Kinematics and 3-D Rotational Trunk Stiffness of the Unilateral Transfemoral Amputee across Five Activities of Daily Living

by

Stewart Chisholm

A thesis
presented to the University of Waterloo
in fulfilment of the
thesis requirement for the degree of
Master of Science
in
Kinesiology

Waterloo, Ontario, Canada, 2015

AUTHOR'S DECLARATION

I hereby declare that I am the sole author of this thesis. This is a true copy of the thesis, including any required final revisions, as accepted by my examiners.

I understand that my thesis may be made electronically available to the public.

Stewart Chisholm

ABSTRACT

Review: Research examining trunk biomechanics in the transfemoral amputee population has only recently begun to include analysis of trunk muscle activity. Based on the absence of knee extensor muscles in the prosthetic limb, the muscles of the hip and trunk will demonstrate compensatory activations. This is an important area of research given that there is an increased incidence of low back pain among transfemoral amputees.

Purpose: The overall purpose of this thesis was to explore movement compensations exhibited by unilateral transfemoral amputees throughout a series of activities of daily living, with a focus on trunk biomechanics. Rotational stiffness about the L4-L5 vertebral joint was recorded and contextualized by spine and lower limb motion, and trunk and hip muscle activity. This was used to identify areas of difficulty and/or risk of injury and LBP.

Methods: Twelve total control participants and four amputee participants participated in this study. The activity of daily living-based tasks included gait, step down, step up, sit-to-stand, and door pull. Lower limb and trunk angles, and trunk and hip muscle EMG were recorded for each participant, and these data were used to model 3-dimensional rotational trunk stiffness about the L4-L5 joint. The results, including individual muscles' percent contribution to the overall stiffness, were compared across tasks separately for controls and amputees, and across subject groups separately for each task.

Results: During the gait task, the frontal plane stiffness was greater during prosthetic single support phase in the absence of a load-bearing cane for the amputees. When stepping down, non-cane using amputees displayed a flexed trunk on impact, and only the microprocessor knee prosthesis displayed knee flexion after contact. The step up task resulted in greater lateral trunk bending from the amputee group, with increased stiffness from non-cane amputees. The amputee participants also showed more sustained trunk stiffness throughout the sit-to-stand task, with a local peak at seat-off.

Conclusions: The altered movement and stiffening patterns of the transfemoral amputees were in an effort to increase trunk stability and mitigate risk of injury. It was concluded that, during some tasks, an increased stiffness is required from amputees to perform the same movement as the control group. The most at-risk tasks are the step down and sit-to-stand tasks, though other tasks also displayed characteristics linked to low-back pain.

ACKNOWLEDGEMENTS

I would like to thank my supervisor, Dr. Stephen Prentice, for giving me this opportunity and for guiding me through the entire process. Over the course of the project, Dr. Prentice was always patient as we encountered challenges, and continued to be supportive when things took longer than expected. This project would not have been completed without his guidance, and I will be forever appreciative of his advice. I would also like to thank my committee, Dr. Andrew Laing and Dr. Stu McGill, for providing perspective and direction when things became unfocused or overwhelming.

Throughout the project, several people assisted with the recording of both control and amputee participants. I am grateful to the entire Biomechanics of Human Mobility lab for their assistance and support. A special thank you goes to Ben Cornish, Raymond Lau, and Glynnis Pardo for directly assisting with data collection.

There are also those outside the lab that deserve recognition for their contributions to this project. I would like to thank Stephanie Beadle for being an invaluable resource in recruiting the amputee participants. Her expertise allowed us to streamline the screening process for potential participants, and gave helpful perspective to incorporate into our analysis. Also, thanks to Jeff Rice for designing and constructing the resistive simulated doorframe used in during the collection process.

Lastly, I would like to thank and family and friends for tolerating my neuroses throughout the last few years. A particular thank you goes out to a certain mouse who is my greatest supporter, and was always there when needed. Thank you.

TABLE OF CONTENTS

LIST OF FIGURES

LIST OF TABLES

LIST OF ABBREVIATIONS

TFA: Transfemoral amputee
TTA: Transtibial amputee
ADL: Activity of daily living
LBP: Low back pain
SACH: solid ankle cushioned heel
MPK/NMPK: microprocessor knee/non-microprocessor knee
Tasks: **GT** (gait), **SD** (step down), **SU** (step up), **STS** (sit-to-stand), **DP** (door pull)
Events: **IC** (initial contact), **TO** (toe off), **Lf** (lead foot), **Tf** (trail foot), **PREP** (preparatory phase), **SO** (seat off), **HE** (hip extension), **STABLE**, **PULL** (door pull)
SS: single support (PSS – prosthetic single support, LfSS – lead foot single support)
Muscles: **ES** (erector spinae), **SLM** (multifidus), **LD** (latissimus dorsi), **RA** (rectus abdominus), **IO** (internal oblique), **EO** (external oblique), **GMax** (gluteus maximus), **QL** (quadratus lumborum)
F/E: flexion-extension (sagittal plane)
LB: lateral bend (frontal plane)
AT: axial twist (transverse plane)
A/P: anterior-posterior
ROM: Range of motion
GRF: ground reaction force
EMG: electromyography
LTM: lateral trunk musculature (IO, EO, QL)
BW: body weight
MVC: maximal voluntary contraction
CCI: co-contraction index
MFCL: Mobility functional classification level
AMP: Amputee mobility predictor
ANOVA: analysis of variance

CHAPTER 1 – Introduction

Maneuvering through environments designed for healthy populations, lower limb amputees face several logistical movement challenges daily. The complexity of the missing limb, coupled with the diversity of prosthetic limb-replacement solutions, make it challenging to predict the compensatory movement strategies that individual amputees will employ. The resultant biomechanical health of amputees is affected by these altered movements and, as such, is difficult to account for. This is made easier by the definition of motion patterns common among lower limb amputees. Variability within this population can be a result of amputee differences – including residual limb properties, comorbidities, and age or gender effects – and prosthesis differences – including sockets, suspension, and knee and ankle joint types (Schuch, 1992). All of these factors affect the likelihood of successful rehabilitation.

One of the primary objectives of amputee rehabilitation is to attain a level of mobility sufficient enough to perform standard activities of daily living (ADL). Amputee mobility is described in a 5-level functional classification system (MFCL), with K0 not having the "ability or potential to ambulate or transfer safely…and a prosthesis does not enhance quality of life", and K4 exceeding "basic ambulation skills, exhibiting high impact, stress, or energy levels" (Ip, 2007). These levels can be ascribed based on an Amputee Mobility Predictor (AMP) questionnaire. This includes sitting and standing reach and balance tests, as well as gait and stair tasks (Gailey et al., 2002). Successful performances of these tasks, however, can be highly variable based on amputee and prosthetic factors, and a high rating on the MFCL does not give any further information into specifics of amputee movement.

In order to achieve a level of K3, "typical of the community ambulator", transfemoral amputees (TFA) employ several common compensatory movements. During ambulation, these compensations include lateral trunk bending, prosthesis circumduction, vaulting, and uneven step length (Berger, 1992). The existing research has a strong foundation in gait-based tasks, though considering the wide range of ADLs, research into other movement tasks is marginal. In adding variables such as ramps or stairs, the available research remains ample, though is more specialized with regards to task performance and prostheses studied. Research examining bending, carrying, or twisting tasks – all common movements in everyday life – is lacking for the TFA population. The qualitative classification of TFA mobility using the AMP does try to account for tasks outside of gait and balance tasks (Gailey et al., 2002), but does little to describe the functional performance.

Much of the research into TFA movement focuses almost exclusively on lower limb joint kinematics and kinetics, while excluding muscle activity, and fewer still have reported on trunk kinematics. Studies examining trunk muscle activity during lower limb amputee gait tasks have only begun to appear very recently (Hendershot and Wolf, 2014; Yoder et al., 2015). This is problematic, as a common issue among TFAs is low back pain (LBP). A reported 64% of TFAs experience LBP, with 39% citing restricted physical activity levels as a result (Devan et al., 2012). As such, an investigation into trunk kinematics and muscle activation would seem appropriate. Based on the absence of knee extensor muscles, the hip extensors would, in theory, produce much of the lost extensor moment. As a result, trunk muscle activation patterns may be altered for certain movements; either to assist in propulsion and braking, or else to stiffen the trunk to allow for the altered distal movements. This compensatory effect has been observed in populations with reduced quadriceps force during gait (Li et al., 2013). One purpose of this thesis was to report TFAs' functional performance of a diverse series of tasks, and compare the tasks with regards to the mechanical demand on the trunk structure.

Manipulation of rotational trunk stiffness through muscle force is used to overcome instability of the spine (Brown and McGill, 2005; Graham and Brown, 2014). Dysfunction of the trunk stabilizing system – including active and passive components – can lead to LBP and injury (Panjabi, 1992; Cholewicki & McGill, 1996). Trunk stiffness and co-contraction are also proportionately correlated (Granata & Orishimo, 2001; Lee et al., 2006), and increased bilateral and flexion-extension co-contraction has been linked with LBP-development during both prolonged standing (Nelson-Wong & Callaghan, 2010) and sitting (Schinkel-Ivy et al., 2013).

The activation of muscles results in a muscular stiffness that helps stabilize the joints about which each muscle acts. Under any given set of conditions, the muscles surrounding the vertebral joints must provide sufficient stiffness as to stabilize the joints and prevent buckling of the spine (McGill, 2014). Though methods of ensuring elevated spine stiffness have been presented (O'Sullivan et al, 1997; Richardson et al., 1999), the role of stiffness in the effective rehabilitation in the TFA population, specifically, is not yet known. Given the hypothesized altered trunk muscle activations, another purpose of this thesis was to investigate the active rotational trunk stiffness and co-contraction of TFAs in relation to a healthy control group, and examine a possible link to the increased incidence of LBP among TFAs.

CHAPTER 2 – Research Purpose

The overall intention of this thesis was to explore movement compensations exhibited by TFAs throughout a series of ADL-based tasks, with a focus on trunk biomechanics. The tasks include gait (GT), step down (SD), step up (SU), sit-to-stand (STS), and door pull (DP). The first purpose of the study was to compare various data across the five ADLs for both a control group and a set of TFA participants. The goal was to identify potential differences in across-task trends between the two subject groups, which may help indicate relative difficulties that TFAs experience during some activities over others. A second objective of this thesis was to report broadly on the separate functional performances of each ADL and compare the TFA participants to the control group in an effort to define motion adaptations common to multiple TFAs. Understanding how TFAs perform both gait and non-gait-related tasks, and how the lower limbs and trunk segments interact, is important in identifying potential areas of difficulty or risk of injury.

Included in these comparisons are trunk and lower limb kinematics, trunk muscle activity, rotational trunk stiffness, and muscular co-contraction of TFAs in relation to a healthy control group and between tasks. Trunk motion that displays excessive stiffness or laxity is not ideal with respect to efficiency and injury prevention. As such, these measurements were used to briefly examine a possible link to the increased incidence of LBP among TFAs, and to provide context with which to assess methods of TFA rehabilitation. Because this study was only intended to be an initial overview of TFA trunk biomechanics during ADL-based tasks in which they are less frequently studied, many of the conclusions will require further, more specified research.

This being a preliminary, exploratory study, much of the resultant data could not be appropriately predicted. While some points of interest were outlined prior to the commencement of the project, much of the direction was to be dictated following data collection. In this regard, specific hypotheses were not established so as to not over-focus and incluence the interpretation of the results.

CHAPTER 3 – Literature Review

3.1 – Introduction to Transfemoral Amputees

In the United States alone, an estimated 60 000-100 000 lower limb amputations are performed each year (Harness & Pinzur, 2001; Dillingham et al., 2002). The various levels of lower limb amputation include partial foot amputation, ankle, knee, and hip disarticulation, and transtibial (TTA) and transfemoral amputation (Bowker & Michael, 1992). Within this demographic, one group of interest – transfemoral amputees (TFAs) – accounts for approximately one third of the population. Common causes for this level of amputation, in descending order of incidence, include vascular disease, trauma, malignant tumors, infections, and limb discrepancies (Mensch and Ellis, 1986; Gottschalk, 1992). Following surgery, TFAs typically exhibit several movement modifications.

Movement variability is prevalent within the TFA population, and can be a result of amputee differences – including residual limb properties, comorbidities, and age or gender effects – and prosthesis differences – including sockets, suspension, and knee and ankle joints (Schuch, 1992). Although each amputee experiences a unique combination of mechanical changes, there are several established strategies that TFAs use to compensate for all of these factors. Well established compensations, observed in a majority of TFAs, include increased lateral trunk bending, greater circumduction – wherein the prosthesis is moved laterally during swing phase to enhance ground clearance, vaulting – wherein the intact leg displays excessive early plantarflexion to assist with prosthetic ground clearance, hip hiking, and uneven step length (Mensch and Ellis, 1986; Berger, 1992; Karacoloff et al., 1992). Outside of these general compensations, there exist several postural and movement asymmetries in TFAs when compared to a control population.

The hip range of motion (ROM) for the prosthetic side of TFAs is smaller than that of the intact side, which is not significantly different from the hip ROM of the control groups (Gaunaurd et al., 2011). This is mainly affected by decreased hip extension, both at rest and when maximum hip extension is attempted. Other postural asymmetries may, in part, be due to the fact that 66% of TFAs have leg-length discrepancies, with the prosthetic limb being 0.63-2.54 cm shorter than the intact limb (Gaunaurd et al., 2011). In comparison to the controls, TFAs experience a slight pelvic tilt towards the side contralateral from the prosthesis (Sapin et al., 2008), as well as increased lateral trunk flexion (Jaegers et al., 1995). Further, task-specific movement changes will be described in later sections. All of these changes are a combined result of altered body and prosthesis mechanics.

3.2 – Prosthetic design

Companies that manufacture prostheses offer a wide variety of designs. It has been reported that the goal of the prosthesis manufacturer is to functionally mimic the missing section of the limb (Pitkin, 2010), and one must take several factors into account when selecting the right combination of prosthetic elements. Aspects to consider include socket and suspension design – important in supporting and containing the residual limb, as well as securing the connection between the residuum and the prosthesis – and knee joint and ankle-foot components, which are important features in controlling movement with the prosthesis (Schuch, 1992). Socket designs include quadrilateral, ischial containment, and flexible sockets, while suspension options are typically limited to suction or soft belt suspension. In some cases, multiple suspension solutions are employed, as a lack of proper prosthetic suspension can result in exaggerated gait abnormalities and a reduction in safety (Schuch, 1992).

Prosthetic knee joints include single-axis or polycentric-axis knees, weight-activated or manual stance phase knee locking mechanisms, and friction or hydraulic swing phase control mechanisms. A more expensive alternative to these components would be a microprocessor-equipped knee joint, which helps control knee joint dynamics by sensing and analyzing various kinetic and kinematic data, and allows for ambulation in more difficult situations (Michael, 1999). Prosthetic feet vary as well, with some feet allowing for ankle flexion, while others utilize a flexible sole (Kapp and Cummings, 1992).

3.2.1 – Biomechanical effects

Although certain socket and suspension types are preferred in different situations, they will typically be matched to the individual amputee by a professional and is less restrictive, financially. Accommodations such as the choice of the flexible socket to allow for substantial residuum volume changes (Schuch, 1992) can therefore be accounted for without significantly affecting the overall function of the prosthesis. As such, they do not have as considerable an effect on TFA movement relative to the knee joint or ankle-foot mechanism.

The majority of recent literature divides knee joints into microprocessor (MPK) and non-microprocessor (NMPK) units. Surveys have shown that using prostheses with microprocessor knee joints result in fewer stumbles and faster walking speeds on both even and uneven terrain when compared to non-microprocessor knees (Kahle et al., 2008). The same is true of ramp negotiation, as a result of both increased stride length and cadence (Burnfield et al., 2012). Using multiple rehabilitation assessment tools, TFAs also displayed improved stair navigation using the MPK (Hafner, 2007; Kahle et al., 2008). During a sit-to-stand task, the GRF of the intact leg was significantly different depending on the power generation in the prosthetic knee joint. The vertical GRF of the intact limb was lower with a powered knee when compared to a MPK, though the change in knee did not affect intact knee or hip power (Wolf et al., 2013). Comparing NMPK to MPK, gluteal muscle recruitment was not found to be significantly different (Burnfield et al., 2012).

With respect to the ankle-foot connection, it has been suggested that an ankle joint that produces genuine plantarflexion – as opposed to simulated, with a solid or semi-flexible ankle – allows for better absorption during heel strike; specifically in the TFA population (Schuch, 1992). A commonly prescribed variant is the solid-ankle cushioned-heel (SACH) foot, which does not allow for significant plantar- or dorsiflexion, but does allow for a cushioned heel strike (Kapp and Cummings, 1992), reducing loading of the hip joint.

3.3 – Amputee variability

Variability in how TFAs respond to rehabilitation, and the subsequent mechanical changes they exhibit, begins with the cause of the amputation. Reasons for transfemoral amputation are varied, with the highest incidence being vascular disease, and descending in frequency with trauma, malignant tumor, infections, and limb discrepancies (Mensch and Ellis, 1986). Immediately following hospital discharge, there were no differences found between vascular and traumatic amputee in mobility test scores, though the predictors of success differed between the two groups. For traumatic amputees, the presence or absence of a medical comorbidity was best at predicting functional outcome, while residual limb characteristics held more influence when assessing potential success of vascular amputees (Melchiorre et al., 1996). Traumatic amputees also tend to skew younger and more male, as a population, in comparison to vascular amputees. Each of the afore-mentioned factors – age, sex, medical comorbidities, and residuum traits – may contribute in some way to the variability of TFA movement.

3.3.1 – Sex effects

In healthy populations, electromyographical (EMG) differences in the leg, hip, and abdominals have been found between sexes for high athletic performance, but not significantly different during less strenuous weight bearing exercises (Bouillon et al., 2012). During rehabilitation exercises, males exhibited decreased hip extension and gluteus maximus (GMax) activation compared to females (Dwyer et al., 2010), but not significant differences were found in gluteus medius or quadriceps when performing a step down task (Cowan and Crossley, 2009). The little research that touches on sex differences in the TFA population does not get this specific, though a review of the existing literature affirms that sex and body mass are not associated with ambulatory abilities (Greive and Lankhorst, 1996).

3.3.2 – Age effects

Higher levels of physical activity are maintained in younger amputees, but this comes with drawbacks. Greater activity results in higher incidence of residual limb irritation or pressure problems, which contribute to limiting mobility (Butler et al., 2014). For elderly amputees, adequate physical fitness and the ability to maintain unipedal stance were seen as good predictors of successful rehabilitation (Schoppen et al., 2003; Hamamura et al., 2009). Age also has an effect on specific task mobility, as more elderly TFAs are less successful at stair navigation (Hobara et al., 2012). Generally, increased age is associated with a lower functional outcome, but it is not clear if this is fully a function of age, or else because of the increase in medical comorbidities (Greive and Lankhorst, 1996).

3.3.3 – Comorbidities and other predictors

Common comorbidities of transfemoral amputation include diabetes, spastic hemiplegia, and hip and knee osteoarthritis. The contributing effects that a diabetes comorbidity has on TFA patients' movement variability is unclear, with several studies producing mixed results (Greive and Lankhorst, 1996). Increased incidence of osteoarthritis in the intact hip and knee in TFAs is both partially caused and contributes to asymmetrical gait (Benichou and Wirotius, 1982; Struyf et al., 2009). Spastic hemiplegia, which affects between 8% and 18% of lower limb amputees, is correlated to less successful rehabilitation outcomes (Hebert et al., 2012). The results of this study also demonstrate improved success in the hemiplegia is less severe, or is on the side ipsilateral to the amputation.

Other predictors of variability in the limitations of mobile activities are muscle strength and balance (Raya et al.., 2010). Specifically, hip abductor strength is associated with increased prosthesis weight bearing and improved mobility (Nadollek et al., 2002). Time since amputation is a good general predictor of variability (Raya et al., 2010), while also providing information of specific task performance, such as stair ascent (Hobara et al., 2012). Motivation and social outcomes should also be considered when examining potential reasons for movement variability (Greive and Lankhorst 1996; Schoppen et al., 2003).

3.3.4 – Residual limb

The length of the residual limb has an effect on the lever function of the leg, the amount of available muscle tissue remaining for muscular control, and the surface area that contacts the socket, which influences both comfort and proprioception (Mensch and Ellis, 1986). The length of the residuum has been reported as decreasing the total energy expenditure of the amputee (Boonstra et al., 1994), and also showing no difference in energy use (Bell et al., 2014). Generally, a longer residuum can allow for more functional gait (Greive and Lankhorst, 1996), though it has also been reported that if the residual limb is at least 57% of the length of the intact limb, there are no clear gait alterations (Baum et al., 2008). An increase in residual limb length decreases frontal pelvis ROM during gait, supposedly as a function of increased intact functional muscles (Goujon-Pillet et al., 2008), while a shorter residuum has contributed to increased trunk excursion and pelvic tilt (Bell et al., 2013). In stair climbing, limb length was found to not be a limiting factor (Hobara et al., 2012), but during a sit-to-stand movement, residuum length correlated with intact hip moment. With a longer residuum, demand on the intact hip was decreased (Highsmith et al., 2015).

Shorter residual limbs also make for a more complicated prosthetic fit. Incorrect fitting of the prosthesis socket reduces function of the residuum (Mensch and Ellis, 1986), and has a negative effect on overall mobility (Butler et al., 2014). Other than length, swelling and muscle characteristics make this difficult to account for. The location and nature of muscle attachments varies with each residual limb (Mensch and Ellis, 1986). TFAs display high variability in residuum muscle activity, unable to maintain a maximum voluntary contraction of constant amplitude. As such, no clear answer as to optimal anchorage of musculature for the residuum has been accepted (Pantall et al., 2011). Hip flexors and extensors in the residuum also correlate temporally with increases in socket pressure, further accentuating the relationship between proper fit and mobility (Hong and Mun, 2005).

3.4 – Activities of daily living

In the course of the day, both control and TFA populations perform several basic tasks. Different activities emphasize different movements which may contribute to challenges in how TFAs execute these tasks. Common ADLs include gait, step navigation (employing mild lunging motion), sit-to-stand movements (crouching and trunk extension movements), and door opening (torso axial twist motion). Existing research has begun to show some differences in how TFAs go about accomplishing these activities.

3.4.1 – Gait

Within the TFA population, gait has the most well-established research base. During a typical, healthy gait cycle, each leg goes through a stance phase (60% of cycle) and swing phase (40%). Stance phase can be further divided into single support (SS) (40%) and double support (20%) phases. TFAs have a 1-5% shorter stance phase (Sapin et al., 2008), while single support (SS) times are significantly shorter for the prosthetic limb when compared to the intact (Schaarshmidt et al., 2012). Comparing TFA gait to a control population, the difference between the step times of the subjects' two legs was significantly more asymmetrical for TFAs (Nolan et al., 2003). With regards to the vertical ground reaction forces (GRF) of TFA during gait, the overall pattern is not significantly different from that of the control, yet the characteristic double peaks appear flatter (Sapin et al., 2008). The vertical GRF pattern is again similar between the prosthesis and intact limb, with the intact limb having a sharper slope, indicating that it was faster to accept the load (Schaarshmidt et al., 2012; Segal et al., 2006). Peak GRFs for the prosthetic limb were lower than the intact limb (Segal et al., 2006), and it was found that the vertical GRFs for TFAs' legs were significantly more asymmetrical when compared to the differences between the legs of the controls (Nolan et al. 2003). Antero-posterior (A/P) GRF was lower in TFA than the controls (Sapin et al., 2008), and that of the intact leg was also greater than the prosthesis, with the difference being more evident at higher speeds (Schaarshmidt et al., 2012). Compensatory asymmetries continue to be evident as the kinetics and kinematics of the limbs are examined.

