb loading as well as COM motion is consistent with the limb load/unload strategy used for medial/lateral postural control (Winter et al., 1993). While there is asymmetry in peak vertical GRF following lift-off from the chair, this loading becomes more symmetric as weight is distributed between limbs and the COM moves back toward the amputated side for standing.
For muscles where significant differences in iEMG between people with and without TTA were observed (Figure 3.3), the largest differences occurred during the phase where the COM was moving toward the amputated side and the vertical GRF was becoming more symmetric in people with TTA (Figures 3.4 and 3.6). During the time of shifting COM, the loading and unloading of limbs is controlled by hip abductor/abductor muscles with increased activity in hip abductors on one side increasing the load supported by that limb (Winter et al., 1996). Therefore, the smaller intact side and greater amputated side iEMG for GM (compared to dominant/non-dominant in people without TTA) that approached significance may reflect the necessary changes in hip abductor activity driving the hip load/unload mechanism. The shift of the load from the intact limb to the amputated limb during the latter half of STS, illustrated in the vertical GRF (Figure 3.6) and movement of the body center of mass (Figure 3.5), requires greater abductor activity in the amputated side because that side is being loaded.

In addition, there was significantly greater amputated side RF iEMG in people with TTA compared to non-dominant side RF activity in people without TTA, which was driven by prolonged activation throughout the movement. The biarticular RF contributes to both hip flexion and knee extension, and is a primary muscle responsible for braking during the latter half of STS (Caruthers et al., 2016). Therefore, the change in RF activity may be a result of the loss of amputated side ankle muscles and subsequent need for an increased contribution of other muscles to braking the body COM during the latter part of STS (Caruthers et al., 2016). Similarly, both the plantarflexors and BFllh contribute to forward acceleration of the COM during STS (Caruthers et al., 2016). The greater iEMG in intact side BFllh in people with TTA compared to the dominant side in those without TTA is consistent with an increased reliance on the hip extensors to compensate for the loss of functional plantarflexors.

We also hypothesized that people with TTA would have altered muscle activity for alignments that deviated from the prescribed, and that there would specifically be greater knee extensor activity under altered alignment conditions. Changes in muscle activity in response to alignment shifts were dependent on muscle and direction of alignment shift, therefore our hypothesis was not supported for all muscles, alignments, and sides. The increased bilateral RF and intact side VL activity in the medial alignment condition compared to the prescribed condition is consistent with the increased difficulty, measured by increased medial/lateral COM displacement, associated with maintaining medial/lateral postural control with narrow stance.
widths (Bonnet et al., 2014). The medial alignment condition reduces the overall base of support and therefore may require greater activity in RF to stabilize the body during STS. The transition from sitting to standing, or time around lift-off from the chair, in STS requires co-contraction of the hip and knee muscles (Roebroeck et al., 1994). Further, the flexed posture and required co-contraction around the time of lift off from the chair in STS may contribute to stabilization during the task (Beyaert et al., 2008; Seyedali et al., 2012). While greater RF activity may be needed for stability throughout the STS movement, the prolonged activity may also compromise socket comfort by increasing the pressure on the distal anterior tibia of the residual limb.

Changes in muscle activity in response to alignment shifts were also observed in the amputated side GM. Lower amputated side GM activity in the lateral condition compared to the prescribed condition is consistent with the reduced EMG magnitude observed in wide stance particularly for proximal muscles (Henry et al., 2001). Some people with a lower limb amputation may have limited mobility and use a prosthesis for transfer tasks only, such as getting into/out of a wheelchair (Campbell and Ridler, 1996). Therefore, the reduction in GM activity for lateral alignments suggests that lateral alignments for transfer-only prostheses may be beneficial in reducing the effort associated with the transfer task.