The majority of ankle flexion and extension in the prosthetic limb is passive, and does not have a significant effect of propulsion. As such, the kinematic compensations of the knee and hip are important in analyzing the movement patterns of the TFA population. At the level of the knee, the prosthesis displays decreased flexion in both stance and swing (Sapin et al., 2008; Segal et al., 2006). During stance phase, the intact knee will give in to some flexion, while the non-microprocessor prosthetic knee usually employs a locking mechanism, keeping the knee full extended through most of stance phase (Kaufman et al., 2012). Hip kinematic patterns during gait are similar for both amputee and control populations, however the hip ROM in decreased for TFAs (Winter 1991; Segal et al., 2006; Gaunaurd et al., 2011; Kaufman et al., 2012). The hips also reach their maximum extension later for TFAs (60-65% of the stride) when compared to control (55%), though they never attain the same level of extension (Jaegers et al., 1995). Some research has reported no significant changes in hip abduction or adduction between TFAs and controls (Jaegers et al., 1995), however it has been shown that TTA hip abductor moments are lesser than their control counterparts during gait (Rueda et al., 2013).

At the hip, the prosthetic limb has a lower peak extensor moment during stance than the intact leg (Nolan et al., 2003); hypothesized as the prosthesis potentially acting as a pendulum, while the intact leg displays greater active propulsion. Contrary to this finding, Segal et al. found no significant difference in peak hip moments during stance phase, but did not comment further (Segal et al., 2006). Temporally, there is increased hip extensor work performed during early stance in the intact limb compared to the control (Seroussi et al., 1996), which may be a compensatory reaction to the lack of prosthetic ankle propulsion. Similar to the knee, the net hip extensor moment of the intact limb of TFAs was found to be greater than that of the control group, while the flexion moment was not (Nolan and Lees, 2000). The 3D motion of the body's centre of gravity (CoG) follows a similar pattern in both amputee and asymptomatic gait (Tesio et al., 1998), with vertical displacement being nearly sinusoidal in nature, and forward velocity displaying an equal and opposite change (Winter, 1991). TFAs, however, experience asymmetries in the extensor work used to propel the body forward. During the prosthetic stance phase, the work performed by the muscles to move the CoG forward decreases by approximately 66% when compared to the intact limb (Tesio et al., 1998). This is, again, hypothesized as a potential pendulum-like effect of the prosthesis as compared to the active propulsion of the intact leg during stance phase.

During the stance phase of the prosthetic limb in TFAs, the gluteus maximus is activated for a longer period of time in comparison to the control, while some upper leg muscles display a double activation that is not present in the controls' EMG recordings (Wentink et al., 2013). Intact limb stance is much the same as those of the controls' stance phases (Wentink et al., 2013), though coactivation of upper leg muscles was found to be greater in the TFAs' intact limbs (Bae et al., 2007/2009). During the prosthetic swing phase, the braking muscles – the gluteus maximus and medius – were reportedly activated earlier (mid-swing) compared to control (end of swing), and are active for a longer period of time (Wentink et al., 2013).

Regarding healthy trunk motion during gait, flexion in the sagittal plane results in approximately 6% of the total trunk ROM, while movement in the frontal and transverse planes result in 13-18% ROM and 21-37% ROM, respectively (Feipel et al., 2001). Given the compensations employed by TFAs, frontal plane motion is the most notable difference between subject groups. A greater pelvic tilt is seen in TFAs during gait, potentially as a function of dysfunctional pelvic drop and a hip hiking compensation (Michaud et al., 2000). The trunk generally shows increased frontal rotation towards the prosthetic side as an aid in prosthetic foot clearance (Berger, 1992; Jaegers et al., 1995). This prosthetic trunk lean has also recently been shown to appear in TTA gait, and occurs concurrently with increased L4-L5 joint contact force and increased contralateral oblique muscle force (Yoder et al., 2015). Generally, TFAs display greater asymmetry in trunk acceleration during gait (Tura et al., 2010), and lower limb amputees show larger ranges of low back muscle activity (Yoder et al., 2015). Rueda et al. (2013) has suggested a need for TFAs to strengthen the proximal musculature and practice frontal plane balance and coordination in order to improve gait.

3.4.2 – Step up/down

In terms of stair ambulation, the absence of pathology is defined by the navigation of stairs with increased confidence and fewer movement adaptations (Kahle et al., 2008), while the inability to descend stairs is associated with a drastic decline in mobility (Ayis et al., 2006). For healthy individuals, the gait cycle duration for both stair ascent and decent is longer than that of walking on level surfaces (Riener et al., 2002), with the cycle duration of stair ascent being longer than that of descent (Protopapadaki et al., 2007; Riener et al., 2002). The proportions of the cycle dedicated to stance (~60%) and swing (~40%) phases are similar for each type of gait (Protopapadaki et al., 2007; Riener et al., 2002). The percentage of the cycle attributed to single-stance, however, decreases with an increase in stair inclination while descending; not true while ascending (Riener et al., 2002).

At first contact to start the stance phase, level surface ambulation is traditionally defined by a heel strike, with the ankle in plantar flexion (Winter, 1991). For stair ambulation, a toe contact is employed, with the ankle in very slight plantar flexion during ascent, and pronounced plantar flexion during descent (Riener et al., 2002). During stair ascent, stance phase consists of knee and hip extension, andankle plantar flexion, while descent stance is the opposite, with knee and hip flexion, and ankle dorsiflexion (Protopapadaki et al., 2007). Overall, hip and knee joint flexion is increased during ascent as compared to descent (Protopapadaki et al., 2007), and flexion along with total joint movement range increases with increased stair inclination (Riener et al., 2002).

In comparison to one another, the vertical GRF during stair descent is greater at foot contact, while stair ascent has a, increased GRF at toe off (Protopapadaki et al., 2007). Taking into account the three main joints of the lower limb, there is a greater overall limb extensor moment, or support moment, during stair walking compared to a level surface to overcome an increased vertical movement demand, though the pattern is similar (McFayden and Winter, 1988). With increases in stair inclination during descent, knee moments are increased while hip moments are decreased (Riener et al., 2002). Overall, the knee extensors are the greatest energy generator for upwards progression, and the knee extensors along with the ankle plantarflexors are the greatest energy absorbers during stair descent (McFayden and Winter, 1988). At the hip, the muscle activity of the gluteus maximus is increased by approximately 50% during stair ascent, as compared to walking on a level surface (Himmelreich et al., 2008), with the muscle activity of the gluteus being greater during ascent than descent (McFayden and Winter, 1988).

Kinetics of TFA stair ambulation were calculated from data of microprocessor knee sample groups. The vertical GRF during stair descent was decreased in the prosthetic limb compared to the control group, though it followed a similar pattern, and the intact limb of the TFAs displayed an increased initial vertical GRF (Schmalz et al., 2007). During descent, TFAs lead uniformly with the prosthesis (Mensch and Ellis, 1986; Karacoloff et al., 1992), and the A/P GRFs are lower in that leading limb than the control population (Schmalz et al., 2007). This allows for the GRF vector to remain anterior to the knee joint during landing, preventing prosthetic knee buckling (Jones et al., 2006). Lower limb joint moments were found to be similar between the control group and the prosthetic limb (Schmalz et al., 2007), though there was less knee flexion at impact for TFAs, contributing to a stiffer limb at landing for TFAs compared to both control and TTA participants (Jones et al., 2006). For subjects using non-microprocessor knees, EMG analysis of the quadriceps and hamstrings showed significantly greater muscle activity in TFAs during stair descent compared to the control group (Bae et al., 2009).

Step-over-step stair ascent cannot commonly be achieved by TFAs without the use of a prosthesis equipped with a microprocessor knee. The majority of the current research on TFA stair ambulation uses sample groups with microprocessor knees and, as they are expensive and using one is not a realistic expectation for most TFAs, the research cannot be generalized over the population. Surveys have shown that using prostheses with microprocessor knee joints result in fewer stumbles and faster walking speeds (self-selected and fast-paced) on both even and uneven terrain when compared to non-microprocessor knees (Kahle et al., 2008). Based on the Montreal Rehabilitation Performance Profile Performance Composite Scores – which are defined by the ideal stair descent being 1 step/second with no taps or stumbles – 63% of those surveyed appeared to have improved stair navigation using the microprocessor knees, while the remaining subjects did not significantly change their score (Kahle et al., 2008).

In healthy populations, the maximum absolute trunk flexion angle is reportedly much greater during stair ascent in comparison to either descent or level gait (Krebs et al., 1992). During ascent, patients with total knee arthroplasty – exhibiting reduced quadriceps and hamstring force similar to TFAs – displayed no significant difference in sagittal trunk motion compared to controls (Bjerke et al., 2014). With LBP, sagittal motion was affected minimally, with few other changes (Keun Lee et al., 2011). Currently, research into TFA trunk motion during step tasks is lacking.

3.4.3 – Sit-to-stand

Along with the various terrains that TFAs must navigate on a day-to-day basis, they must perform other activities of daily living (ADL). One such ADL – the sit-to-stand or stand-to-sit maneuver – is considered ‘‘the most mechanically demanding functional task routinely undertaken during daily activities’’ (Riley et al., 1991). It has been reported that roughly 60 sit-to-stand actions are performed daily (Dall and Kerr, 2010), though most TFA research, again, focuses of prostheses with microprocessor knees and cannot be generalized for such an activity. For healthy subjects, one sit-to-stand cycle takes between 1.3 and 2.5 seconds (Nuzik et al., 1986), as the CoG follows a distinct anterior then vertical trajectory, with little arc (Roebroeck et al., 1994). During the first 35% of the cycle, the trunk flexes forward, and extends after 45% of the cycle has been completed, while the hip joint flexes for the first 40%, then extends for the remaining 60% (Nuzik et al., 1986). The knee extends throughout the sit-to-stand cycle, beginning dramatically around 25 % (Roebroeck et al., 1994), as the knee displaces anteriorly and slightly downward for the first half of the cycle, before moving back up and backwards (Nuzik et al., 1986).

The knee and ankle moments are almost always in a state of extension throughout the sit-to-stand cycle, while the hip exhibits an extensor moment from 30% onwards, after the initial forward flex of the trunk (Roebroeck et al., 1994). Some asymmetry in knee joint moments has been reported, with the left knee moment peak occurring prior to the right, though with a lesser magnitude (Schofield et al., 2013). From 40%-70% of the cycle, the right knee moment is larger than the left, while the left is greater than the right from 70% to the end (Schofield et al., 2013). This study, however, did not appear to take into account the participants’ dominant side.

In terms of muscle activation during the sit-to-stand movment, the gluteus maximus and hamstrings showed moderate muscle activity (10-20% MVC) beginning at seat off (~30% of the cycle) that was sustained until the end of the movement (Roebroeck et al., 1994). When performing a sit-to-stand-to-sit action, they were active until just beyond seat down (Ashford and De Souza, 2000). The knee extensors showed a pronounced increase in activity (50-80% MVC) at seat off, which then steadily reduced until the end of the sit-to-stand cycle (Roebroeck et al., 1994). The erector spinae would activate just prior to seat off and remain active until the end of the task cycle (Ashford and De Souza, 2000).

Within the TFA population, some asymmetry and kinetic values were reported for the sit-to-stand and stand-to-sit tasks. It was found that the vertical GRF of the prosthetic leg was significantly less than that of a control group, with the GRF of the intact leg was significantly greater for both the sit-to-stand and stand-to-sit movements (Highsmith et al., 2011). This dynamic was shown to change with differing capabilities of the prosthetic knee joints, as the intact leg GRF decreased with increased power generation from the prosthesis (Wolf et al., 2013). The intact knee moment showed no difference between TFAs and controls, however the intact hip moment was significantly greater than the control, while the prosthetic hip moment was unchanged (Highsmith et al., 2011). These asymmetries have been hypothesized to contribute to increased instability during TFA STS movement (Gao et al., 2011).

3.4.4 – Door pull

Another common ADL is that of opening doors. With both pulling and pushing motions, and sometimes working against resistance, it can be a relatively strenuous task that does not appear to have been researched in any detail. The closest topic would be that of pushing and pulling objects along floors, but this does not require the same type of foot planting and rotational movement. For a population such as TFAs, the asymmetrical planting create greater challenges for trunk musculature.

3.4.5 – TFA Rehabilitation

Amputee rehabilitation is mainly focused on development of strength, sensation, balance, coordination, and ROM (Gailey and Clark, 1992). Before the TFA patient begins any task-specific rehabilitation, isolated muscle strength exercises are performed. Decreased ROM from muscle contractures is an eminent concern in this population and, as such, rehabilitation begins with residuum exercises to increase movement, strength, and flexibility, and encourage limb function (Mensch and Ellis, 1986; Gailey and Clark, 1992). From there, the patient moves on to balance and weight transfer exercises, including shifting weight in the anterioposterior and mediolateral directions, as well as twisting and reaching (Mensch and Ellis, 1986; Karacoloff et al., 1992). Progressing to gait training, TFAs begin by performing isolated step practice, stepping forwards, backwards, and laterally to improve prosthesis coordination and balance. The subsequent gait exercises are focused on foot clearance and stability, though the action that each amputee develops to achieve locomotion is different.

The remaining ADLs are taught in a similar way, developed to optimize success for a wide range of TFAs. Mensch and Ellis (1986) and Karacoloff et al. (1992) each outline standard protocols for performing several ADLs. For step descent, body weight is sustained with the intact limb while the amputee leads with the extended prosthesis and follows shortly with the intact limb. This is sometimes aided by a crutch or cane, usually supporting the prosthetic side. Stair ascent is the opposite, in that body weight is supported by the extended prosthesis, as the intact leg leads followed by the prosthetic limb. Step over step ascent is not regularly taught, though step over step descent can be achieved by agile TFAs, employing a ‘jackknife’ technique to flex the prosthetic knee. The STS protocol involves placing the prosthetic foot forward and the intact foot backward. The amputee then leans slightly forward and pushes upward with the help of crutches, canes, or armrests. During extension, the prosthesis is brought forward to be under the base of support. No specific methods of door opening were described, though the variability the performance of that task coupled with the differences in door mechanisms makes it difficult to standardize. Pre-gait training of twisting the upper body with planted feet would help with balance and coordination in this task.

3.5 – Rotational trunk stiffness

3.5.1 – Muscle force-stiffness relationship

Within the existing literature, there is a well-established relationship between muscle force, muscle stiffness, and rotational joint stiffness. Early modelling estimates of muscle stiffness accounted for independent muscles, theorizing the energy storage capacity of muscle cross-bridges (Zahalak, 1990), but did not model the muscle within an active-passive anatomical structure. Bergmark (1989) developed hypotheses linking muscle force, muscle stiffness, and elastic stiffness. Muscle stiffness was modelled to be proportional to muscle force and inversely proportional to its length (Bergmark, 1989), with increased muscle activation leading to a linear increase in muscle stiffness (Crisco and Panjabi, 1991). Later modelling made a point of accounting for passive structures, such as tendons and parallel elastic components, and reported a non-linear relationship between muscle force and stiffness, with the muscle stiffness slope becoming less steep with increased force (Cholewicki and McGill, 1995). Actively manipulating the non-linearity of the muscle force-stiffness relationship was found to result in joint rotational stiffness values decreasing past a critical force level (Brown and McGill, 2005). This was hypothesized to relate to unstable events leading to injury at high levels of muscle activation.

Generally, however, muscular stiffness was reported to stabilize joints against perturbation from external load and movement (Bergmark, 1989). Dysfunction of the stabilizing system may have three results: immediate compensatory response, long term adaptation, or injury (Panjabi, 1992).

3.5.2 – Trunk stiffness and stability

Rotational stiffness of the trunk has been repeatedly cited as an important contributor to spinal stability (Cholewicki and McGill, 1996; Cholewicki et al. 1999; Gardner-Morse and Stokes, 2001; Lee et al., 2006). Directly, increased muscle tension has also been linked to increased spine stability (Panjabi, 1992). Both passive and active tissues contribute to overall joint stiffness, though it has been reported that, in many movement situations, passive trunk stiffness – most notably the ligamentous spine – without active muscle support results in a lack of stability (Crisco and Panjabi, 1992; Brown and McGill, 2009). Muscles act to stabilize the joint when muscle stiffness itself is increased. As the muscles crossing a joint are activated, they act as spring becoming stiff, storing elastic potential energy. These stiff springs perform the function of cables or guy wires to compress the spinal column and generate increased stability (McGill, 2014). The geometry of the stiff, muscular 'guy wires' also has an effect on their stabilizing potential. Both a wide base of muscular support and muscles parallel to the compressive axis of the spine combine to help stiffen the vertebral joints.

This active trunk stiffness enhances the load bearing capacity of the spine (McGill, 2014), while a lack of stiffness can lead to spinal buckling (Choleqicki and McGill, 1996). This concept of insufficient stiffness has linked light, unloaded activity to a risk of injury. On the other hand, the increased muscle forces associated with increased active joint stiffness causes an increase in loading, and potential risk of bony tissue injury (Butler et al., 2003), while overloading of healthy structures or normal loading of weakened structures can lead to soft tissue injury and pain (Panjabi, 1992).

Increased joint stiffness has also been associated with increased performance in both the trunk and lower extremity (Butlet et al., 2003; McGill, 2014). During gait, the total body system stiffness was shown to increase linearly when external loads were applied incrementally (Caron et al., 2015). This was found to be a satisfactory compensation up to an additional 40% of body weight (BW) before the gait pattern was altered. In comparison, other high performance tasks utilizing repeated faster movements correlated to limited spine stability (Granata and England, 2006). Examining standing – a low performance task – has also been shown to be unstable even with increased stiffness if other factors, such as positive feedback loops, are absent (van Soest et al., 2003). This shows a link to how both movement patterns and sensory information affect the stabilizing effect of muscle stiffness.

3.5.3 – Co-contraction

Flexion of the lateral abdominal muscles was modelled to only slightly increase spine stability, though past 20% activation, no increase was reported (Stokes et al., 2011). Even so, trunk stiffness has been correlated to an increased voluntary recruitment of antagonistic musculature during both static standing postures (Granata and Orishimo, 2001) and isolated sagittal trunk extension (Lee et al., 2006). In postures usually associated with ADLs, increased trunk muscle co-contraction is proportionally related with trunk stiffness (Brown and McGill, 2008). Further excursion into the end ROMs, however, displayed reduced stiffness with greater activation for both flexion and lateral bend movements, while extension movements remained stiff (Brown and McGill, 2008). In lifting an unstable load, trunk muscle co-contraction increased to control stiffness and help stabilize the spine (van Dieen et al., 2003). During active response to perturbations, trunk co-contraction was also found to increase rotational stiffness (Vera-Garcia et al., 2006). In this case, it was hypothesized to help negate the need for complex muscular responses to disturbances.

A secondary effect of antagonistic abdominal contraction is the development of intra-abdominal pressure. Increased intra-abdominal pressure is thought to alleviate spinal loading and help stabilize the spine (Arjmand and Shirazi-Adl, 2006; Cholewicki et al., 1999). Indeed, modelling of intra-abdominal pressure has presented a proportional increase in spine stability (Stokes et al., 2011). Without active co-contraction, simply breathing has a positive effect on stiffness outcomes. Trunk stiffness was shown to increase in the F/E direction at both inspiration and expiration, correlating to changes in intra-abdominal pressure (Shirley et al., 2003).

3.5.4 – Movement effects

Joint stiffness helps maintain stability in the presence of perturbations from both external load and movement (Bergmark, 1989), with direction and timing of the stimulus playing a role in the effectiveness of the stiffening (Kavcic et al., 2004; Granata and England, 2006). Specifically for vertebral joints, the muscles acting around the joint must provide sufficient stiffness so as to prevent that joint from buckling and risking injury. A modest amount of activation is typically required to maintain sufficient stiffness during most daily activities (Cholewicki and McGill, 1996; Cholewicki et al., 2000). Commonly, ADLs require trunk musculature to be active only to approximately 5-10 %MVC (McGill, 2014). In pathological populations, however, the demand on the trunk musculature to maintain adequate stiffness appears to increase. In patients with LBP, erector spinae activity increases during gait, potentially as a result of spine instability (Ghamkhar and Kahlaee, 2015), while TFAs have demonstrated increased muscle forces for several stabilizing trunk muscles (Yoder et al., 2015). As mentioned prior, however, the geometry of the stiff, muscular ‘guy wires’ also has an effect on the stiffening reaction.

In accounting for movement perturbations during a series of rehabilitation exercises, Kavcic et al. (2004) found that the stiffening role of different muscles depended on the direction of motion. No muscle contributed more greatly over others, universally. Stiffening to stabilize the spine during external loading was found to also have a directional effect. With an active manipulation of spinal stability based on the changing height of an external load, co-contraction – and, in effect, stiffness – increased proportionally with an increase in load height and decrease in stability (Granata and Orishimo, 2001). In the frontal plane, EO and ES muscles displayed the largest responses to added lateral load, helping to initiate stiffness (Chiang and Potvin, 2001).

During reflexive responses to perturbations, passive trunk stiffness is not sufficient to eliminate risk of injury (Brown and McGill, 2009). Unknown perturbations have shown delays in muscle activation and stiffness (Cort et al., 2013), which increases the risk of injury due to instability. Activating the trunk musculature prior to a sudden perturbation was shown to increase the onset latency, but resulted in an overall decrease in magnitude of muscle activation while maintaining stability (Vera-Garcia et al., 2006). Preactivation has also been found to increase trunk stiffness (Chiang and Potvin, 2001). In this case, lower levels of trunk muscle preactivation were correlated to increases in reactive co-contraction.

Direction and timing of the perturbations also have a combined effect on trunk stiffness. During repetitive lifting exercises, muscular contribution to trunk stiffness and stability was lessened, with increased rate of lifting decreasing the measurements (Graham and Brown, 2014). Sudden loads applied to the transverse axis appear to be the most potentially damaging, as reactive stiffness is not sufficient (Vera-Garcia et al., 2006). However, if perturbations were anticipated, Cort et al. (2013) found that participants exhibited significantly larger joint rotational stiffness in the frontal and transverse planes.

3.5.5 – Low back pain and injury

Even with added heavy external loads, the force required for muscle stiffening is a main contributor to spine loading (McGill et al., 2009). As previously mentioned, this high loading may make one prone to injury, but insufficient stiffness is also an injury risk factor. TFAs, specifically, report between 64% and 81% incidence of LBP (Kulkarni et al., 2005; Devan et al., 2012), but pain and no-pain groups do not exhibit differences in disc degeneration (Kulkarni et al., 2005). This suggests that TFA LBP may affect targeted muscle groups over passive structures, marking a sensitivity to the active components of trunk stiffness. A proportional active stiffness correlate, increased co-contraction in healthy populations has been linked to development of LBP in both prolonged sitting (Schinkel-Ivy et al., 2013) and standing (Nelson-Wong and Callaghan, 2010), being thought of as a predisposing, and not adaptive, factor.

Pharmaceutically induced LBP has shown a decreased muscle contribution to trunk rotational stiffness and local dynamic stability, supposedly in an effort to avoid pain (Ross et al., 2015). By comparison, increased transverse trunk stiffness has been reportedly used as a guarding action in restricting trunk movement during gait in subjects with LBP (Selles et al., 2001). Restricted axial twist movement during gait is also a characteristic exhibited by TFAs (Mensch and Ellis, 1986).