Changes in MG activity in response to alignment may be influenced by the loss of functional ankle muscles as well as postural control strategies used for frontal plane stability. Medial/lateral movement and stability are generally controlled at the hips, however there is a mechanical coupling of the hips and ankles during lateral movement that is reduced with smaller stance widths (Day et al., 1993). Further, a greater challenge associated with maintaining medial/lateral stability due to decreased stance width (Day et al., 1993), soft surfaces, or closed eyes (Riemann et al., 2003) results in decreased corrective action at the ankles and increased corrective action at the hips. The loss of active control of the amputated side ankle due to amputation combined with the decreased reliance on ankle control for medial/lateral movement under challenging conditions may contribute to the lower MG activity observed for people with TTA in the medial alignment condition where the base of support is smaller. Just as significant differences in muscle activity between people with and without TTA are primarily a result of differences during the amputated side loading phase (50-100% of STS) and shift of the body COM in people with TTA, significant changes due to alignment shifts also occur during this time. Thus, the medial/lateral motion strategy observed in people with TTA is likely dependent on prosthetic
alignment. Alignment adjustments, specifically in the lateral direction, could help minimize differences in muscle activity in people with TTA compared to those without an amputation with potential to reduce overall muscular demand for people with TTA.

Decreased BFh activity in both the medial and lateral alignment conditions compared to the prescribed condition may be a response to the decreased total motion time observed in those conditions. Because the STS motion is being completed faster, greater braking is required to slow the body COM, which is reflected in the increased posterior GRF for the medial and lateral alignment conditions compared to the prescribed condition (Figure 3.6). The BFh is a main contributor to forward acceleration of the body COM during STS (Caruthers et al., 2016), and smaller activity may be a strategy to assist braking. In addition, a faster STS motion may also require less co-contraction, and thus less BFh activity. During the STS motion, the period of transition from sitting to standing (time surrounding the point of lift-off from the chair) involves complex co-contraction patterns of the quadriceps, hamstrings, and gluteus maximus (Millington et al., 1992). Studies of daily tasks such as walking and STS have indicated a possible relationship between task speed and level of co-contraction, with more co-contraction occurring at slower speeds (Bishop et al., 2005; Fallah-Yakhdani et al., 2012). A greater understanding of the effect of speed on co-contraction in people with TTA during STS would better indicate the potential contribution of changing levels of co-contraction on overall change in muscle activity in response to alignment shifts.

The results of this study are subject to potential limitations including the challenges associated with collecting EMG data. EMG sensors were not placed on the remaining portions of amputated side MG and TA to ensure socket comfort and other muscles (bilateral paraspinals and amputated side VL and BFh) were compromised by signal artifact, meaning there was no analysis of paraspinal muscle activity and limited analysis for amputated side limb muscles. Analysis of paraspinal muscles and additional amputated side limb muscles may give further insight into the load/unload mechanism of weight transfer observed in people with TTA and the potential role of trunk-pelvis motion in the medial/lateral shift of the COM. In addition, while factors such as chair and starting position were controlled during the protocol, there are many other determinants of the STS motion, including age and sex, which may influence the results of this study. Further, a standardized protocol enabled the comparison between participant groups and alignment conditions but the motion strategies used in daily life may be highly variable over
time and individuals. Therefore, the standardized STS motion observed in this study may not reflect STS performed in daily life. In addition, we chose to evaluate iEMG from normalized EMG signals because these measures incorporate both magnitude and duration. Signals were normalized to the peak value during the motion and the sit-to-stand motion is a sub-maximal task. Thus, differences between people with and without an amputation are largely due to the shape of the processed EMG signal, and conclusions cannot be made regarding the magnitude of the EMG signal between groups.

3.5. Conclusion

While STS is generally analyzed in the sagittal plane, the primary source of differences in muscle activity and associated whole body biomechanics in people with TTA compared to those without TTA was the medial/lateral motion driven by a limb load/unload mechanism controlled by the hips. These results indicate that frontal plane analysis of STS in people with an amputation is important for understanding altered motion strategies as well as the potential implications of these strategies on long-term health and mobility, including socket comfort and LBP development. Further, the altered medial/lateral motion strategy and underlying muscle activity are affected by the medial/lateral alignment of the prosthesis, which changes the muscular demands and balance control during STS. Generally, medial alignments result in a greater challenge associated with maintaining stability and thus greater muscle activity compared to prescribed alignments as well as greater differences from people without TTA. Therefore, prosthetic intervention, and lateral alignments in particular, may help reduce muscle activity differences in people with TTA compared to those without, as well as reduce muscular effort associated with the STS task for people with TTA.
CHAPTER 4
CONCLUSIONS AND FUTURE WORK