3.5.6 – TFA

Research into stability or stiffness of the trunk in the TFA population is lacking. Generally, during gait, TFA subjects displayed less local dynamic stability, and increasingly variable and unpredictable trunk motions than a control group (Lamoth et al., 2010), and the STS movement required greater muscular effort to maintain trunk symmetry in the frontal plane (Gao et al., 2011). Lower extremity amputees also show increased and asymmetric peak forces and moments at the low back (Hendershot and Wolf, 2014). Specifically, lateral forces and lateral bend moments were 83% and 41% larger than the control for prosthetic and intact limbs, respectively. A theoretical increase in stiffness would be associated with these increased loads, and, coupled with the variability in trunk motion of some established ADLs, would suggest that TFAs may require additional trunk stiffness to maintain stability.

3.5.6.1 – Rehabilitation

No trunk stiffness rehabilitation programs unique to the TFA population have been reported. In a pain-developing population, however, core stability training programs focusing on deep abdominal muscle activation have reported shown a moderate lasting effect on reducing LBP (Puntumetakul et al., 2013), while gait based exercises have been shown to exhibit less low back loading than slow, restricted motion exercises (Callaghan et al., 1999). With this said, long term isometric core training has been recently shown to increase passive trunk stiffness with greater efficiency than dynamic training (Lee and McGill, 2015). The appropriate method of rehabilitative intervention for TFAs remains unclear, though movement stability was said to improve with intelligent microprocessor transfemoral prostheses (Lawson et al., 2011).

CHAPTER 4 –Methods

This study examined lower limb and trunk kinematics, and trunk muscle EMG to compare 5 ADL-based tasks for control and TFA participants. Prior to any task performance, each participant completed an Owestry Low Back Pain Disability Questionnaire in order to assess the presence of LBP that may affect the performance of the tasks. An example of the questionnaire can be found in Appendix A.

4.1 – Participants

Participants in this study were divided into three groups: a young, healthy control group (8 male, 2 female), an elderly, healthy control group (1 male, 1 female) and a group of 4 TFAs. Given the high level of variability amongst the transfemoral amputees, their data may be presented individually, as a series of case studies to be compared to the control group. Pertinent information for each TFA participant can be found in Table 4.1. All participants were injury-free within the past six months. This study received ethics clearance from the University of Waterloo's Office of Research Ethics review board prior to participant recruitment.

Table 4.1: Demographic information of transfemoral amputee participants.

Participant	Age	Prosthesis*	Anthropometry	Additional Information
TFA_01	59	**Side**: Left **Knee**: Microprocessor **Suspension**: Suction	**Height**: 1.78 m **Weight**: 90.9 kg	- Right ankle replacement (2 yrs) - Right hand cane
TFA_02	69	**Side**: Right **Knee**: Stance lock **Suspension**: Belt	**Height**: 1.77 m **Weight**: 86.6 kg	- 2 canes
TFA_03	49	**Side**: Left **Knee**: Hydraulic control **Suspension**: Suction & belt	**Height**: 1.81 m **Weight**: 79.2 kg	- N/A
TFA_04	34	**Side**: Left **Knee**: Hydraulic control **Suspension**: Suction	**Height**: 1.83m **Weight**: 75.0 kg	- N/A
**** All TFA participants had been using their current prosthesis for a minimum of 2 years.***				

4.2 – Experimental Setup

Lower limb and trunk kinematics and trunk muscle electromyography were recorded over a series of ADL tasks. Task events were defined using a series of kinematics and ground reaction force (GRF) data recorded from a offset cluster of four AMTI force plates (OR6, AMTI, Watertown, MA), sampled at 1024 Hz, located in the centre of the collection space (Figure 4.1). For various tasks, items were added to the force plate area, including blocks during the step tasks, a stool during the STS task, and a simulated door frame during the DP task. Throughout performance of each task, participants wore their own pair of shoes.

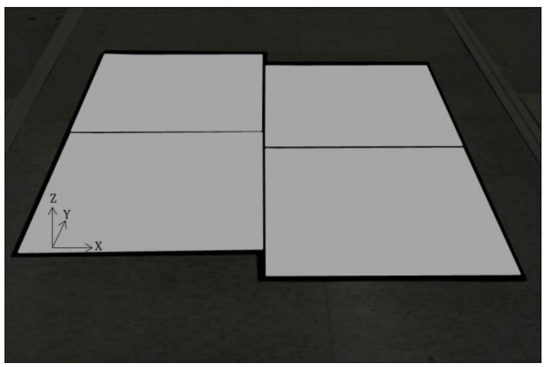

Figure 4.1: Offset configuration of force plates within the lab space

4.3 – Task protocol

The chosen tasks represent activities that are common in the daily life of TFAs. If any of the participants were visibly having trouble performing any of the tasks, or else mentioned their discomfort, their trials were noted. The activities studied were gait, step up and down, a sit-to-stand movement, and a simulated door opening task, the order of which were randomized. Each task was performed 3 times by the TFA participants and 5 times by the control participants. There was at least a 2 minute rest between performance of different tasks, or if the participant was feeling fatigued.

4.3.1 – Gait

These trials consisted of the participants walking the length of the collection space (approximately 7 m). They were aligned so that their left foot made contact with the first force plate in the cluster located in the centre of the collection area, while the following right foot made contact with a second, staggered force plate. This was done in such a manner as to not draw the participants' attention, in order to maintain a natural stride. Task events were defined using force plate GRF data or peaks of vertical foot velocity data if the event should occur off of the force plate (O'Connor et al., 2007). Task events included: right initial contact (RIC), left initial contact (LIC), right toe off (RTO), the second right initial contact (RIC2), left toe off (LTO), and the second left initial contact (LIC2), with a full trial being defined from RIC to LIC2. This encompassed one full step cycle from both the right and left limbs. The task was further subdivided into stance (IC-TO) and swing (TO-IC) phases, and single support phases (RTO-RIC; LTO-LIC).

4.3.2 – Step up/down

As step over step stair ambulation is not typically accomplished with non-microprocessor knee prostheses, the stair walking trials consisted of only one step. In this way, it may also simulate navigation of curbs and other everyday obstacles. On the cluster of force plates in the collection area, two 20 cm high blocks (Protopapadaki et al., 2007) were placed on separate force plates, alongside two other adjoining ground level plates. Each block was 40 cm wide and 40 cm deep, and they were placed with 0.5 cm of one another to simulate a connected, unbroken step. The step up task began with the participants standing on the ground level force plates. When prompted, they stepped up onto the raised blocks with both feet and stay there until stable. Similarly, the participants also performed trials stepping down on to the ground level force plates from the raised blocks. If the TFAs were unable to perform the step tasks without assistance, a note was made. Task events – measured exclusively from force plate GRF data – included: lead foot toe off (LfTO), lead foot initial contact (LfIC), trail foot toe off (TfTO), and trail foot initial contact (TfIC), with a full trial being defined from LfTO to TfIC. This encompassed one full step cycle from both the right and left limbs. The task was further subdivided into stance (IC-TO) and swing (TO-IC) phases, and single support phases (LfTO-LfIC; TfTO-TfIC). All event and phase definitions were common to both SD and SU.

4.3.3 – Sit-to-stand

Participants were positioned on a stool (0.46 m) without armrests. The stool was positioned on a force plate and each foot was also planted on separate force plates. When prompted, they rose from their seat in the way that felt most natural, staying standing for several seconds. Notes were made of whether or not the subject utilized their arms to assist with the movement. Task events were defined by the vertical GRF data, and they included: preparation (PREP), seat off (SO), full hip extension (HE), and stable stance (STABLE). PREP was defined by the vertical GRF recorded from the leg force plates declining 2.5% from their resting measurement. SO was found by matching the peak GRF of the leg force plates with a zero reading from the stool force plate. HE was defined by a GRF equaling participant BW following the first oscillation below said value after peak GRF was reached. The STABLE event was satisfied when the amplitude of the oscillations were within 2.5% of the participants' BW. Subdivisions of the task included preparatory phase (PREP-SO), rising phase (SO-HE), and stabilization phase (HE-STABLE), while a full trial was defined from PREP to STABLE. All task definitions were in accordance with the method utilized by Lindemann et al. (2007).

4.3.4 – Door pull

Using a weighted, hinged mechanism of standard height (1.17 m) positioned within a frame (width 1.2m) over the force plate cluster, participants performed trials of simulated door opening. They began the task standing off of the force plates. When prompted, they stepped forward, pulled open the door in the manner which felt most natural, and stepped through the frame and off the force plates. The full trial was defined from initial contact on the force plates to final TO following door opening.

4.4 – Data Collection

4.4.1 – Kinematics

Kinematic data were collected using an Optotrak motion capture system (NDI, Waterloo, ON), sampled at 64 Hz. Rigid bodies were placed bilaterally on the thigh, shank, and foot segments, and a rigid "dorsal fin" was placed perpendicular to the lumbar spine in order to track the segmental motion (Figure 4.3). Each rigid body consisted of five markers, while a digitizing probe was used to define anatomical landmarks for each segment.

For control participants and the intact legs of TFAs, digitized points on the thigh included the medial and lateral femoral epicondyles distally, as well as the greater trochanter, proximally. The shank rigid bodies defined imaginary points proximally at the tibial tuberosity, styloid process of the tibia, and fibular head, and distally at the medial and lateral malleoli. The foot segment rigid bodies were referenced to landmark the calcaneus, the heads of the 1st and 5th metatarsal, and the tip of the big toe. Digitization of the prosthetic limb defined sagittal joint centres, as well as distal foot and proximal socket of the prosthesis.

With respect to the trunk segment, the imaginary markers were the same for both TFAs and control subjects. The rigid body on the lumbar spine was used to landmark the 7th cervical vertebrae (C7), 1st lumbar vertebrae (L1), right and left acromion, and the jugular notch and xiphoid process of the sternum. Additional individual markers were placed on the right and left iliac crests, posterior superior iliac spines (PSIS), and L5 in order to help define the pelvis segment.

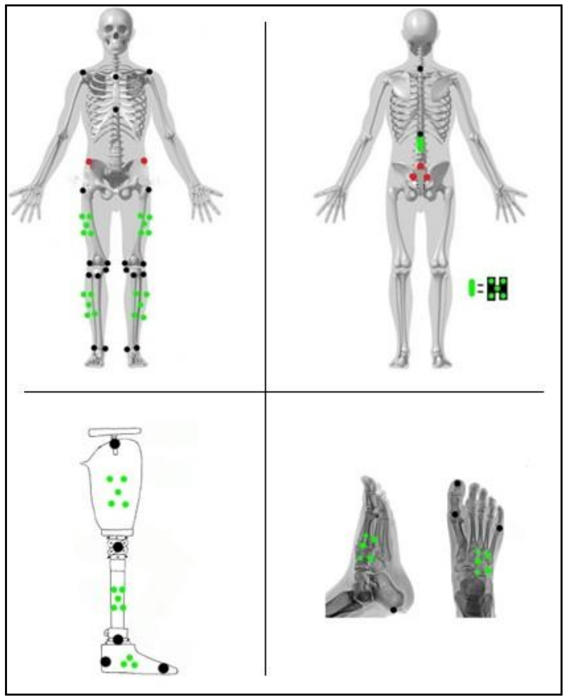

Figure 4.3: Motion capture marker placement (van Sint Jan, 2007). Green – rigid body markers; Red – individual markers; Black – digitized markers

4.4.2 – Electromyography

Pairs of surface EMG electrodes were applied bilaterally to the following muscles: rectus abdominus (RA), internal and external obliques (IO, EO), lumbar erector spinae (ES), latissimus dorsi (LD), superficial lumbar multifidus (SLM), and gluteus maximus (GMax), for a total of 14 EMG channels. Detailed placement locations can be found in Table 4.2 and Figure 4.4.

Table 4.2: Surface electrode placement locations.

Lumbar erector spinae (ES)	Location of largest muscle mass, approximately 4 cm from the midline at L3 (McGill, 1991; Drake et al., 2006)
Superficial lumbar multifidus (SLM)	At L5, parallel to a line connecting the PSIS and L1-L2 joint (Dankaerts et al., 2006; O'Sullivan et al., 2006)
Latissimus dorsi (LD)	Most lateral portion of the muscle at T9 (McGill, 1991; Drake et al., 2006)
Rectus abdominis (RA)	3cm lateral to abdominal midline, 2cm above the umbilicus (Drake et al., 2006)
External oblique (EO)	15cm lateral to the umbilicus at a 45º angle (McGill, 1991)
Internal oblique (IO)	Below the EO, just superior to inguinal ligament at a 45º angle (McGill, 1991)
Gluteus Maximus (GMax)	Midway between sacrum and GT, along the line between PSIS and posterior thigh (Hermens et al., 2000).

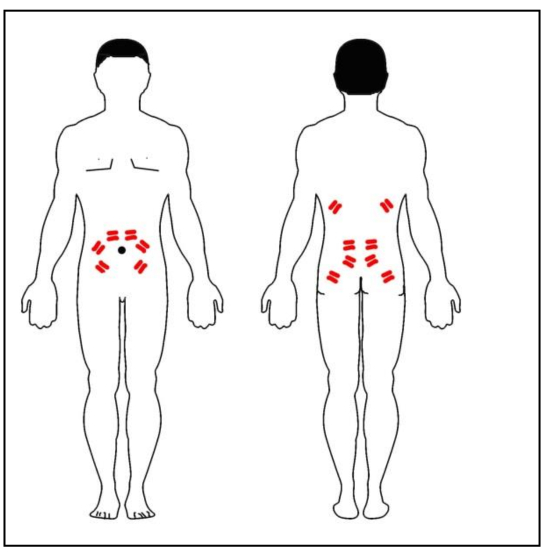

Figure 4.4: Surface EMG placement on trunk and hip musculature.

To calibrate the EMG data, a 60 second resting trial – consisting of lying supine on a therapy table with legs extended and arms loosely at their sides – was performed by each participant. This was followed by determining the maximum voluntary contraction (MVC) for each EMG channel. The MVCs for each set of muscles were found through a series of manually-resisted contractions, similar in technique to those reported by McGill (1991) and Drake et al. (2006). The MVC contraction task for the abdominal muscles (EO, IO, RA) consisted of enhanced sit-ups, with manually-resisted trunk flexion and axial twists to each side. The LES and SLM MVCs were found with a resisted back extension. The participants lay face down, with their trunk suspended over the therapy table, and attempted to attain a horizontal body position through resisted back extension after beginning slightly flexed. For the LD, participants placed their shoulders in abduction, with their elbows at 90º, and hands facing forward. They then pulled down against manual resistance, with the elbow moving in an inferior and slightly posterior direction. The trials for GMax consisted of manually-resisted hip extension. These trials were performed while lying on the side contralateral to that being tested, with hip extensions continuing past neutral position (Pantall et al., 2011). Each contraction task was performed up to 3 times, with the maximum value from the trials to be used later for EMG normalization. Between trials, a minimum 30s rest was allotted in order to not tire the muscles. All EMG data was recorded at a frequency of 1024 Hz.

4.5 – Data Analysis

Data were processed using Visual 3D V5 (C-Motion Inc., Germantown, MD). Kinematic data were filtered using a 2nd order dual pass Butterworth filters with a cutoff of 6 Hz (Winter, 1991). Missing data points were interpolated using a 3rd order cubic spline up to a maximum of 10 frames. A rigid link model was then constructed consisting of the trunk, pelvis, and bilateral thigh, shank, and foot. These segments were used to calculate relative angles bilaterally at the ankle, knee, and hip, as well as the angle of the trunk relative to the pelvis.

EMG data from the rest, MVC, and task trials were full wave rectified and filtered using high-pass and low-pass Butterworth filters with cutoff frequencies of 30 Hz (Drake and Callaghan, 2006) and 2.5 Hz (Brereton & McGill, 1998), respectively. Maximums were then found for each channel from the MVC trials. Average values were taken from the rest trial to determine resting activity level. Resting activity level was then subtracted from all EMG signals, which were then normalized to a %MVC.

Force plate GRF data were used for event definition of all tasks. Gait events occurring off the force plates were found using vertical foot velocity (O'Connor et al., 2007). The sit-to-stand task events were defined in accordance with the method described by Lindemann (2007), defining the beginning of preparatory phase, seat off, hip extension, and stabilization.

4.5.1 – Trunk Stiffness Model

Utilizing the normalized EMG and 3-D trunk angle data described prior, rotational trunk stiffness was modeled about the L4-L5 joint for every task using Matlab software (Mathworks, Inc., Natick, MA). Anatomical model data for 23 muscle fascicles were taken from Cholewicki and McGill (1996), and each fascicle's contribution to trunk stiffness about each of the joint's three anatomical axes – flexion/extension (F/E), lateral bend (LB), and axial twist (AT) – was calculated as the second derivative of the muscle's stored elastic potential energy (Potvin and Brown, 2005). The following equation is an example used to calculate the stiffness contribution of a muscle fascicle (m) about the z-axis:

$$S(m)_z = F_m * \left[\frac{A_x B_x + A_y B_y - r_z^2}{l} + \frac{q * r_z^2}{L}\right]$$

where

S(m)$_z$ = rotational stiffness contribution of muscle m about the z-axis of the joint in question (N*m/rad)
F_m = force of muscle m (all muscle forces balanced such that the net moment about each axis is zero)
l = 3-D muscle length from origin to insertion/node
L = muscle length (m) from origin to insertion
r = muscle moment arm (m)
q = proportionality constant relating muscle force and length to stiffness
$\langle A_x \quad A_y \quad A_z \rangle$ = muscle origin coordinates relative to the joint in question
$\langle B_x \quad B_y \quad B_z \rangle$ = muscle insertion/node coordinates relative to the joint in question
Note: a detailed account of the derivation of the above equation can be found in Potvin and Brown (2005).

Following this, the stiffness values from each fascicle were sorted and combined into their functional groups (ES, SLM, LD, RA, IO, EO, and QL), with muscle activity for QL predicted from the ES surface electrodes (McGill et al., 1996). Total stiffness about the x, y, and z axes were then calculated by summing the stiffness contribution values from each of the 7 muscles groups, bilaterally. Percent contribution values were then found for each muscle group through dividing the absolute stiffness contribution by the total rotational stiffness value in the respective plane.

The q value has been estimated in previous literature as ranging from 0.5-50, with an approximate mean of 10 (Cholewicki and McGill, 1995). In this study, q ranged non-linearly from 10 to 6.4 (Brown and McGill, 2005), representing 0%MVC to 100%MVC or maximal activation across all tasks. Using this method, Brown and McGill reported a critical q value of 6.8, wherein any value under it would be characterized by having a destabilizing force, though not enough to be dominant over the trunk stiffness created by such an activation. A more detailed breakdown of every component of the equation can be found in Appendix B.

4.5.2 – Co-contraction

As co-contraction has been associated with joint stiffness, bilateral and F/E co-contraction values were calculated for additional context. A co-contraction index (CCI) which accounted for magnitude and duration of activation was used, so as to compare total activity from the entirety of each task performance (Nelson-Wong and Callaghan, 2010). Co-contraction was calculated with the following equation:

$$CCI_{mp} = \sum_{i=1}^{N} (\frac{EMG_{low}}{EMG_{high}})(EMG_{low} + EMG_{high})$$

where

CCI_{mp} = co-contraction index for muscle pairing *mp* (%MVC)
EMG_{low} = muscle pairing EMG signal with the lower magnitude at each instant of time
EMG_{high} = muscle pairing EMG signal with the higher magnitude at each instant of time
N = number of data points recorded

One CCI value was calculated for each muscle group pairing, with total co-contraction values being an average of all muscle pairing CCIs in their respective category. For the bilateral CCI calculations, the muscles pairings were RES-LES, RSLM-LSLM, RLD-LLD, RRA-LRA, RIO-LIO, REO-LEO, and RGMax-LGMax. The F/E co-contraction muscle pairings were RES-RRA, RES-LRA, RES-RIO, RES-LIO, RES-REO, RES-LEO, LES-RRA, LES-LRA, LES-RIO, LES-LIO, LES-REO, and LES-LEO. The GT, SD, SU, and STS tasks were all performed with one or two gross movements, but the DP was a combination of several movements and, as such, took longer to perform. The DP CCI would then be skewed higher as a function of time, and comparison of values would be contextualized accordingly. CCI values were also normalized to the task length to give further context for comparison.

4.6 – Variables & Statistics

The data from this study were compared from two perspectives. The first comparison was between the 5 ADL-based tasks for both the control group and the 4 TFA participants. The independent variables in this case were the GT, SD, SU, STS, and DP tasks, while the dependent variables were the 3D trunk ROM, peak EMG activation of all muscles, 3D peak rotational trunk stiffness, bilateral and F/E co-contraction, and the percent contributions of each muscle towards the overall stiffness of the L4-L5 joint. Similar dependent variables were also compared between participant groups, with the control group compared to both the individual TFA participants, and the TFA group as a whole.

Trunk kinematics, EMG, stiffness, and CCI were individually compared, statistically, between-tasks for each subject group using one-way analysis of variance (ANOVA) tests. Within subject factors included the ADL-task, which consisted of GT, SD, SU, STS, and DP activities. Alpha was set to 0.05 prior to the collections. If significant differences were found, pairwise comparisons were performed using the Tukey post-hoc test.

Similar results were also compared between subject groups for each task using one-way ANOVA tests. Between-subject factors – participant groups – consisted of the control and TFA groups. Alpha was set to 0.05 prior to the collections. Because of the high variability of TFA participants, these statistical results were rarely used. Instead, the preferred method of comparison consisted of case-studies, wherein the control group was compared to each individual TFA participant.

A summary of the statistics performed can be found in Table 4.3. Generally, ROM was used as the predominant passive measure of trunk motion, while rotational stiffness was used as the active measure of trunk motion. These variables were compared non-statistically to provide a foundation upon which to interpret the recorded movements.

Table 4.3: Summary of statistical analyses performed on the recorded data.

Independent Variables		**Dependent Variables**	**Statistical Model**
Factor	**Level**		
ADL (within-subjects factor)	GT vs. SD vs. SU vs. STS vs. DP	**Kinematics**: Sagittal trunk ROM Frontal trunk ROM Transverse trunk ROM **Peak EMG activation**: RES, LES, RSLM, LSLM, RLD, LLD, RRA, LRA, RIO, LIO, REO, LEO, RGMax, LGMax **Rotational trunk stiffness**: F/E, LB, AT **CCI**: bilateral, F/E **Percent stiffness contribution**: F/E, LB, AT (*RES, LES, RSLM, LSLM, RLD, LLD, RRA, LRA, RIO, LIO, REO, LEO, RGMax, LGMax*)	One-way ANOVA (Tukey post hoc)
Participant group (between-subjects factor)	Control vs. TFA		One-way ANOVA

CHAPTER 5 – Results

For the all tasks, the right side is defined as the intact side in amputees, and the left side refers to the prosthetic side. The control group maintains their anatomical sides, with the exception of the SD and SU tasks. During SD, the left side denotes the lead leg side and the right refers to the trail leg side. The opposite is true for the SU task. TFA1 utilizes a cane in his right hand for the GT, SD, and DP tasks, while switching to his left hand for the SU and STS tasks. TFA2 maintains canes in both hands for each task.

5.1 – General Results

The control group was made up of 10 (8M, 2F) healthy university-aged participants. Given the wide range of TFA participant ages, an additional 2 (1M, 1F) healthy older control participants were recorded to provide reference data when accounting for age effects, but will not be included with the control group data. The average age of the control group – not including the 2 elderly participants - was 25.25 (2.49), and the average height and weight were 1.79 (0.05) m and 77.93 (12.90) kg, respectively. The elderly participants had average age, height, and weight measures of 68.8 (6.1), 1.66 (0.03) m, and 71.2 (15.2) kg. Information on each amputee participant's demographics can be found in earlier sections.

Each task took longer to perform for the TFA participants than the control group. These absolute values, along with the event markings as a percent of each normalized task cycle can be found in Table 5.1. For the GT task, the single support phase of the right and left legs for the control group were 26.2 (1.5) and 26.6 (1.3) % of the task cycle, respectively. For the amputees, prosthetic single support (PSS) phase lasted 22.2 (2.0) % of the task, while the intact single support phase accounted for 24.7 (1.6) %. The SD and SU task had the control participants unpredictably choosing which foot to lead with. The TFA group uniformly led with the prosthetic limb during SD, and with the intact limb during SU. While performing the STS movement, the TFA group demonstrated greater incidence of arm assistance – either with canes or pushing off of their thighs – than the control group. The TFA2 participant was unable to perform the STS movement at the regular height, and instead performed from an elevated massage table surface (0.64 m). The DP task was highly variable in its execution for both groups.

Table 5.1: Task length and event marking information for all tasks across subject groups.

Task	Group	Measure	Task length (s)	RIC (%cycle)	LIC	RTO	RIC2	LTO	LIC2
GAIT	**Control**	Mean	1.68	1	33	40	66	73	100
		SD	0.11	0	1.2	1	0.9	1.3	0
	Amputee	Mean	2.38	1	30	42	61	73	100
		SD	0.38	0	8.7	1.4	7.1	1.6	0
				LfTO (%cycle)	**LfIC**	**TfTO**	**TfIC**		
STEP DOWN	**Control**	Mean	1.19	1	48	66	100		
		SD	0.1	0	4.2	3.1	0		
	Amputee	Mean	2.2	1	53	79	100		
		SD	0.8	0	13.5	5.1	0		
				LfTO (%cycle)	**LfIC**	**TfTO**	**TfIC**		
STEP UP	**Control**	Mean	1.52	1	40	62	100		
		SD	0.25	0	5	3.2	0		
	Amputee	Mean	2.06	1	32	46	100		
		SD	0.59	0	4.7	3.8	0		
				PREP (%cycle)	**SEAT OFF**	**HIP EXT**	**STABLE**		
STS	**Control**	Mean	1.98	1	40	76	100		
		SD	0.23	0	4.1	8.4	0		
	Amputee	Mean	2.57	1	41	69	100		
		SD	0.3	0	4.9	3.7	0		
				START (%cycle)	**PULL**	**END**			
DOOR PULL	**Control**	Mean	3.24	1	72	100			
		SD	0.46	0	3.9	0			
	Amputee	Mean	4.9	1	64	100			
		SD	3.01	0	10.6	0			

5.2 – Comparison of tasks

In comparing tasks, the control and TFA participants were divided into their respective groups. Because of the potentially high variability of the TFA participants, it is more difficult to generalize whether certain tasks present different demands for all TFAs. This issue is discussed in a later section (see *7.3 – Limitations*).

Examining the between-task differences in trunk ROM displayed by the two subject groups, flexion/extension and axial twist trunk ROM were found to have trends that were most similar between the two groups. With this said, there were differences, most notably in the frontal plane. For the control group, significant differences in trunk ROM were found across the five activities of daily living around all three axes. In the sagittal plane (df=4, F=156.87, p<0.0001), STS had the significantly largest ROM, followed in descending order by SU and DP tasks, which were not significantly different from one another, GT, and SD tasks. In the frontal plane (df=4, F=71.84, p<0.0001), the DP task had the significantly largest lateral bend trunk ROM, followed by SU and GT, then SD and STS tasks. In the transverse plane (df=4, F=80.11, p<0.0001), the DP task again had the largest trunk ROM, followed by the GT task, with SD, SU, and STS tasks having similarly low ROMs.

The TFA group had fewer differences in trunk ROM across tasks. In the sagittal plane (df=4, F=95.17, p<0.0001), STS had a larger ROM than all other tasks, which showed no further differences. In the frontal plane (df=4, F=3.05, p=0.028), the STS and SD tasks were significantly different, while all other comparisons were not. In the transverse plane (df=4, F=27.60, p<0.0001), the DP task was significantly greater than all other tasks, while the GT task had a significantly higher ROM than the SD and STS tasks. A summary of these comparisons are displayed in Figure 5.1.

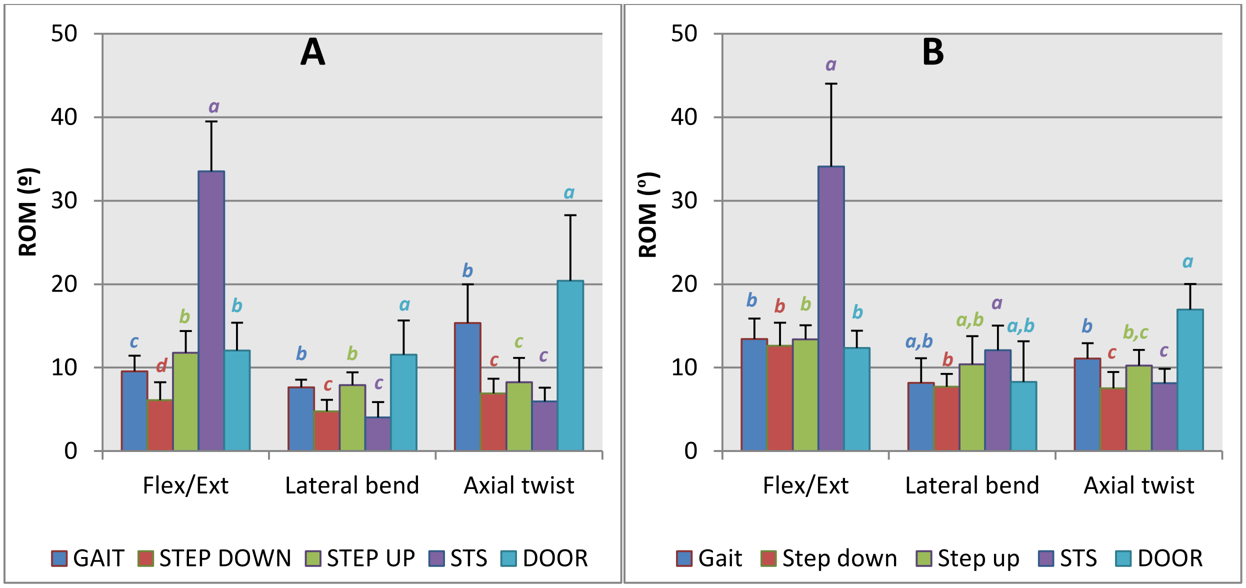

Figure 5.1: Comparison of activities of daily living in terms of the relative spine-pelvis ROM in 3-dimensions. A – Control group; B – TFA group. Similar lowercase letters denote non-significant differences.

Displayed in Tables 5.2 and 5.3 are the peak muscle activation values for all recorded muscles across all tasks, for the control and TFA group, respectively. Tasks were compared to one another within each muscle. Because of the great number of comparisons across 5 tasks in 14 muscles, not all pairwise comparisons will be mentioned in the text. All significant differences are indicated in Tables 5.2 and 5.3.

For the control group, RES, RSLM, LSLM, and RLD values were significantly higher (df=4, $p<0.05$) during the STS task than the remaining 4 tasks, while LES was significantly higher during STS than all tasks except SU. Activation of the LLD was significantly greater during DP than all other tasks except SU, which was, in turn, greater than SD. The SD task consistently had among the lowest peak muscle activation for the back musculature. The RRA had significantly higher peak activation during the SD task, while the LRA was lowest during STS. Both the RIO and REO values were also significantly greatest during the SD task. LIO and LEO revealed virtually no significant differences across tasks. Activation of the RGMax was significantly greater during SU compared to all tasks except for STS, while LGMax was greatest during GT, SU, and STS.

For the TFA group, RES peak muscle activation was greater during the SD task than the remaining tasks, though no significant difference was observed between tasks for the LES, RSLM, LSLM, RLD, or LLD. A similar lack of significant difference was present in the comparisons of the abdominal and gluteal muscles, with the exception of REO peak activation being greater during STS than GT, SU, and DP. As can be seen in Table 5.2, there are several muscles that have large differences between task means, though with such a small, variable group it is difficult to generalize without sufficient perspective. All significant differences are labeled in the tables.

For comparison, average muscle activations across all tasks for both subject groups are displayed in Figures 5.2 and 5.3. A statistical analysis of the average muscle activations was not performed, though the overall trend of activation between muscles followed a similar pattern for both the control and TFA groups. On the back, the SLM displayed slightly higher average activation than the ES or LD. For the abdominals, the average IO activation was slightly greater than that of the RA or EO. The GMax activations were about on par with the SLM values, with the TFAs displaying relatively higher values than the control group, possibly due to normalization error (see *7.3 – Limitations*). Between-task comparisons within each muscle displayed generally similar trends to the peak activation comparisons for both subject groups. Further contextualized muscular activity will be discussed in later sections.

Table 5.2: Average peak muscle activation (%MVC) for trunk and hip musculature across the five activities of daily living tasks for the control group. Similar subscript letters denote non-significant differences.

TASK	Measure	RES	LES	RSLM	LSLM	RLD	LLD	RRA	LRA	RIO	LIO	REO	LEO	RGMax	LGMax
GAIT	Mean	11.54$_{b,c}$	14.32$_{c}$	21.57$_{b,c}$	19.1$_{b,c}$	9.13$_{b}$	10.92$_{b,c}$	5.84$_{b}$	5.72$_{a,b}$	13.63$_{b}$	13.25	10.34$_{b}$	10.26$_{a,b}$	15.25$_{b,c}$	22.69$_{a}$
	SD	8.53	9.1	9.96	9.67	3.09	3.34	2.14	2.24	5.58	6.03	5.12	5.55	10.24	16.9
STEP DOWN	Mean	7.02$_{c}$	15.99$_{b,c}$	12.45$_{d}$	16.71$_{c}$	7.76$_{b}$	10.25$_{c}$	7.96$_{a}$	6.89$_{a}$	17.87$_{a}$	14.29	13.03$_{a}$	11.55$_{a}$	9.43$_{c}$	10.24$_{b}$
	SD	5.81	6.66	4.93	6.6	3.91	4.44	2.15	2.39	7.99	4.07	5.69	4.23	4.16	4.69
STEP UP	Mean	13.79$_{b}$	20.95$_{a,b}$	25.91$_{b}$	21.89$_{b}$	9.4$_{b}$	14.5$_{a,b}$	6.12$_{b}$	5.43$_{a,b}$	14.24$_{a,b}$	12.74	9.75$_{b}$	8.09$_{b}$	25.15$_{a}$	16.45$_{a,b}$
	SD	6.73	7.88	4.78	7.07	2.92	3.62	2.64	2.83	6.72	6.05	4.11	2.98	14.2	8.91
STS	Mean	22.68$_{a}$	26.53$_{a}$	32.74$_{a}$	28.74$_{a}$	13.82$_{a}$	11.69$_{b,c}$	5.21$_{b}$	4.69$_{b}$	9.23$_{c}$	11.26	8.04$_{b}$	8.93$_{a,b}$	19.47$_{a,b}$	21.23$_{a}$
	SD	13.35	13.67	9.86	8.64	6.98	6.25	2.91	3.16	3.52	4.05	3.4	3.98	12.77	14.81
DOOR	Mean	12.9$_{b}$	17.6$_{b,c}$	18.07$_{c}$	20.0$_{b,c}$	8.83$_{b}$	17.41$_{a}$	5.89$_{b}$	6.67$_{a}$	13.09$_{b,c}$	14.73	9.36$_{b}$	11.01$_{a}$	14.72$_{b,c}$	12.06$_{b}$
	SD	7.28	9.2	5.41	8.92	2.38	12.83	2.31	3.32	5.14	7.84	3.73	5.16	7.79	3.18

Table 5.3: Average peak muscle activation (%MVC) for trunk and hip musculature across the five activities of daily living tasks for the TFA group. Similar subscript letters denote non-significant differences.

TASK	Measure	RES	LES	RSLM	LSLM	RLD	LLD	RRA	LRA	RIO	LIO	REO	LEO	RGMax	LGMax
GAIT	Mean	19.27$_{b}$	56.79	67.56	63.66	52.41	42.81	26.26	19.34	45.24	52.34	21.19$_{b}$	71.6	170.44	42.35
	SD	4.65	10.2	45.47	28.45	17.82	19.68	15.93	10	27.26	16.08	10.67	71.39	124.67	15.19
STEP DOWN	Mean	58.42$_{a}$	47.62	58.86	84.12	81.12	46.23	16.74	28.1	36.38	40.95	60.31$_{a,b}$	33.03	54.81	204.2
	SD	19.63	19.18	32.64	19.53	49.82	14.11	6.89	10.35	14.26	14.42	81.93	13.39	29.3	132.83
STEP UP	Mean	23.44$_{b}$	67.02	90.47	73.5	46.41	102.82	32.94	25.02	25.86	59.26	21.7$_{b}$	76.64	167.25	76.58
	SD	5.25	7.18	60.58	25.53	13.55	74.82	19.7	15.14	15.21	25.43	7.91	59.6	93.6	34.22
STS	Mean	29.81$_{b}$	57.05	73.69	71.06	57.99	45.04	32.22	22.56	58.18	50.43	68.55$_{a}$	83.5	84.09	68.51
	SD	6.28	12.85	26.09	13.47	16.18	16.65	9.92	10.48	51.56	10.74	64.25	31.71	39.25	20.52
DOOR	Mean	20.98$_{b}$	65.27	62.59	66.5	48.73	52.57	27.08	23.02	30.47	45.55	22.63$_{b}$	64.41	115.5	44.59
	SD	5.39	15.27	35.68	28.13	31.56	13.64	15.44	15.13	17.11	16.18	15.02	56.52	58.54	14.23

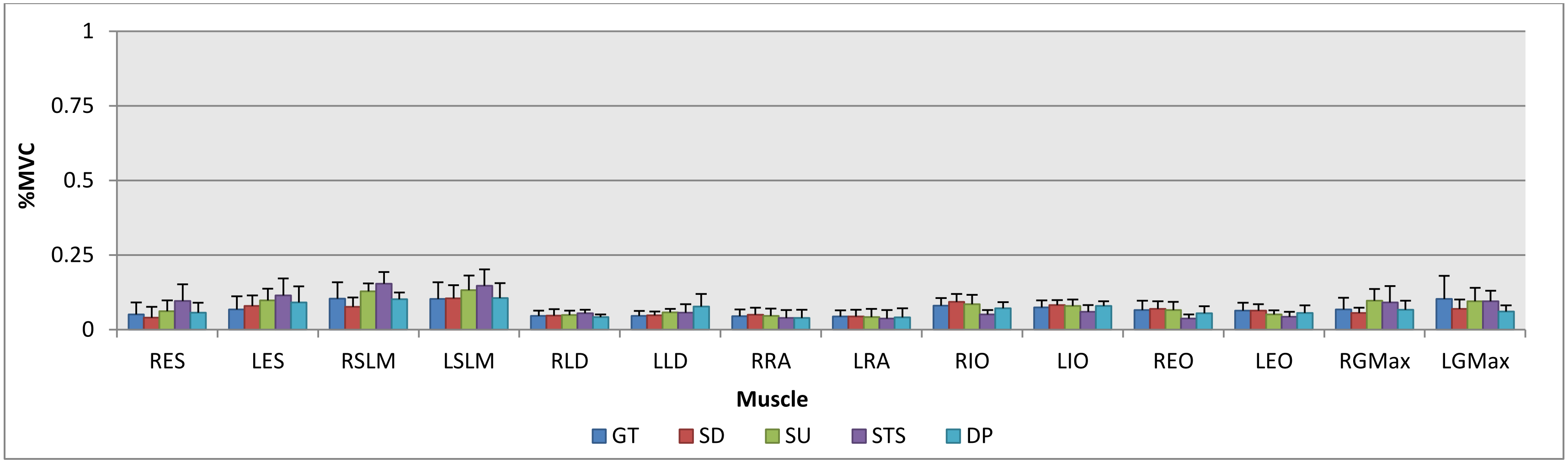

Figure 5.2: Comparison of average control group muscle activation during five activities of daily living.

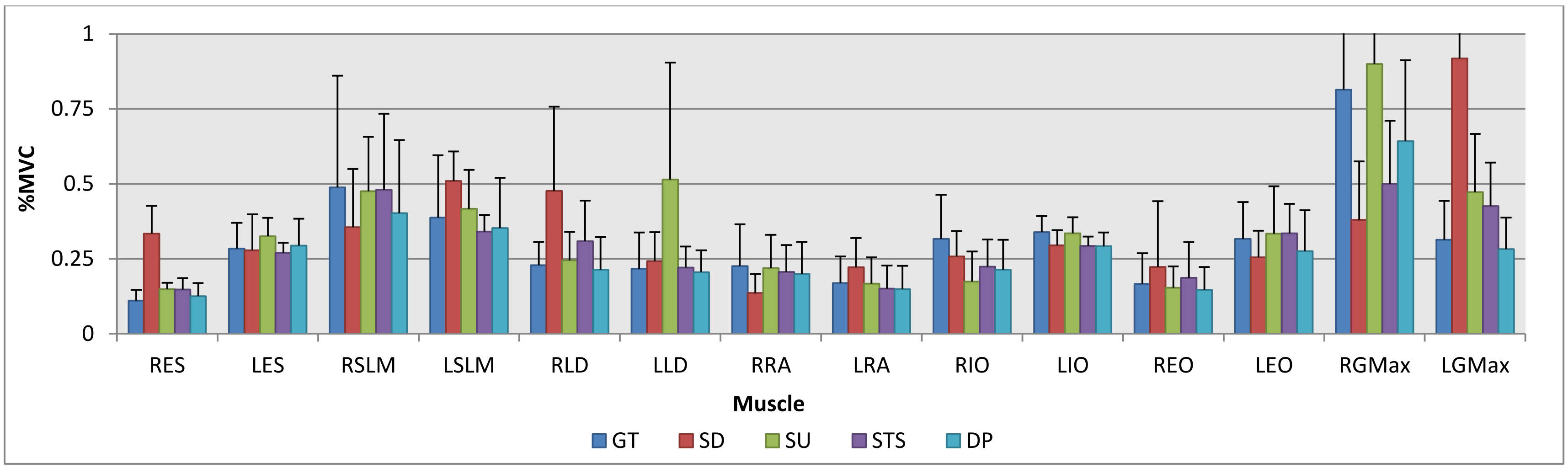

Figure 5.3: Comparison of average TFA group muscle activation during five activities of daily living.

Over all tasks, the general trends of rotational trunk stiffness and co-contraction were to present higher values for the TFA group than the control group. A comparison of the average rotational trunk stiffness between groups for each task is included for reference in Appendix C. This contains comparisons to the pair of older control participants to demonstrate that the change in stiffness is not necessarily due to the higher age of the participants. In both the control and TFA groups, bilateral co-contraction skewed higher than flexion-extension co-contraction. For trunk stiffness, the lateral bend was consistently higher than the flexion-extension or axial twist stiffness values. This was common to both groups. Averaged across tasks, TFA group CCI was roughly six times larger than the control group in both bilateral (C:11.51, A:68.13) and F/E (C:8.36, A:48.82) categories, while trunk stiffness was approximately three times larger in the F/E (C:126.24, A:389.06), LB (C:271.37, A:830.03), and AT (C:110.79, A:349.60) planes. As such, group figures will be presented separately to more clearly display between-task differences.

Bilateral co-contraction of the control group was greatest, significantly across tasks (df=4, F=45.86, $p<0.0001$), during the DP and STS tasks. This was followed by a lesser value during the SU task, which, in turn was significantly greater than the SD task. Flexion-extension co-contraction was significantly greatest (df=4, F=23.14, $p<0.0001$) during the DP task, while the STS task was also significantly greater than the SD task. The control group exhibited no significant differences (df=4, F=1.61, p=0.17) in peak rotational stiffness measurements across tasks around the flexion-extension axis (Figure 5.4). In the frontal plane, the SD and STS tasks displayed significantly greater (df=4, F=8.92, $p<0.0001$) peak stiffness values than the GT and SU tasks. The axial twist stiffness also presented significant differences (df=4, F=7.93, $p<0.0001$), as the STS task had greater peak values than the remaining tasks. This is displayed in Figure 5.4.

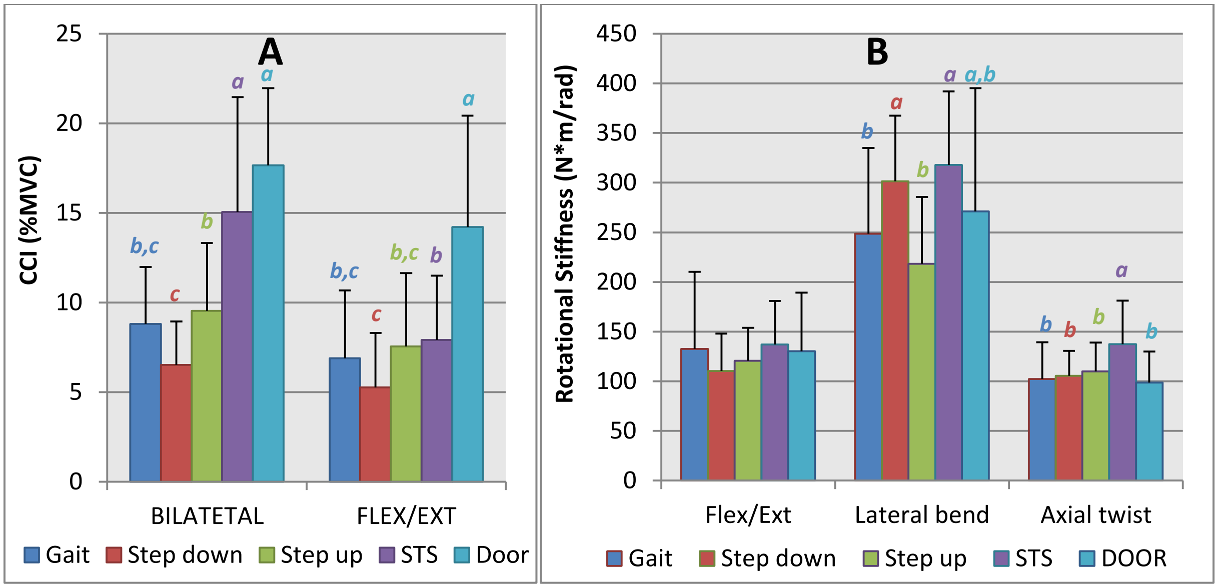

Figure 5.4: Comparison of total bilateral and flexion/extension co-contraction values (A) and peak rotational trunk stiffness (B) in 3-dimensions across activities of daily living for the control group. Similar lowercase letters denote non-significant differences.

Flexion-extension (df=4, F=0.21, p=0.93) and axial twist (df=4, F=1.36, p=0.26) peak rotational stiffness values were not found to be significantly different across tasks for the TFA group. The lateral bend peak stiffness also failed to show significant differences across tasks (df=4, F=2.02, p=0.10), though the STS task's value was notably higher, albeit with a relatively high standard deviation.