The purpose of this work was to characterize differences in muscle activity between people with and without a unilateral transtibial amputation (TTA) during sit-to-stand (STS) and to investigate the effects of medial/lateral prosthetic alignment shifts on muscle activity in people with TTA. People with TTA performed STS trials using their prescribed alignment as well as with altered alignments achieved by 10-mm medial and lateral translations of the prosthetic pylon. Total muscle activity (integrated EMG: iEMG) was compared across alignments and to people without TTA, and other quantities including ground reaction forces, trunk-pelvis flexion/extension angle, and medial/lateral center of mass (COM) were analyzed to provide insight into movement coordination. We hypothesized that people with TTA would have altered muscle coordination during STS compared to people without TTA, with greater hip extensor and abductor activity as previously seen during walking. In addition, because of increased rates of energy consumption in response to alignment changes during walking, we hypothesized that the muscle coordination for people with TTA would change under altered alignment conditions. Specifically, we hypothesized that there would be greater activity in muscles central to STS execution, specifically the knee extensors, for alignments that deviated from the prescribed alignment.

People with TTA had greater amputated side rectus femoris iEMG during STS compared to people without TTA. In addition, differences in gluteus medius coordination between people with and without TTA approached significance with decreased intact side and increased amputated side iEMG in people with TTA compared to the dominant and non-dominant sides in people without TTA. These changes in muscle activity magnitude over the duration of STS is in part a result of the medial/lateral COM shift observed in people with TTA during STS. The movement of the COM towards the amputated side was accompanied by the loading and unloading of limbs and is driven by altered hip abductor activity during the second half of STS, and was not observed in people without TTA. Further, the greater iEMG during STS in the intact side biceps femoris long head and amputated side rectus femoris (compared to dominant and non-dominant sides in people without TTA) is likely a response to the loss of functional ankle
muscles and subsequent need for contributions to forward acceleration of the body COM during the first half of the STS motion and a slowing the body COM at the end of the STS motion.

Just as for differences in muscle activity between people with and without TTA, changes in muscle activity in response to altered alignment conditions also occurred during the shift of the body COM toward the amputated side. This phase of movement is influenced by the ability to maintain medial/lateral stability, and therefore, increases in knee extensor activity and decreases in medial gastrocnemius activity in response to medial alignments is a response to the greater effort associated with maintaining stability with smaller stance widths and the necessary shifting of postural control from the ankles to the hips. Similarly, the smaller amputated side gluteus medius activity in the lateral alignment condition is a response to the reduced effort associated with maintaining stability in wider stance widths.

These results suggest that prosthetic intervention, and lateral alignments in particular, may help reduce muscle activity differences in people with TTA compared to those without, as well as reduce muscular effort associated with the STS task for people with TTA. Lateral alignments may be particularly beneficial for transfer only prostheses, which are used exclusively for transfer tasks such as getting into and out of a wheelchair, where medial/lateral postural control of the body center of mass is critical, and do not need to make considerations for other types of movement tasks. Further, these results indicate that while studies typically focus on sagittal plane motion during STS, frontal plane analysis is important to understand the movement strategies of people with TTA and how they may affect long-term health and mobility. For example, this study showed greater amputated side rectus femoris activity in the medial condition compared to the prescribed condition likely related to the greater challenge in maintaining medial/lateral stability. Prolonged RF activity may result in increased pressure on the distal anterior tibia of the residual limb, thus compromising socket comfort and potentially causing residual limb pain.

People with TTA have an altered frontal plane motion strategy during STS compared to people without TTA, which results in altered muscle activity compared to people without TTA and also contributes to the changes in muscle activity observed in response to alignment shifts. Future work characterizing changes in muscle activity in people with TTA who also experience residual limb pain will help to better understand how changes in muscle activity may cause pain development in the residual limb. Similarly, further investigation directed at understanding changes in paraspinal muscle activity in people with TTA, and across alignment conditions, as
well as an understanding of muscle activity in people with TTA and LBP could expand our
understanding of LBP development. Expanding our knowledge of changes in muscle activity in
people with TTA compared to those without, in response to alignment conditions, and in
response to pain can help to determine the potential relationships between altered muscle activity
and pain development. Ultimately, an understanding of the specific biomechanical sources of
pain development has the potential to positively impact therapy and device interventions to
reduce pain and improve quality of life for people with TTA.
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