The TFA group also exhibited no statistically significant co-contraction differences across tasks for either the bilateral (df=4, F=1.87, p=0.14) or F/E (df=4, F=2.38, p=0.069) measurements. In Figure 5.5, the increased values of the STS co-contraction measurements are noted, though the large deviation in recorded values resulted in no significant difference. Similar to the control group, the DP had notably larger values, though they were, again, highly variable. As the co-contraction index is a sum of information collected from each time-series data point, the resultant CCI values increase the longer the task takes to perform. As such, these values appear to be a function of the increased time needed for some subjects to complete the DP relative to other tasks. When normalized to task length, the values are lower than the other tasks in both co-contraction categories. Taking this into account, and acknowledging that even the unnormalized CCI value is not especially different between tasks, the DP task's inflated co-contraction was not considered to have any mechanical significance.

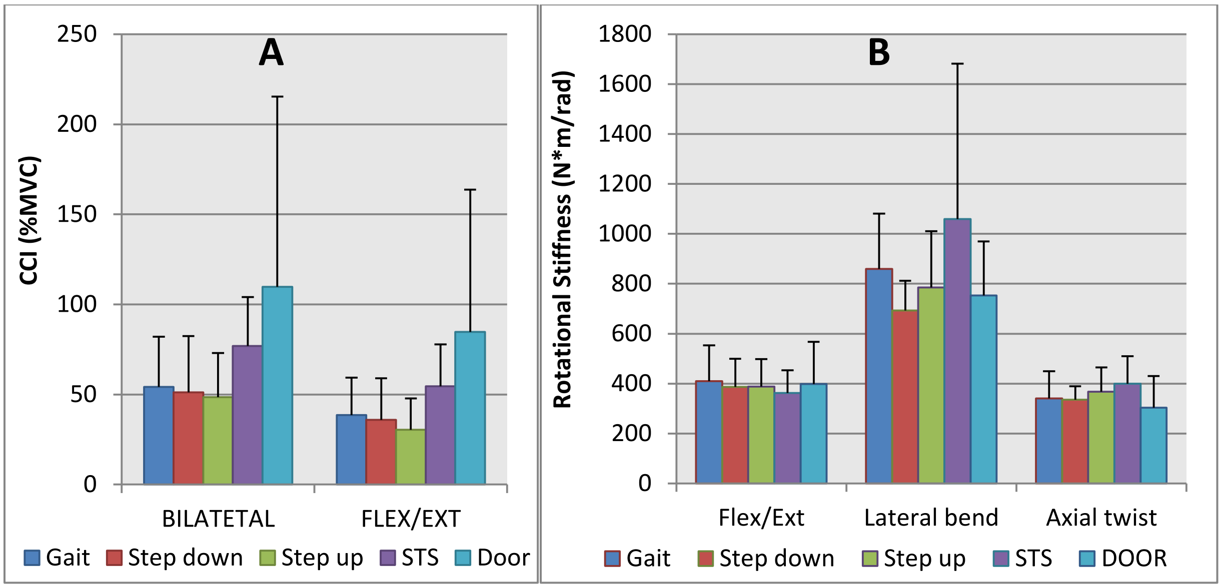

Figure 5.5: Comparison of total bilateral and flexion/extension co-contraction values (A) and peak rotational trunk stiffness (B) in 3-dimensions across five activities of daily living for the transfemoral amputee group. No significant between-task differences were found.

When the rotational stiffness percent contribution of each muscle is taken into account, the control and TFA groups display similar overall patterns with regards to muscle groups. Flexion-extension trunk stiffness (Figure 5.6) was predominantly a factor of ES, SLM, and IO activity, with slightly lesser contributions from the RA and EO, and little from the LD or QL. For lateral bend stiffness (Figure 5.7), the IO, EO, QL, and ES were the major contributors. Axial twist stiffness (Figure 5.8) was mainly produced by the ES, IO, and EO, with small contributions from SLM and QL. Across-task comparisons of trunk stiffness percent contribution for the two subject groups, however, revealed differences.

Flexion-extension trunk stiffness for both subject groups was mostly created by the main back extensors and abdominals, with little contribution from the LD or QL muscle groups. Again, because of the volume of comparisons, only notable outliers and differences between tasks will be mentioned in the text. In terms of significant percent contribution differences ($p<0.05$), the RES was greatest during STS and DP for the control group, and lowest during SD for both groups. LES during GT was lower than all other tasks for the control group and higher during SD than all other tasks for the TFA group. RSLM contribution was similar across tasks for both subject groups, with SD profiting least. SD was the biggest beneficiary of LSLM activity for TFAs, while control participants showed no significant difference across tasks. For the TFA group, the RRA and LRA each contributed significantly more during STS than all other tasks, with the exception of SD for LRA. With controls, DP was least affected by LRA activity. RIO contributed less to the STS in controls, and less to SU for TFAs. No significant difference was found between tasks for LIO contribution in either group. For the control group, GT and SD received the largest percent contribution from REO and LEO compared to the remaining tasks, while TFAs were most affected by the EOs during the STS task. LLD was also a significantly greater ($p<0.0001$) contributor to sagittal plane stiffness during the DP than all other tasks in the control group, but overall did not make much of an effect.

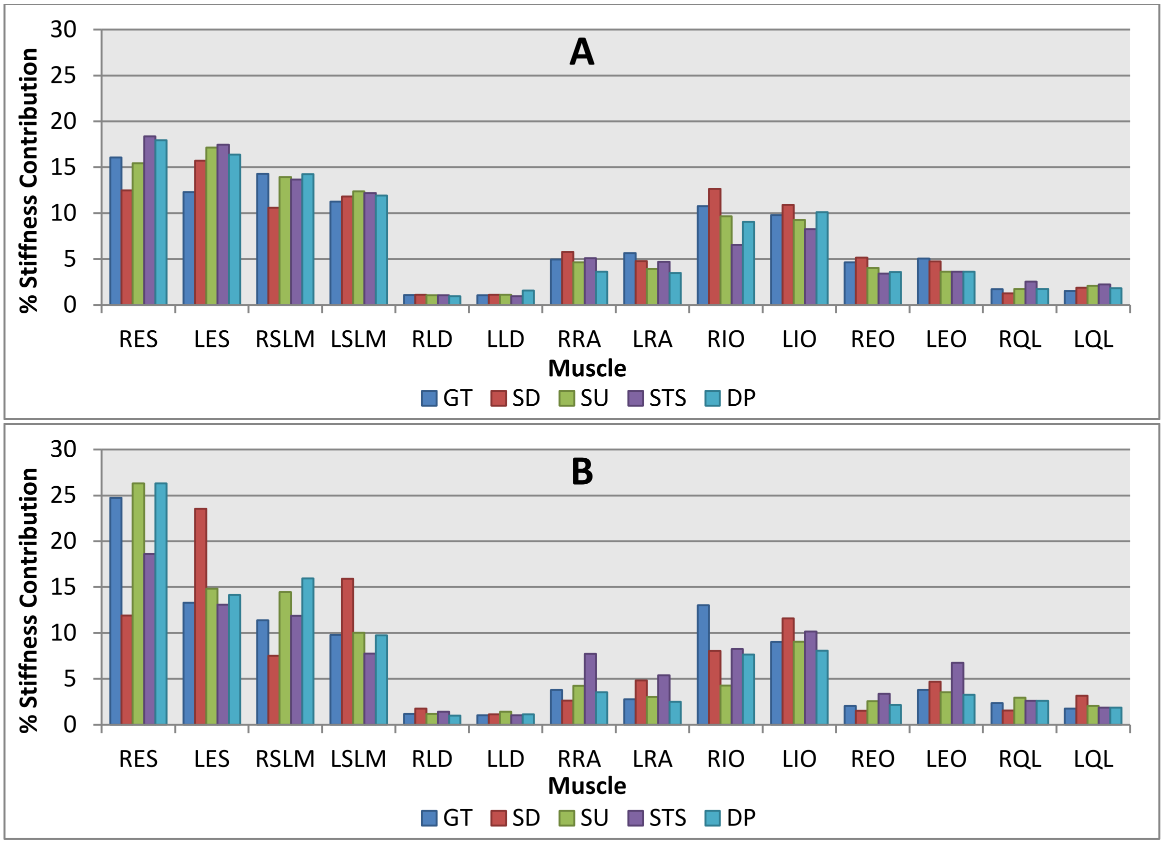

Figure 5.6: Comparison of individual muscles' percent contribution to flexion/extension rotational trunk stiffness for the control group (A) and TFA group (B).

In the frontal plane, lateral bend rotational trunk stiffness for both subject groups was mainly a function of ES and QL in the back, and the oblique muscles in the front. For both RES and LES, the control group showed significantly greater ($p<0.05$) contribution during the STS task above all others. The TFA group displayed the lowest RES but the highest LES input during the SD task. All oblique muscles in the control group were found to contribute significantly less to STS than all other tasks, with the exception of LEO during SU and DP. For TFAs, only GT and SD benefitted more than STS, from RIO and LIO, respectively. RQL contribution was greatest during STS and lowest during SD for the control group, while SD was less affected than all tasks but GT in the TFA group. Control LQL contribution was greatest during STS, but was greatest during SD for TFAs. Generally, the two subject groups could be differentiated in the frontal plane through muscle contribution to the STS and SD tasks. The control group back muscles contributed a relatively greater amount to STS, with lesser relative contribution from the obliques. By comparison, the TFA group displayed relatively average contribution from all muscle groups during STS compared to the other tasks. The TFA group also revealed greater asymmetry in stiffness contribution during the SD task compared to the control group.

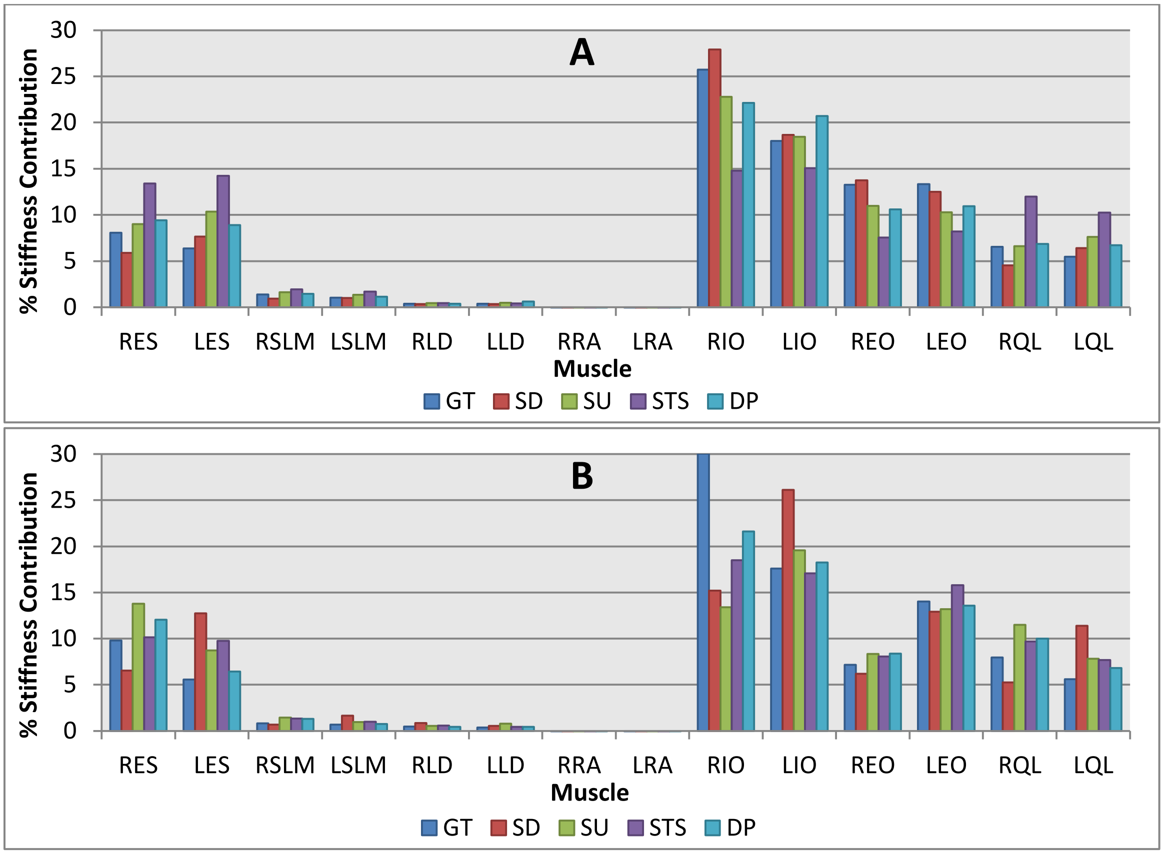

Figure 5.7: Comparison of individual muscles' percent contribution to lateral bend rotational trunk stiffness for the control group (A) and TFA group (B).

Axial twist trunk stiffness for both subject groups was mostly produced by the ES and oblique muscles. The significant defferences between tasks for these muscles' contributions are almost identical to those described in reference to lateral bend stiffness. In addition to the TFA group's bilateral asymmetry – most notably during the SD task – there are again differences between the two subject groups with regards to the STS task. The control group ES muscles again contribute more during STS than the other tasks, while the obliques contribute less to STS than any other tasks. These differences are not present in the TFA group, as the STS task displays similar muscle contributions to most other tasks.

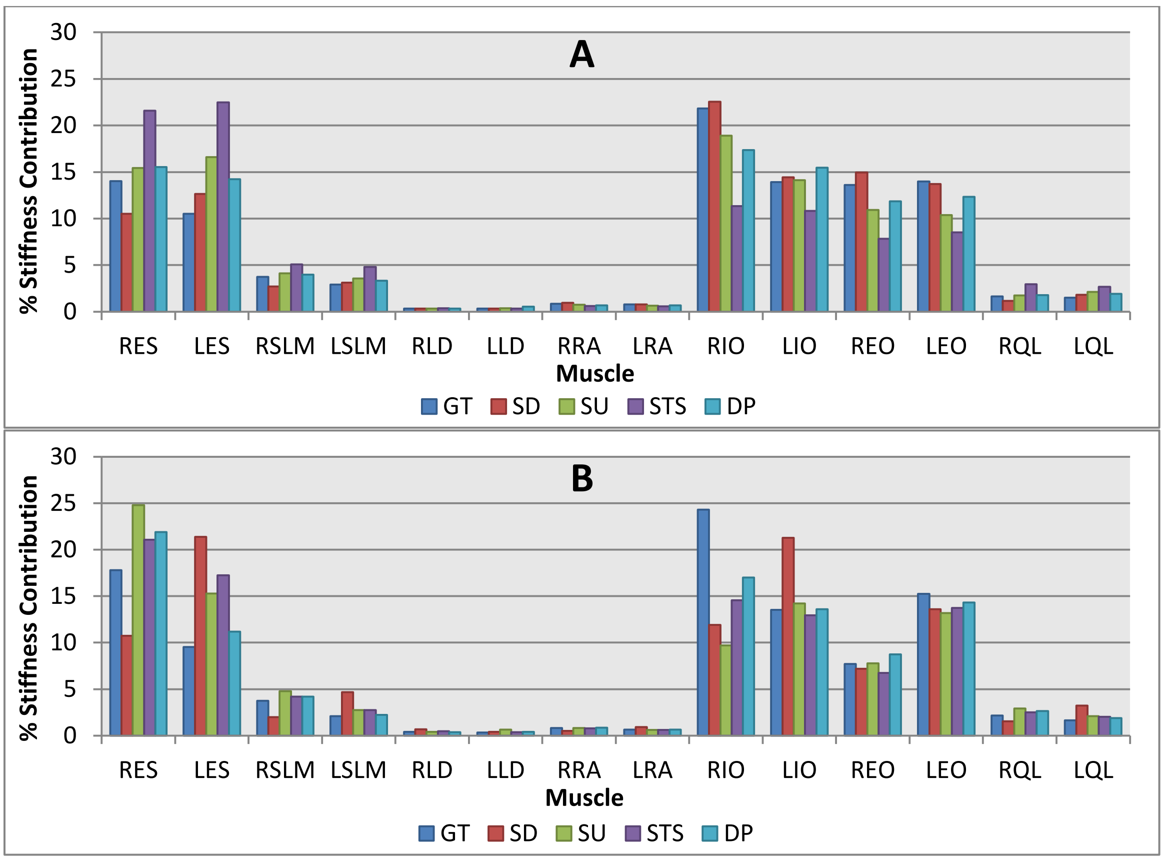

Figure 5.8: Comparison of individual muscles' percent contribution to axial twist rotational trunk stiffness for the control group (A) and TFA group (B).

5.2.1 – Case studies

As the respective primary passive and active outcome measures, 3D trunk ROM and 3D trunk stiffness were also compared on a case study basis, separating the 4 TFA participants. This separation is displayed in Figures 5.9. General similarities include relatively high F/E ROM during STS and AT ROM during DP. The relative increase in F/E ROM compared to the control group appears to be consistent for all TFA participants, as does the relative increase of LB ROM and decrease of AT ROM during STS and GT, respectively. The stiffness measurements of the TFAs also range from approximately 2-4 times those of the control group, with TFA1 being the lowest and TFA2 being the highest. While TFA1 and TFA2 most accurately mirror the control groups stiffness in the frontal plane, TFA1 is the only one maintaining a relatively high LB stiffness during SD; though this increase is reflected within the other two planes, unlike the control group. Examining the relationship between ROM and rotational stiffness reveals further differences.

For the control group, the F/E stiffness remains quite even despite relative highs and lows during STS and SD, respectively. LB ROM and stiffness appear to be inverted, with relative high stiffness matching low ROM in SD and STS. Relative AT stiffness is only increased during STS, while GT and DP tasks show greatly increased ROM not reflected in stiffness differences.

TFA1, unlike the control group, displays increased relative F/E and AT stiffness during SD, though this is only accompanied by a relative ROM increase in the sagittal plane. While the LB stiffness comparisons do mimic those of the control group, it no longer appears as an inverse of the LB ROM, with relatively higher STS and lower DP displayed by TFA1. AT ROM also displays lower relative values during both GT and DP than the control, but no reflective change is found in AT stiffness.

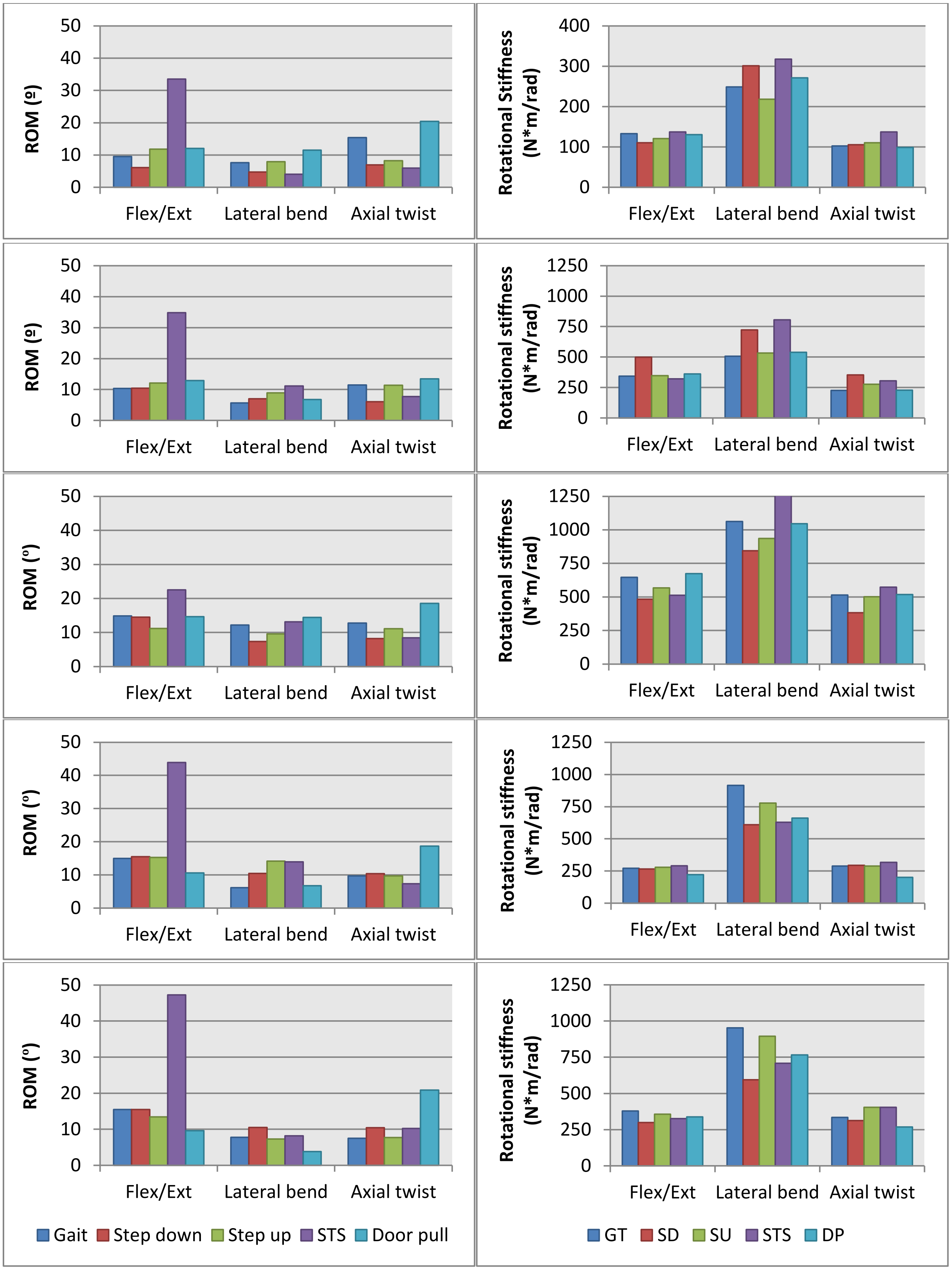

Figure 5.9: Comparisons of 3-dimensional trunk ROM and stiffness across all tasks for the control group (A), TFA1 (B), TFA2 (C), TFA3 (D), and TFA4 (E).

As with all other participants, TFA2 exhibited relatively high F/E ROM during STS, though the difference between tasks was much reduced due to TFA2's performance of the task from an elevated seat. However, this diminished difference in ROM did not appear to have any correlated change in F/E trunk stiffness. Rotational stiffness measures for TFA2 exhibit relatively low values during SD in all three planes. This remains despite no apparent differences in relative ROM when compared to TFA1. As with the other TFA participants compared to control, LB ROM is relatively higher during STS, and TFA2's LB stiffness value is also high during STS, similar to TFA1 and control, but differing from TFA3 and TFA4. AT ROM is again highest during DP for TFA2, though, like the control group and TFA1, AT stiffness remains relatively even with the other tasks.

Unlike the previous three comparisons, TFA3 displayed relatively low LB stiffness during SD and STS tasks. This was accompanied by relative increases in LB ROM during those same tasks compared to the control group. TFA3 also displayed a relative increase in AT ROM during SD compared to control, TFA1, and TFA2, though TFA3's AT stiffness during SD remained relatively average, equivalent to its position in the control group data. ROM during DP was also relatively low in the sagittal and frontal plane, and high in the transverse plane. Though this latter difference was common to all participants, TFA3's alternating extremes were paired with relative lows in both F/E and AT stiffness during DP. Among the other participants, this was only reflected in the transverse plane of TFA4.

Similar to TFA3, TFA4 demonstrated relative F/E and LB ROM lows and an AT ROM high during DP. In this case, however, the only change in stiffness during DP was a relative low in the transverse plane. The LB stiffness comparisons of TFA4 are similar to those of TFA3, representative of the inverse of those of the control group. The LB ROM of both SU and STS are much more in line with the remaining tasks than they had been for TFA3, but again appear to represent the inverse of TFA4's LB stiffness levels.

5.3 – Summary of each task

5.3.1 – Gait

Compared to the control group, the TFAs' prosthetic hip joint displayed an increased ROM value. This was by virtue of increased hip flexion (C: 24.34º, A: 36.20º), as the hip extension was slightly less in the TFA group (C: -12.49º, A: -9.93º). The amputee trunk ROM was greater in the sagittal plane and significantly lesser in the transverse plane in comparison to the control group. When compared statistically between subject groups, no significant difference was found in frontal plane trunk ROM during gait, though an examination of the extreme values of said range (C: -3.89º–3.75º; A: -1.54º–6.64º) resulted in the TFA group displaying a consistent lateral bend ipsilateral to the prosthesis (Figure 5.10).

Further analysis of individual amputee results as case studies shows the difference in lateral trunk movement between those with canes and without. Those with canes on the prosthetic side (TFA1 and TFA2) displayed a maximum lateral bend of 3.49º (0.34) toward that side, similar to that of the control group. TFA3 and TFA4, with no canes, displayed lateral ROMs entirely on the side ipsilateral to the prosthesis, from 2.36º (0.89) to 9.05º (0.80).

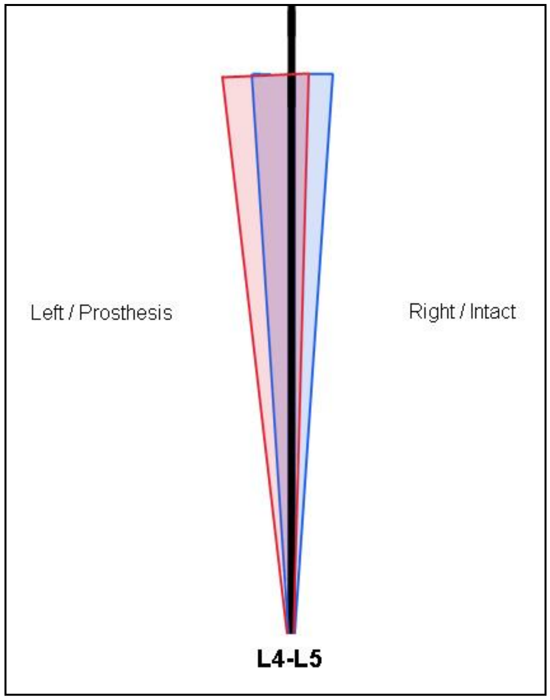

Figure 5.10: Lateral trunk ROM (º) during gait relative to a neutral upright posture. Blue – Control group; Red – TFA group.

In comparison to the control group, the TFA group displayed higher average peak muscle activation for all muscles (Appendix B). No statistical analyses between right and left muscles were performed, though a few differences are noted. The clearest difference is between the RES and LES (Figure 5.11), whose values are similar in the control group but quite different in the TFA group, with the prosthetic side (LES) displaying greater than double the peak activation value. Both the control and TFA groups display a double peak pattern for the ES muscles, coinciding with TO and IC events. A second bilateral difference is observed in the RLD and LLD (Figure 5.12), though this can be attributed to the use of canes. With a cane on its side, the respective peak activation values of RLD and LLD are 64.69 (4.78) %MVC and 69.41 (8.03) %MVC. Without an ipsilateral cane, TFA values are reduced to 27.86 (1.68) %MVC and 29.50 (4.18) %MVC, remaining greater than the control group values of 11.54 (8.53) %MVC and 14.32 (9.10) %MVC.

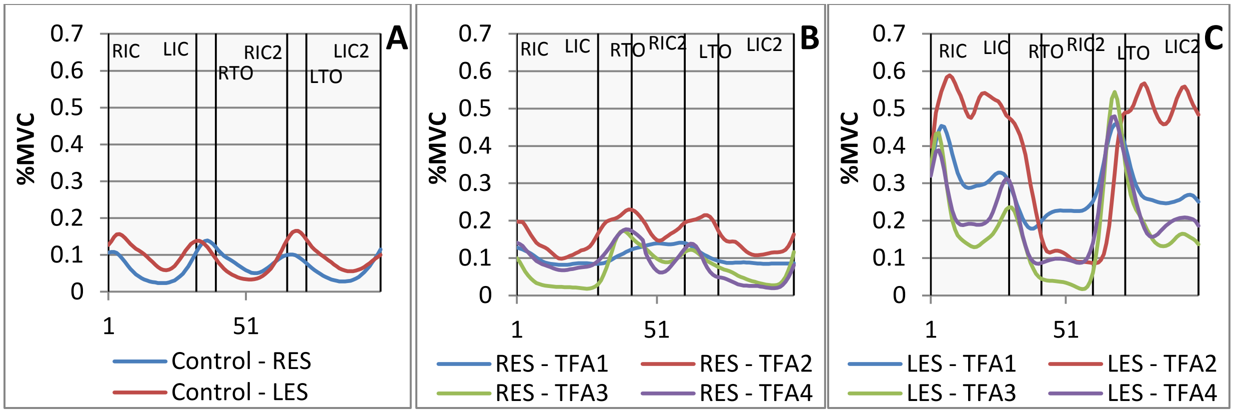

Figure 5.11: RES and LES activation during gait task for both subject groups. A – Control, B – Intact side TFA, C – Prosthetic side TFA.

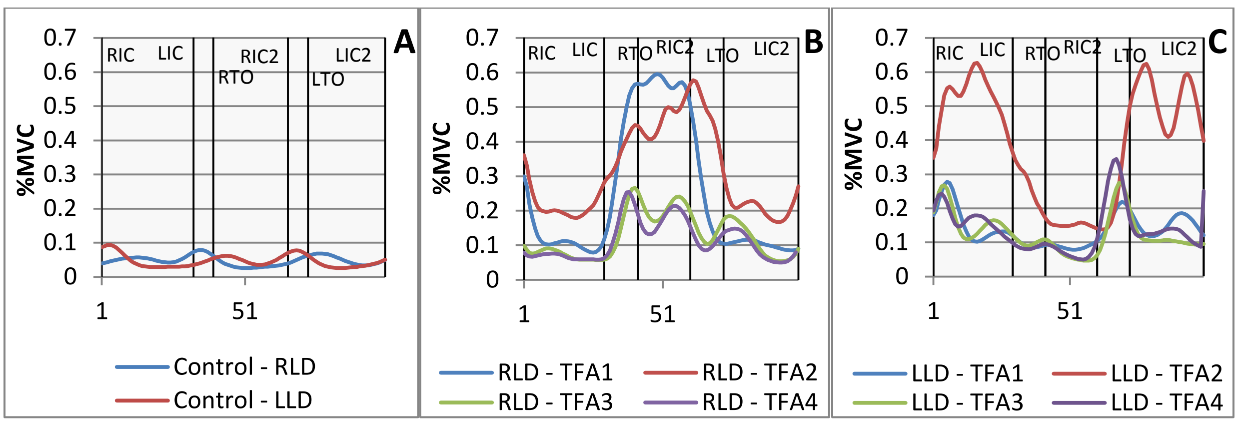

Figure 5.12: Anticipatory RLD and LLD activation during the gait task, displaying TFA1 and TFA2's canes' contribution to stability during stance phase. A – Control, B – Intact side TFA, C – Prosthetic side TFA.

During the GT task, the TFA group exhibited larger peak rotational stiffness values than the control group in all three planes, as well as larger co-contraction measurements in both categories. Investigation of frontal plane trunk stiffness reveals bilateral differences. During the single support (SS) phases of gait, the control group displays similar average lateral bend stiffness values for both right SS and left SS. For the TFA group, average trunk stiffness is greater during prosthetic SS compared to the intact leg SS. Separating the individual amputees, this trend remained consistent (Figure 5.13). Further differences can be seen when examining muscle contribution to trunk stiffness.

The lateral trunk muscles (LTM) – including the IO, EO, and QL – are major contributors to frontal plane trunk stiffness. Dividing this grouping of muscles into right and left sides, the bilateral differences during SS phases are reinforced. The stiffness contributions of the right lateral trunk musculature (RLTM) and left lateral trunk musculature (LLTM) during right/intact SS phase were 40 (3)% and 44 (7)%, respectively, for the control group, and 42 (6)% and 41 (4)% for the TFA group. During left/prosthetic SS phase, the control group had a RLTM contribution of 46 (6)% and a LLTM contribution of 38 (6)%, compared to 52 (7)% and 30 (4)% in the TFA group.

The similar contribution values during right/intact SS were common to all amputees, which was not the case during left/prosthetic SS. Through prosthetic SS, TFA1 displayed RLTM and LLTM contribution values similar to the control group, at 43(0.3)% and 37 (0.4)%. TFA2, TFA3, and TFA4 had RLTM and LTLM values of 45 (0.4)% and 32 (0.2)%, 61 (2)% and 29 (2)%, and 62 (2)% and 18 (1)%, respectively. Generally, TFA participants without canes relied more heavily on the lateral trunk muscles on the contralateral side to produce lateral bend trunk stiffness during prosthetic SS phase.

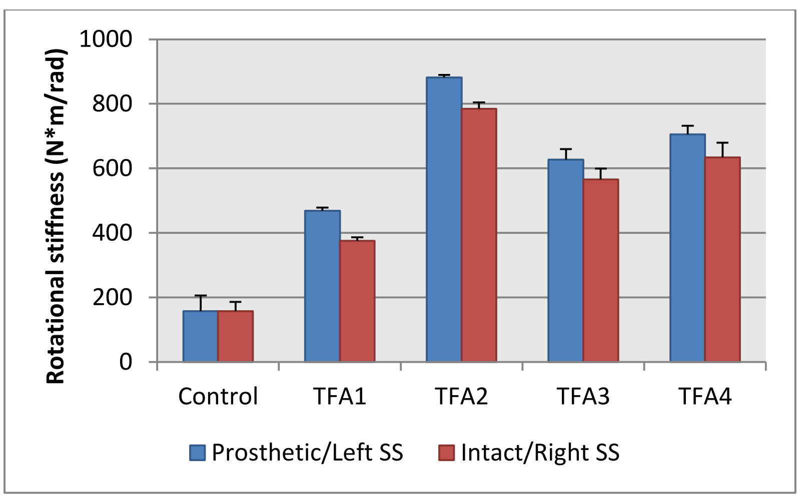

Figure 5.13: Comparison of average lateral bend trunk stiffness during single-support gait phases of the right and left legs across subject groups.

Examining the 3-dimentional trunk stiffening patterns of the control and TFA groups throughout the GT task cycle (Figures 5.14-5.15), there is a clear waveform pattern with peaks corresponding to the IC events of the right and left feet. Though the overall values are different, the patterns of the F/E and AT stiffness are relatively similar between the control and TFA groups. In the frontal plane, however, the groups differ between the LIC and the second RIC. During the SS phase on the prosthetic limb, the lateral rotational stiffness does not decrease as consistently as it does in the control group. Instead, TFA1 and TFA2 maintain a relatively high lateral bend stiffness before decreasing during the SS phase of the intact leg. TFA3 and TFA4 each display an additional local stiffness peak during prosthetic SS, which is lesser or not present throughout the intact SS.

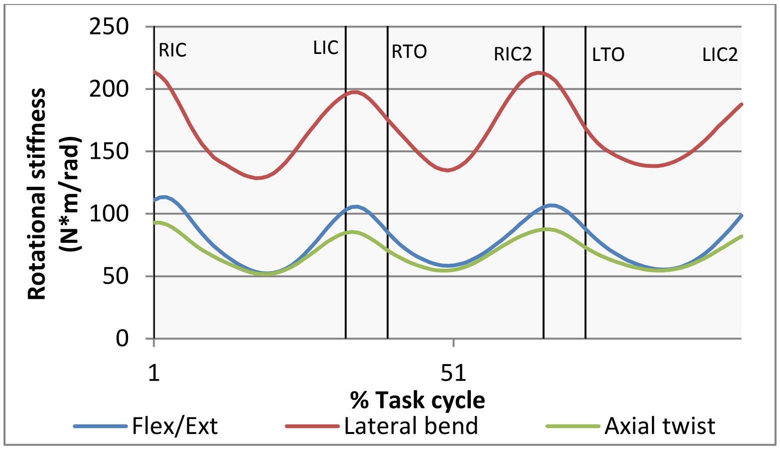

Figure 5.14: Comparison of the average 3-dimensional rotational stiffness patterns of the control group during the gait task.

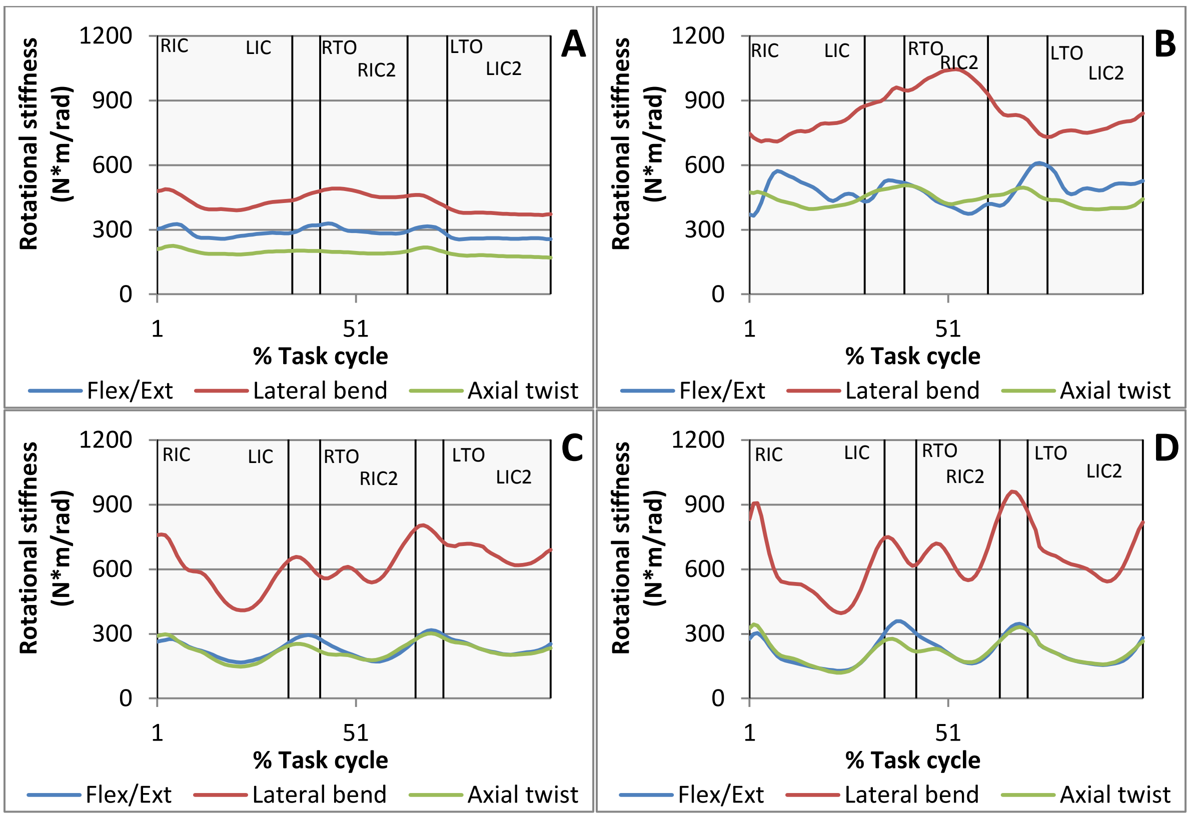

Figure 5.15: Comparison of the average 3-dimensional rotational stiffness patterns of TFA1 (A), TFA2 (B), TFA3 (C), and TFA4 (D) during the gait task.

5.3.2 – Step down

During SD, the lead limb knee ROM was found to be greater in the control group (38.99 [8.16]º) compared to the TFA group (23.67 [12.92]º). For lead limb hip ROM, the TFA group (32.71 [11.09]º) was greater than the control group (21.73 [7.02]º). These values were predominantly a result of initial knee and hip flexion just following LfTO but prior to LfIC. Further differences in lead limb knee and hip angles were seen following LfIC (Figure 5.16). In the control group, the leading knee goes through an additional period of flexion to absorb the body weight, while the hip continues a gradual extension. TFA1 displayed similar knee and hip angle patterns, albeit with lesser knee movement. This participant, however, was the only TFA to demonstrate a clear period of knee flexion following LfIC. TFA2, TFA3, and TFA4 each showed initial periods of knee flexion, but little to no knee motion following LfIC. While TFA2 continued hip extension after LfIC, TFA3 and TFA4 each go through an additional period of flexion when accepting BW.

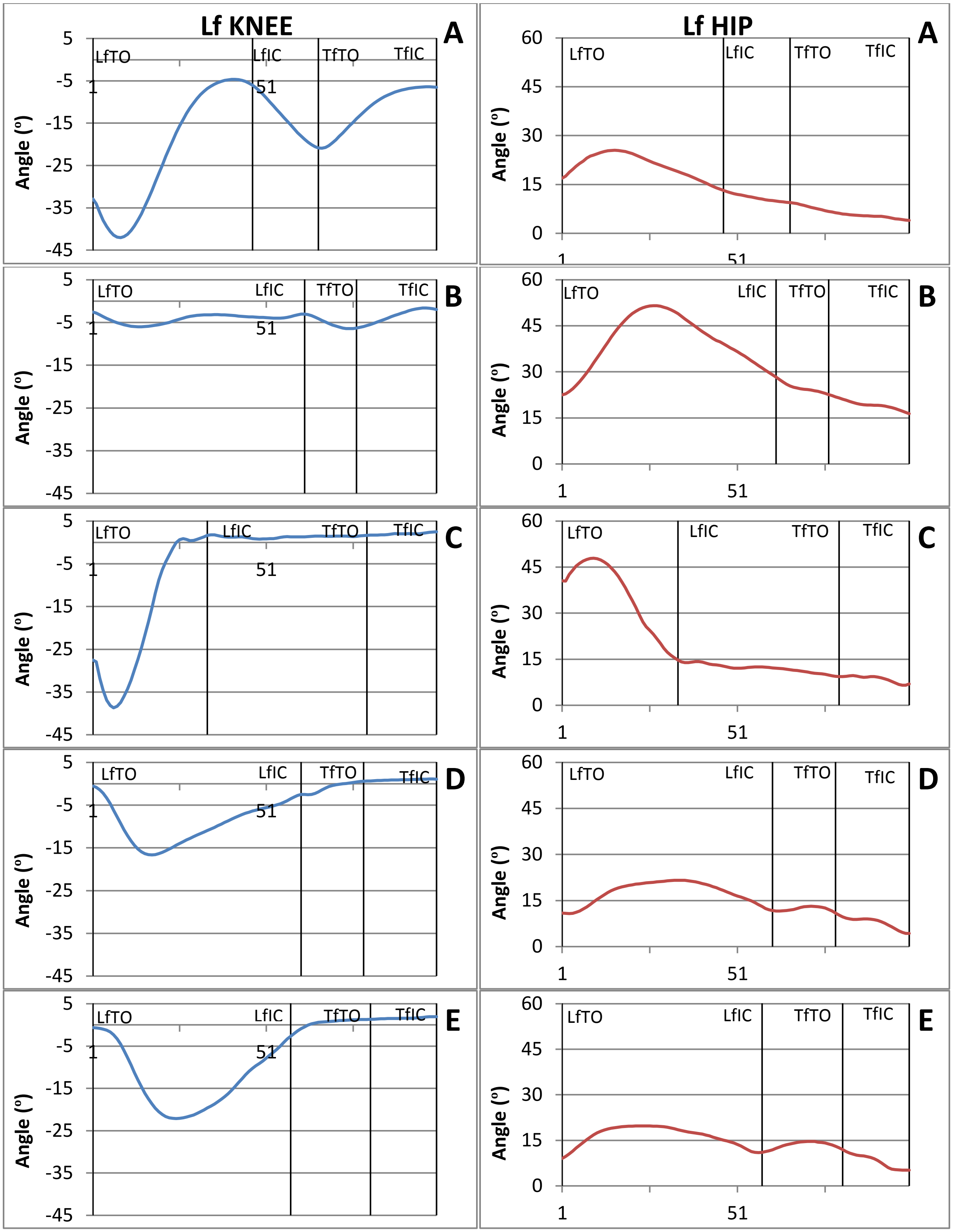

Figure 5.16: Lead limb knee (blue) and hip (red) angles during the step down task, for the control group (A), TFA1 (B), TFA2 (C), TFA3 (D), and TFA4 (E).

For the step down task, kinematic trunk data for TFA4 is absent due to technical trouble during recording. Comparing the controls and TFAs, trunk ROM was notably different between subject groups in the sagittal (C: 6.12º, A: 12.63º) and frontal (C: 4.75º, A: 7.7º) planes, while little difference was seen in the transverse plane (C: 6.93º, A: 7.51º). At the LfIC event, the body is absorbing its weight. TFA3 – the only remaining participant without a cane to support his descent – displayed a much higher sagittal trunk flexion angle than the control group and other amputees, while the lateral bend values are closer together, and the axial twist values are not significantly different (Figure 5.17). In the sagittal plane, the TFA trunk angles all reach a peak at or prior to LfIC, and begin extension following this. Conversely, the control group displays a period of further flexion following LfIC, beginning extension at TfTO (Figure 5.18).

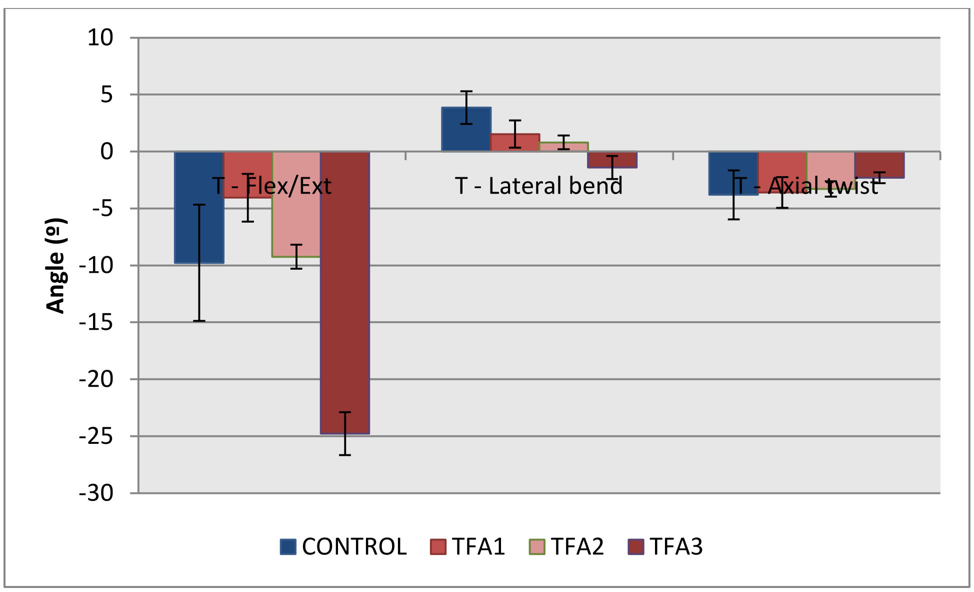

Figure 5.17: Comparison of 3-D spine angle at the LfIC event during the step down task (TFA4 trunk data missing).

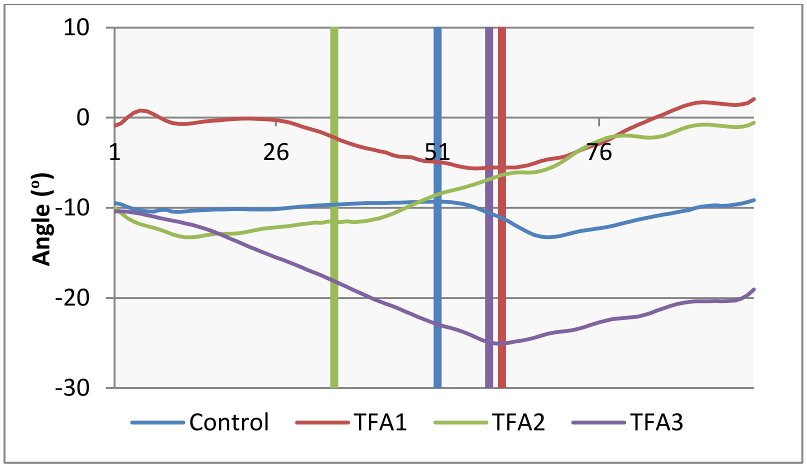

Figure 5.18: Comparison of the sagittal trunk motion across subject groups during the step down task, with the LfIC event marked for the control group and individual TFAs.

Figure 5.19 shows the ramped increase of stiffness in all planes from the control group, reaching a peak within 5% of the LfIC event. While they share a similar overall shape to the control group, the rotational stiffening patterns of the TFA group present two distinct patterns from the four participants. It should be noted that, because of the absence of kinematic trunk data for TFA4, rotational stiffness was calculated with the angle data from TFA3 and the EMG data from TFA4. Reasoning that it would be more accurate than constant zero trunk angles (comparison of the two methods included for reference in Appendix C), this solution was chosen based on the similar strategies observed between the participants in hip and knee kinematics. Graphs A and B of Figure 5.20 display a less distinct increase in stiffness, with TFA1 and TFA2 having begun the trial with a higher level of stiffness. Of note, these normalized outputs begin at the TO of the lead foot, and were preceded by the placement of canes on the lower surface. Another common element of these two patterns is the increased F/E stiffness relative to the AT. The two non-cane users (graphs C and D) are much more similar to that of the control in all three dimensions, though it is made distinct through a marked increase in stiffness late in the task cycle. All participants began to increase trunk stiffness just prior to the LfIC event.

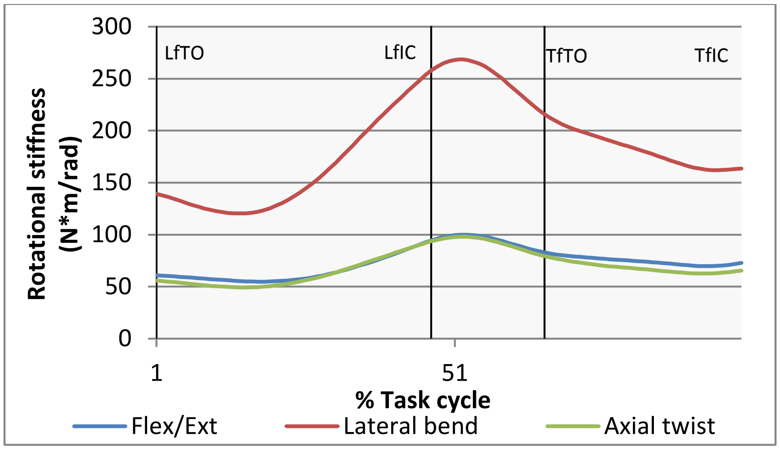

Figure 5.19: Comparison of the average 3-dimensional rotational stiffness patterns of the control group during the step down task.

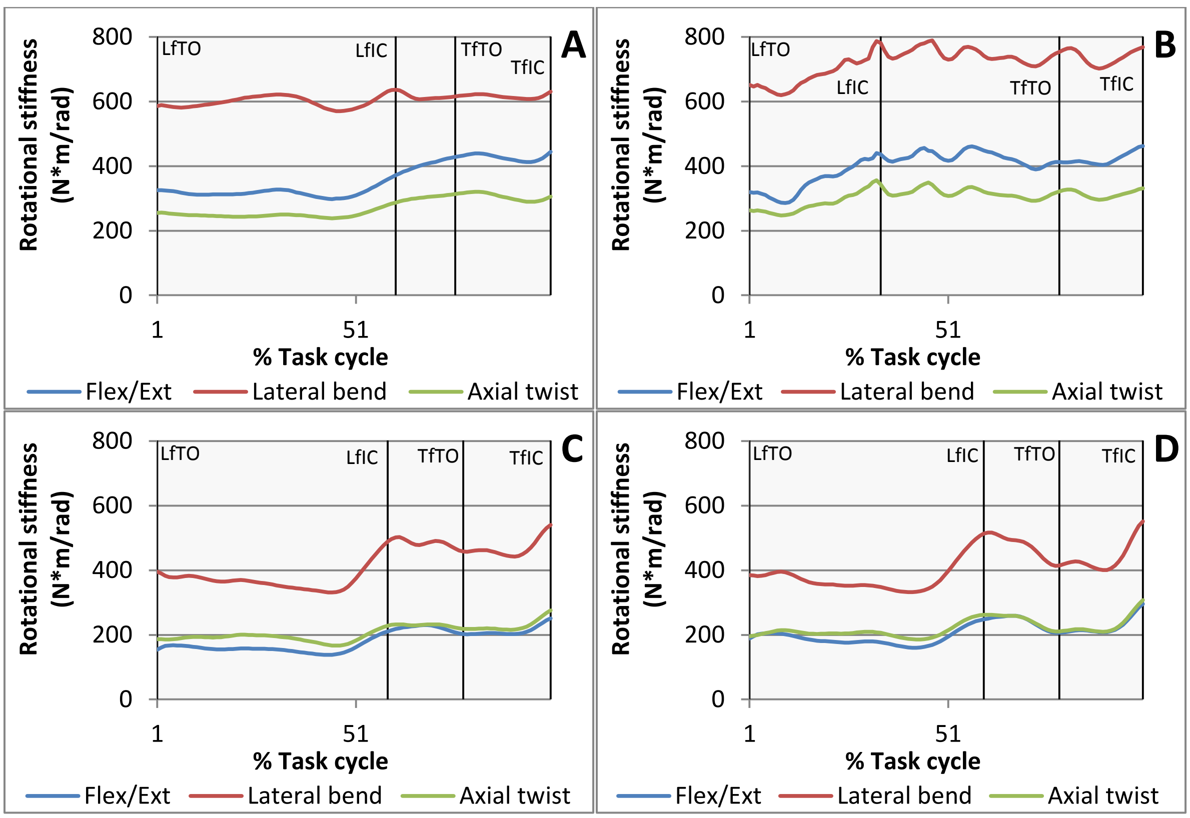

Figure 5.20: Comparison of the average 3-dimensional rotational stiffness patterns of TFA1 (A), TFA2 (B), and an average of TFA3 (C) and TFA4 (D) during the step down task.

With all amputee subjects ostensibly leading with the prosthesis, there is a greater contribution from the muscle on that side of the body to overall stiffness in the TFA group. Specifically in the sagittal plane, the RES, RSLM, and RQL each display greater percent values than their contralateral partners, while the LLD has a marginally higher contribution to the F/E stiffness. Dividing the muscles into abdominal and back groupings, the TFAs show greater asymmetrical muscle contribution than the controls for both muscle groups, with little variation seen between TFA participants (Figure 5.21).

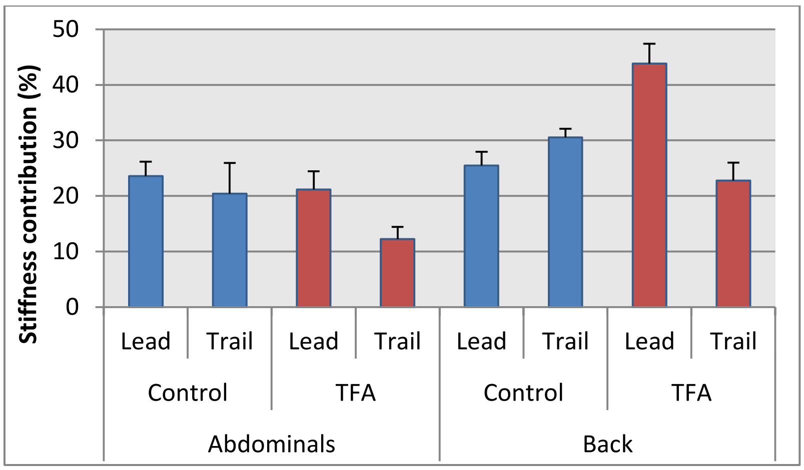

Figure 5.21: Percent contribution of the abdominal and back musculature to sagittal plane trunk stiffness during the step down task for the control and TFA groups.

5.3.3 – Step up

All TFA participants led with the intact leg, while the prosthesis was the trailing leg. Comparing the sagittal joint ROMs of the control and TFA group, only the LA, LK, RH, and lateral trunk bend are significantly different ($p<0.05$) between groups. The lateral bend of the trunk went to a maximum of 1.89º on the lead limb side for the control group and 6.96º for the TFA group, while the trail leg knee joint of the TFA group barely flexes (C: 39.87º, A: 5.58º). Though the LH ROM is not significantly different in the sagittal plane, a further examination of the trail leg movement shows differences in maximum hip extension values (C: -8.13º, A: 3.03º), as well as lateral and axial hip ROM (Figure 5.22). Lateral and axial hip ROM values are skewed by the large rotations exhibited by TFA1, though prosthetic lateral hip ROM is consistently greater than the control group for all TFAs.

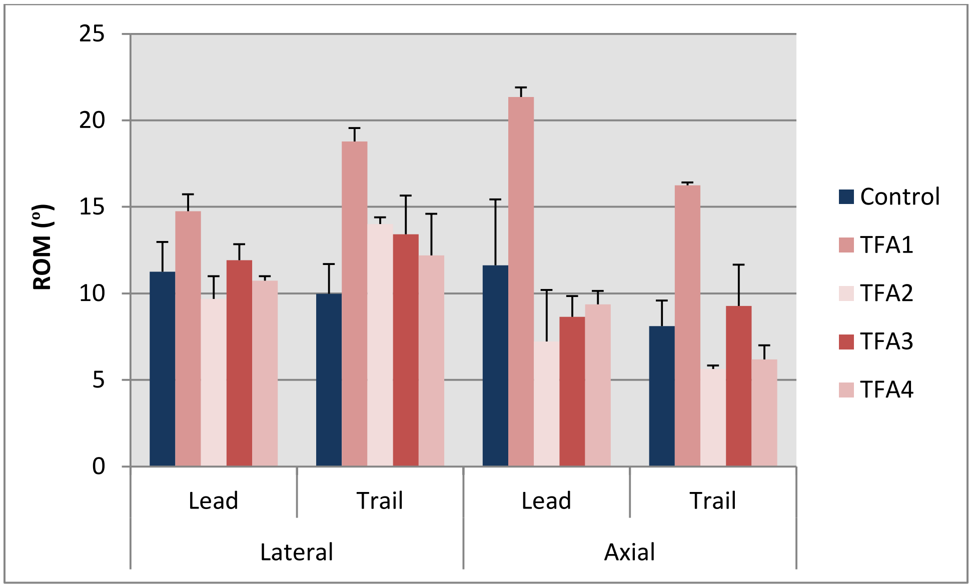

Figure 5.22: Lateral and axial hip ROM between subject groups for the step up task.

In Figure 5.23, the control group displays an initial local maximum of lateral stiffness right at the start of the task, as they prepare for a single support phase. This is reflected in Figure 5.24 for TFA3 (C) and TFA4 (D), and, to a degree, in TFA1 (A). It should again be noted that the cane support for TFA1 and TFA2 (B) occurred prior to the LfTO event at the beginning of this normalized task stiffness representation. This is followed in the control group by a double peaked stiffness pattern in all three planes. TFA1 is closest to this in the frontal plane, while TFA3 and TFA4 exhibit an increase in stiffness in 3-dimensions followed by a plateau and a further increase in the frontal plane. TFA2 was found to have maintained a reasonably consistent stiffness in every plane, but displayed a F/E stiffness relatively higher than AT. This was also the case with TFA1, as the control group and remaining amputee participants maintained similar sagittal and transverse stiffness. Generally, each TFA subject – with the notable exception of

TFA2 – reached a peak stiffness at the same point in the task cycle as their control counterparts, though the TFAs maintained lateral stiffness for a longer period of time.

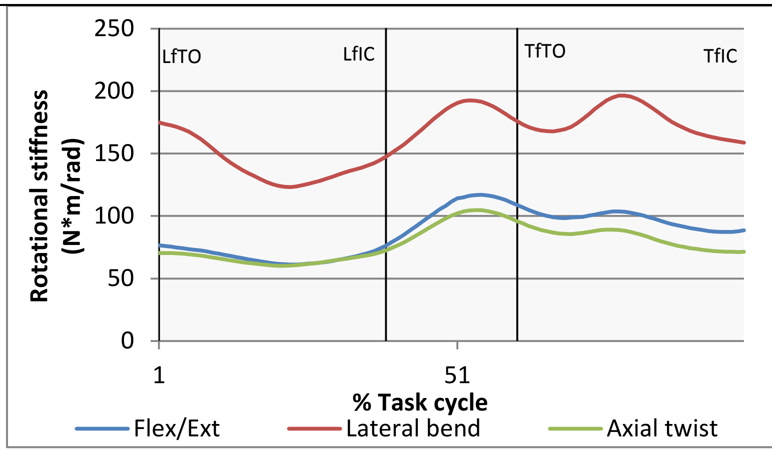

Figure 5.23: Comparison of the average 3-dimensional rotational stiffness patterns of the control group during the step up task.

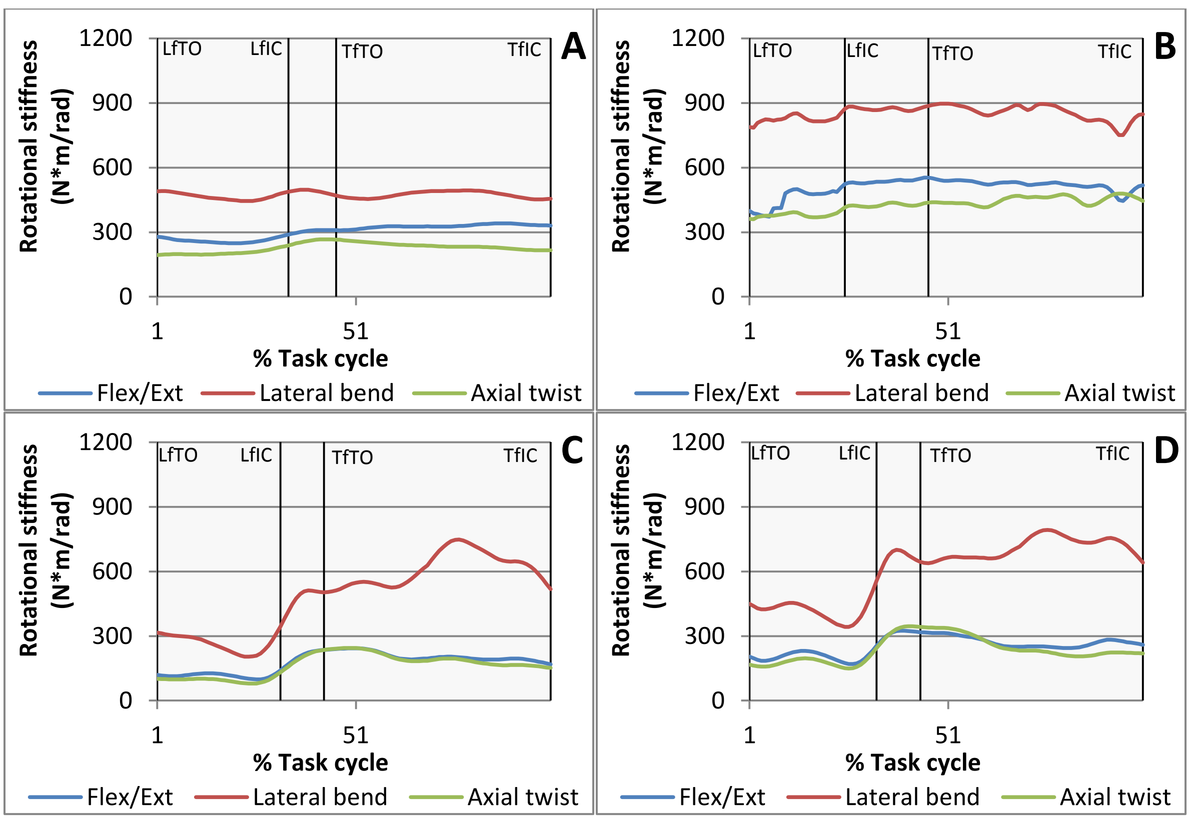

Figure 5.24: Comparison of the average 3-dimensional rotational stiffness patterns of TFA1 (A), TFA2 (B), TFA3 (C), and TFA4 (D) during the step up task.

5.3.4 – STS

Between subject groups, trunk ROM values were significantly different ($p<0.05$) in all but the sagittal plane, with the lateral bend ROM representing the largest difference. However, the maximum lateral bend of the trunk in the amputee group differs in direction. TFA1 exhibited a maximal lateral bend of 9.08º (0.612) to the prosthetic side (left), toward the cane that he had switched to his left hand for this task. This is compared to a maximal left lean of 3.63º (2.20) for the control group. TFA2, TFA3, and TFA4 each exhibited maximal lateral leans towards the intact side, of 6.11º (5.36), 11.32º (3.18), and 7.04º (6.24), respectively. Figure 5.25 shows the averaged time-normalized frontal plane trunk motion and, as such, does not exactly reflect the previously reported maximal leans, instead focusing on the direction and relative timing of the lateral bend.

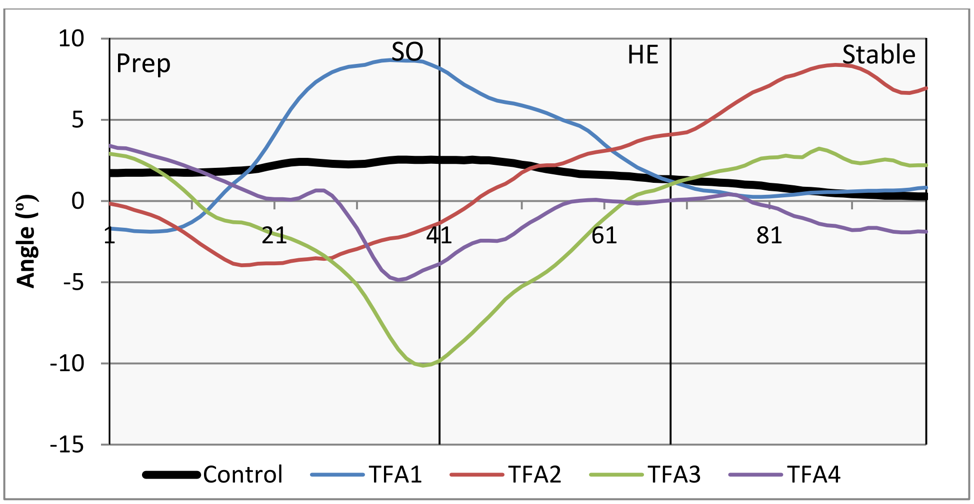

Figure 5.25: Lateral trunk motion during the sit-to-stand task for both subject groups.

Comparing average peak muscle activation between groups, the only muscle not notably different between groups was the RES. Bilaterally, TFAs showed notable differences between RES and LES activation, with the LES peak activation being much greater. This difference is most clearly seen in TFA3 and TFA4 (Figure 5.26). The peak activation of the RES and LES in both the control and amputee group corresponds approximately with the SO event, when knee and hip extension begin. By comparison, the peak activation of the RGMax and LGMax of the control group, and LGMax (prosthetic) of the TFA group occurred following SO. The RGMax of the amputee group displaying a peak activation at close to 85% of the task cycle, just after hip extension. Timing of other muscle group activations will be discussed in a later section.

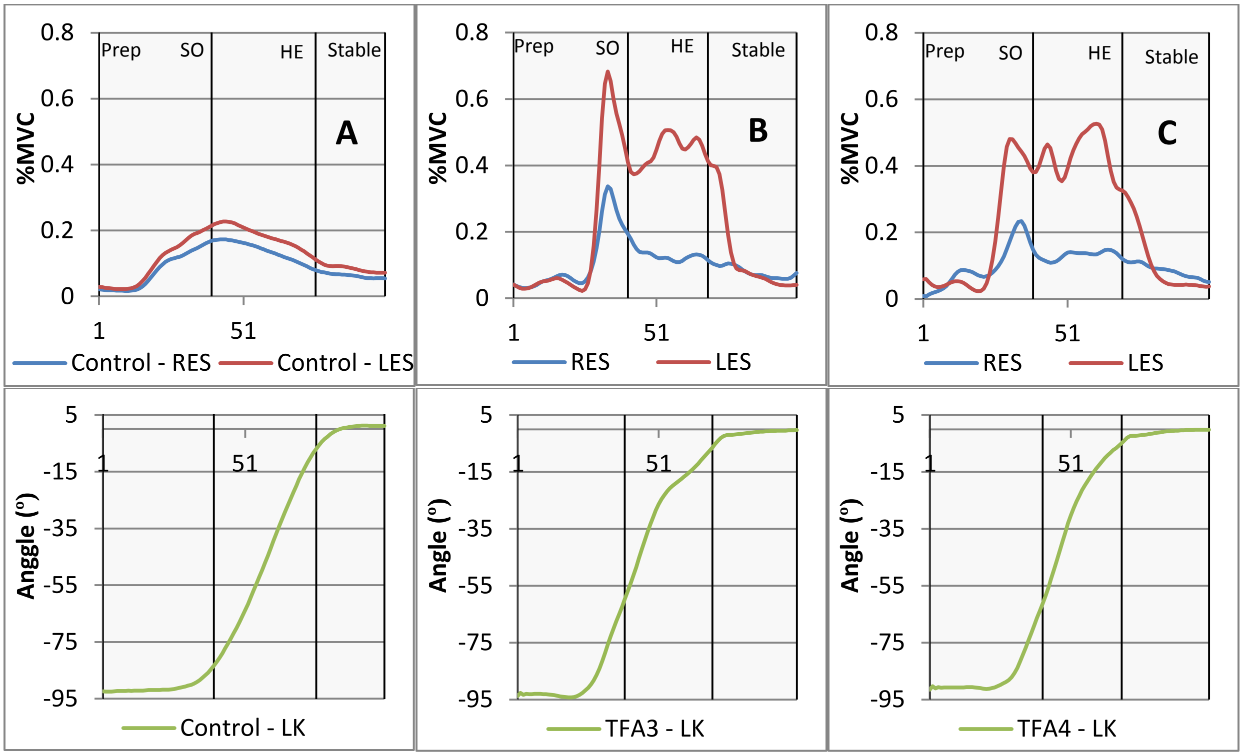

Figure 5.26: RES and LES activation during sit-to-stand task, displaying activation timing in relation to left/prosthetic knee extension. A – Control, B – TFA3, C – TFA4.

Trunk stiffness patterns exhibited by the control group were variable enough to be separated into three categories. On display in Figure 5.27, method A was demonstrated by 3 control participants, while method B and C were used by 5 and 2 participants, respectively. All control stiffening patterns display a marked increase in 3-dimensional trunk stiffness just prior to the seat off event. Method A reveals a sharp drop off in stiffness for the remainder of the task cycle, while method B exhibits a more gradual decline over the course of the task. Method C is distinguished by a sharp peak and partial decline, followed by a plateau until the hip extension event whereupon the stiffness decreases during the stabilizing phase.

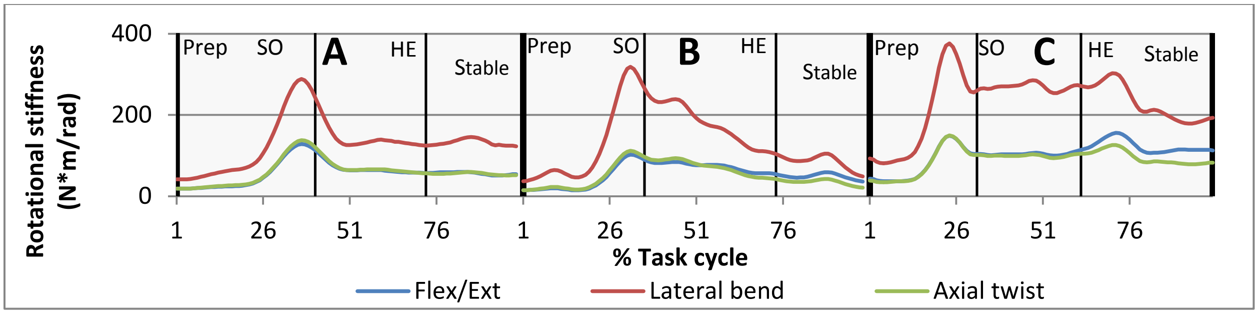

Figure 5.27: Control group variation in 3-dimensional rotational stiffness during the sit-to-stand task (three individual control group participant example trials).

TFA1, TFA3, and TFA4 each displayed a stiffening increase in all three planes, with an initial peak well prior to the seat off event, similar to the control group (Figure 5.28; A, C, D). This was succeeded by a second increase in trunk stiffness following the seat off event, which was to be maintained until full hip extension, at which point the participants experienced a gradual decline in 3-dimensional stiffness. This was most similar to method C in the control group, which was exhibited by the fewest participants. TFA2, performing the task from a more elevated seat, began the task with incredibly high lateral stiffness (graph B scaled down compared to other TFAs), only to decrease to the seat off event. Immediately following seat off, there was a distinct peak in F/E and LB stiffness, with notable oscillations during the stabilization phase.

Anecdotally, TFA subjects exhibited a ‘hitch’ at the beginning of the task, in which subjects performed what could be described as a rapid seated jump, followed by a slower steadying and extension movement. This was most apparent in TFA2 who already was seated at an elevated level, being unable to perform the task from the standard seat height. The ‘hitch’ was least notable in TFA3 and TFA4, who exhibited motions similar to the control group: a smoother beginning followed by a quicker extension.

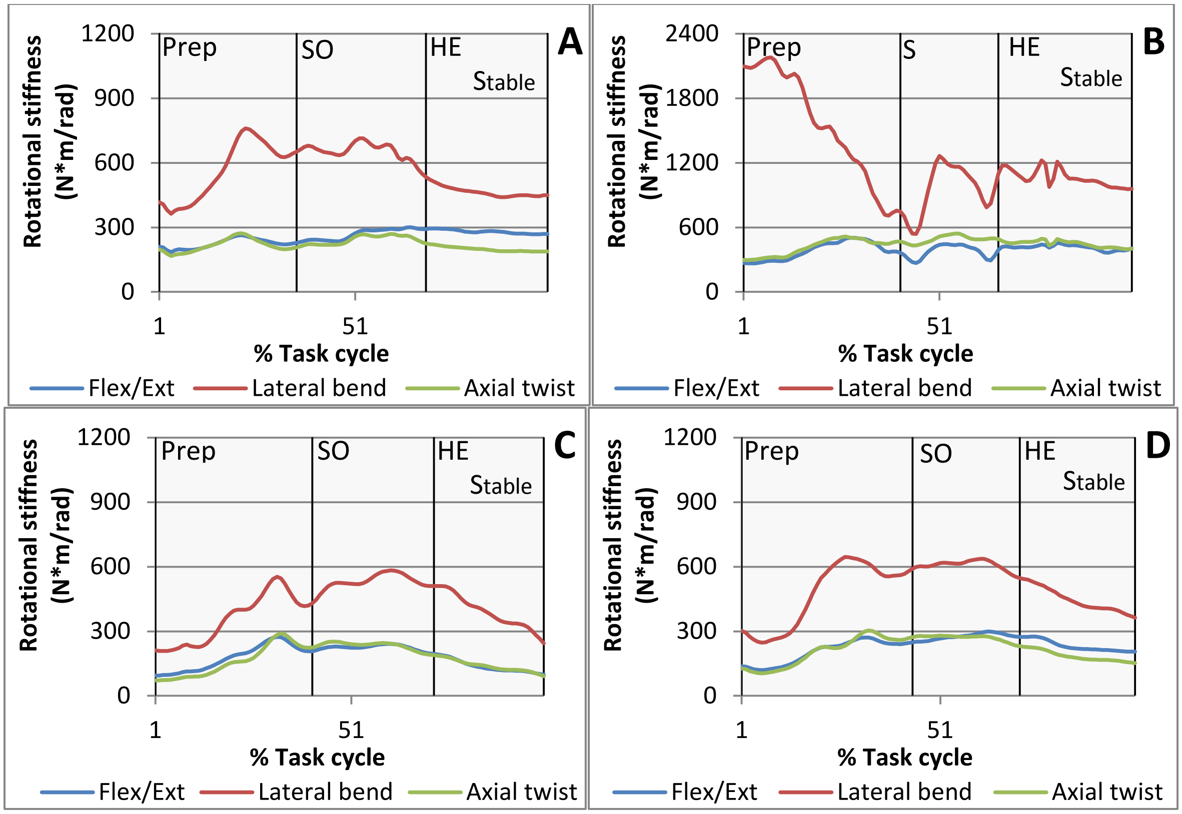

Figure 5.28: Comparison of the average 3-dimensional rotational stiffness patterns of TFA1 (A), TFA2 (B), TFA3(C), and TFA4 (D) during the sit-to-stand task.

5.3.5 – Door pull

The control and TFA groups performed the door pull task quite similarly, considering the inherent variability of the activity. The 3-dimensional trunk ROMs were not notably different between subject groups, and the extreme angle values of the trunk were also quite similar between groups in the frontal and transverse planes. The sagittal trunk angle ROM was biased towards flexion for the control group.

The control group displayed a stiffness pattern with a clear peak in all three planes between 70 and 75% (Figure 5.29). The TFA group (Figure 5.30) was found to have a similar stiffening pattern for F/E and AT, with the most notable peak occurring in the sagittal plane of TFA4. This peak, corresponding approximately to the door pull event, was earlier in the task cycle by percentage, but much closer to the control group in absolute time, as the amputees took longer to get started through the door frame after opening. The amputees also demonstrated relatively increased lateral bend stiffness prior to the door pull as each participant utilized multiple, smaller steps when approaching the door frame compared to the deliberate step forward of the control group.

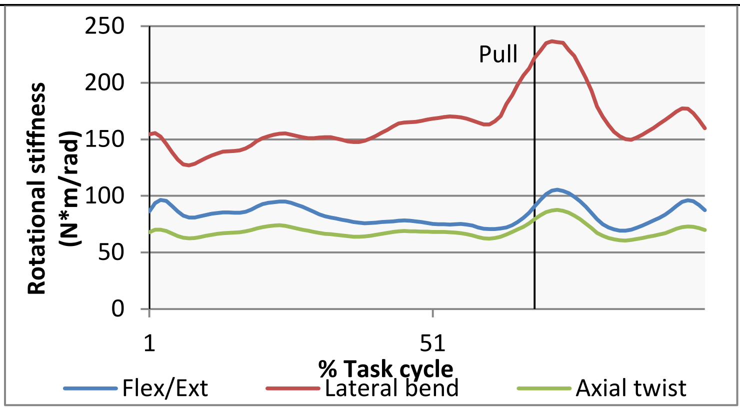

Figure 5.29: Comparison of the average 3-dimensional rotational stiffness patterns of the control group during the door pull task.

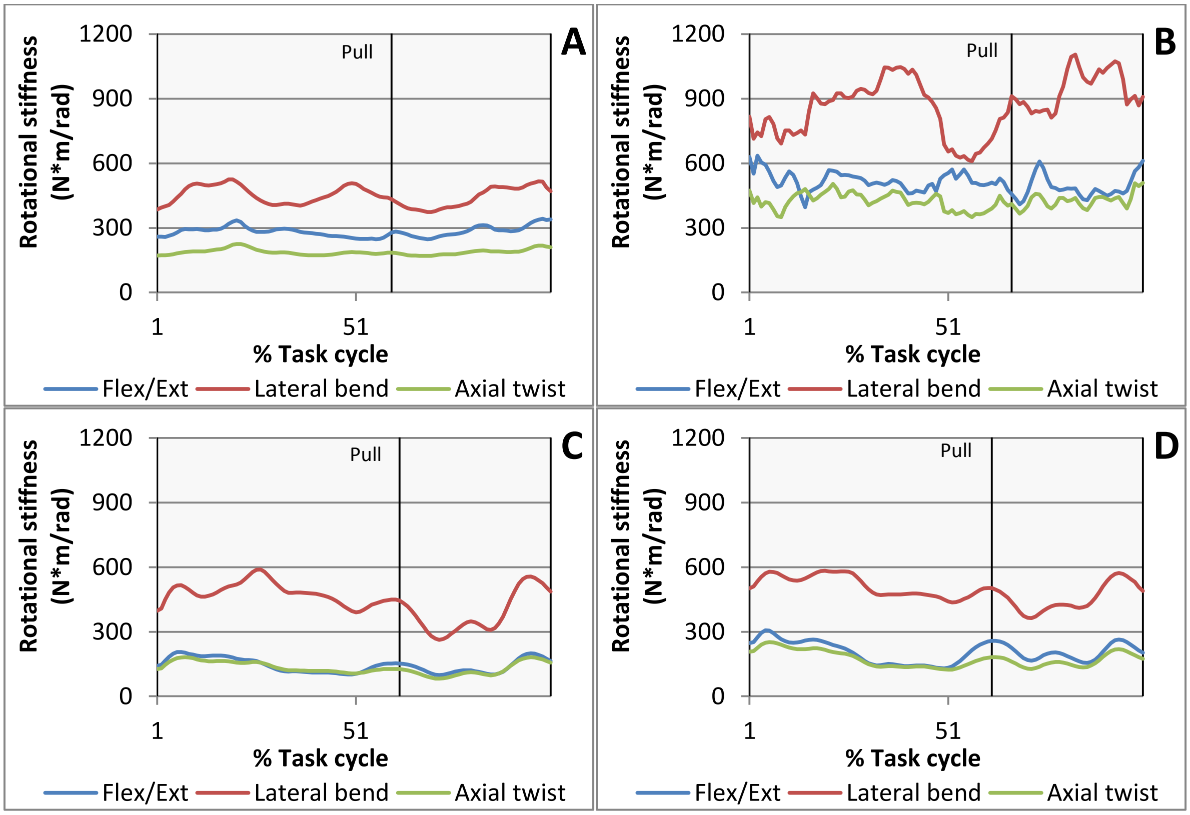

Figure 5.30: Comparison of the average 3-dimensional rotational stiffness patterns of TFA1 (A), TFA2 (B), TFA3(C), and TFA4 (D) during the door pull task.

CHAPTER 6 – Discussion

6.1 – Task comparison

The TFAs took longer to perform each task than the control group. For both groups, however, the shortest to longest tasks were SD, SU, GT, STS, and DP. When comparing the relative differences between tasks, the control group was relatively consistent, while the TFA group showed high variation that may have been masking significant differences. An example of this is activation of the LLD, which displayed peaks of 103%MVC during SU and 43%MVC during GT in the TFA group. Since two participants did not use canes, and one participant switched cane hands between these two tasks, the average peaks are highly variable and do not show significant difference, though they may exist with individual participants. Nevertheless, task comparisons still produced cogent information, as both groups and individual TFA participants were examined.

When muscles are activated, they produce their own inherent stiffness. For a given joint, the stiffness of all the muscles that cross it combine and act as 'guy wires' to stabilize the joint (McGill, 2014). This joint stiffness restricts the movement of the joint and helps reduce any excessive rotation. In the case of the spinal column, the trunk musculature that surrounds the lumbar vertebrae all help to stiffen the joint, guarding against instability and mitigating risk of injury through buckling of a joint. In terms of the active stiffness produced by these muscles, the values depend both on the level of activation and the geometric orientation of the muscle. This latter factor also determines to which plane the muscles will contribute stiffness.

Generally, rotational trunk stiffness was greatest in the frontal plane, with sagittal and transverse plane stiffness being about half the value of LB stiffness in both the control and TFA groups. F/E and AT stiffness values were approximately even, though F/E stiffness more often displayed slightly higher values with the notable exception of the STS task, which displayed higher AT stiffness for both subject groups. In the sagittal plane, neither group revealed differences in between-task stiffness values. LB stiffness was highest during SD and STS tasks in the control group, in accordance with predictions. For the TFA group, however, STS displayed the highest stiffness while SD displayed the lowest, yet neither difference was significant. The relatively low SD LB stiffness value in the TFA group may have been a result of the asymmetrical contributions of the trunk musculature, as the trailing limb side consistently displayed close to half the percent contribution of the leading, prosthetic limb side. Though SD was the least laterally-stiff task for the TFA group, the peak value was still much higher than that of the control group, as per the previously noted general differences in control and TFA stiffness values (see *7.4 – Limitations*). In the transverse plane, STS was again the task with the highest stiffness value in the control group. This was also the case in the TFA group, but the between-task comparisons were much closer, and the result was not significant.

Comparing every task individually to the remaining tasks helps give an indication as to the unique challenges each ADL may present to the TFA population. Beginning with GT, the control group demonstrated AT ROM that was significantly greater than all tasks except DP. With TFAs, though still greater than SD and STS, GT was much closer to these remaining tasks, implying a comparatively decreased reliance on transverse trunk movements during amputee gait. Upon closer inspection, however, this average was brought down exclusively by TFA1, as the remaining TFA participants continued to demonstrate a relatively higher AT ROM during gait. In patients with LBP, AT trunk stiffness has been presented as a guarding technique, restricting trunk rotation and mitigating potential for pain (Selles et al., 2011). However, the apparent decrease in TFA gait AT ROM was not as universal as it seemed, and did not coincide with a change in relative AT stiffness during GT compared to other tasks for the TFA participants. For TFA1, the decrease in AT ROM saw no difference from the control group in relative AT stiffness, and the same lack of change was seen in TFA2 for both the ROM and stiffness in the transverse plane. TFA3 and TFA4 each showed slight relative increases in AT ROM, but were accompanied by decreases in AT stiffness. This would imply that rather than an active guarding movement, the AT rotation of the TFAs was more of a passive reaction to another aspect of the gait pattern. Considering the task as a whole, there were no key attributes – other that high IO stiffness contribution in TFAs, notably skewed by TFA3 – that distinguished GT as an extreme to the remaining tasks.

During performance of the SD task, the TFA group showed comparatively increased F/E and LB ROM relative to control task differences, with trunk motion now being more on par with the other ADLs. This was coupled with no relative change in F/E stiffness, with the notable exception of TFA1 who displayed relatively increased stiffness for SD in all three planes. Opposing this, TFA1 was the only amputee participant to mimic the relative stiffness of the control group in the frontal plane, with the remaining participants showing marked decreases in stiffness. Although the peak LB stiffness was typically lower compared to other amputee tasks, TFAs showed increased back and IO muscle contribution to trunk stiffness from the leading, prosthetic side. This was not the case for the control group, as SD LB stiffness was relatively higher than other tasks, but the percent contribution values were in line with an across-task average. TFAs with canes also showed increased intact side LD activation, as the contralateral canes bore more weight during descent than other tasks. This discrepancy characterized the comparatively asymmetrical stiffening of the TFA participants. The increase in trunk rotation – specifically in the sagittal and frontal planes – was not accompanied by relatively equivalent stiffness to the control group, and the prosthetic side was responsible for most of the stiffening production. This would indicate that – for the complete cycle of the task – there is excessive trunk movement not accompanied by stabilizing stiffness. Therefore, SD was flagged as a potentially risky task with regard to acute injuries, pending further analysis.

In the control sample, the SU task required less LB stiffness than the other tasks. For the TFA group, this was true for TFA1 and TFA2. During this task, the cane using TFAs bore most of the weight with the left arm and, as such, TFAs displayed comparatively higher LLD activation during SU and lower overall relative LB stiffness values. Examining TFA3 and TFA4, however, one can see that the relative LB stiffness is higher than three of the remaining tasks. Overall, this task appeared to have few differences that were unique to one subject group over the other, though the differences between TFA participants suggests that there is a further compensation that is employed in the absence of external support.

Execution of the STS movement shows the greatest F/E trunk ROM among the five ADLs for both subject groups. Compared to the other tasks, all TFA participants showed a relative increase in LB ROM over the control group – sometimes going from lowest LB ROM in controls to highest LB ROM in TFAs – though that did not convert into significant differences in LB trunk stiffness for TFA1 or TFA2. Non-cane-using TFAs showed a relative decrease in LB stiffness for the STS task. Further examining muscular influence on stiffness, however, shows a change in percent contribution. In terms of both average and peak muscle activation, the IO and EO are the lowest of the five ADLs for the control group. Conversely, the TFA group displays average oblique activity during STS compared to the other tasks. This would appear to directly translate to changed muscle contributions to both LB and AT trunk stiffness between subject groups. In the control group, the obliques contribute less to STS stiffness than most other tasks, while TFA STS only has a lesser contribution from RIO and LIO than GT and SD, respectively. Additionally, the control group displays ES and QL contributions to LB and AT stiffness far surpassing those found in other tasks, which is not the case with TFAs. Compared to the controls, the TFA group also shows a relative between-task increase in F/E co-contraction during STS. Generally, it would appear that the STS movement require increased abdominal effort compared to other tasks for the TFA group. This is especially clear in the lateral abdominal muscle groups, which is much different from the trend displayed by the control group, who present a larger relative demand on the back musculature. This was thought to be a possible bracing compensation to account for the relative instability of the TFA STS movement (Gao et al., 2011).

During the DP task, the control group exhibited larger AT trunk ROM values than for other tasks. In the TFA group, the relatively large AT ROM was maintained but was only paired with a relative stiffness similar to the control group in TFA1 and TFA2. The DP AT trunk stiffness relative to the other tasks was lower in TFA3 and TFA4A, suggesting a passive, uncontrolled transverse rotation of the trunk. A notable difference in muscle activation was seen in the control LLD, as the left arm was used to open the door. However, in the TFA group, this unique task increase was not present, implying a different door opening strategy.

Overall, the TFA group as a whole appears to experience similar relative trunk challenges to the control group for the GT, SU, and DP tasks. During SD, the TFA group shows a progressively asymmetrical execution, with substantial increases in muscular demand on the prosthetic side compared to other tasks. The STS task appears to require greater effort from the abdominal muscles, especially laterally, in an effort to preserve the same relative stiffness as other tasks for the TFA group, while controls attain sufficient stiffness by increasing back muscle contribution. To further understand these challenges, performance of each task was examined separately.

6.2 – Summary of tasks

6.2.1 – Gait

The GT task cycle was longer, overall, for TFAs than the control group. Although it was found that – contrary to previous reports (Sapin et al., 2008) – the stance phases of the prosthetic and intact limb were not different in relative length for the TFA group, the single support times were shorter for the prosthesis, in accord with the existing literature (Schaarshmidt et al., 2012), with shorter overall step times for the prosthetic limb (Nolan et al., 2003). This leads to shorter step lengths and, as a result, the TFAs displayed less hip extension in both limbs, which was reached later in the gait cycle (Jaegers et al., 1995). While the kinematics of lower limbs have been well reported for the TFA population in existing literature, the effect that they have on trunk biomechanics has been less thoroughly covered.

The TFA trunk experienced lower sagittal trunk ROM, but greater lateral bending angles than the control group, as has been reported previously (Sapin et al., 2008; Jaegers et al., 1995). Frontal plane rotation of the trunk was in the direction of the stance leg in both groups, though the TFA group demonstrated greater lateral leans towards the prosthesis during stance phase. This was most clearly evident in TFA3 and TFA4, who did not have canes. TFA1 and TFA2 displayed lateral trunk motion more resembling that of the control group, utilizing canes to maintain medio-lateral support during stance. The use of canes has been linked to a peak transmission of 44% BW to the cane arm (Goh et al., 1986), and can increase symmetry during gait, with bilateral canes showing the greatest improvement (McDonough and Razza-Doherty, 1988). The LD muscles display activation with greater magnitude on the sides with canes compared to TFAs without canes and the control group. They show anticipatory activation to the swing phases of their respective sides (Figure 5.12), indicating the canes' role as supporting their side of the body during contralateral stance. With this clear demonstration of the BW support from the canes, TFA1 and TFA2 are shown to have eased the demand on the support limb and trunk musculature during periods of single support. TFA1 uses a cane to maintain his posture during the prosthetic single support phase, while TFA2 uses canes on either side to support the body during both the prosthetic and intact single support phases. As the canes are lateral to the body, they mainly assist in the frontal plane. Without canes, TFA3 and TFA4 compensate with their trunk musculature mainly in the LB dimension, adjusting their stiffening strategies in order to maintain spine stability. The components of this compensation will be discussed in later paragraphs.

A combination of lower prosthesis GRF (Nolan et al., 2003; Segal et al., 2006; Sapin et al., 2008), lower prosthesis braking and propulsion impulses (Schaarshmidt et al., 2012), and lower peak hip extensor moments during stance phase on the prosthetic side (Nolan et al., 2003) has led to the formation of the hypothesis in the literature that the prosthesis acts as a passive pendulum during gait, while the intact leg provides most of the forward motion. The ES muscles of the prosthetic side, however, show higher peak activation than either the intact side or control group. The timing of the prosthetic side activation takes the form of a double peak, activating just prior to TO and just prior to IC. This is a similar pattern to the intact side and control group, though the first of the two peaks is noticeably greater for the prosthetic ES in TFA1, TFA3, and TFA4 (Figure 5.11). Existing research has discovered the double peak ES activation during gait, indicating the presence of uneven peaks in normal populations (Cromwell et al., 1989; Thorstensson et al., 1982). Indeed, the control and intact sides of the current study displayed irregular peaks, but the exaggerated initial ES activation on the prosthetic side reveals a clearer disparity. Amputees have been recorded to have reduced quadriceps and hamstring activity on the prosthetic side (Bae et al., 2009), and the ES has been shown to compensate for this phenomenon in total knee arthroplasty patients by contributing more to both vertical and fore-aft CoG acceleration (Li et al., 2012). This suggests that the ES may serve a dual purpose.

During a typical gait cycle, the ES activation has been reported to help prevent both antero-posterior trunk motion – at ipsilateral IC – and lateral trunk bending – at contralateral IC – displaying a greater activation during the latter (Cromwell et al., 1989; Thorstensson et al., 1982). Certainly this appears to be true for the amputee participants in this study, but the consistently elevated activation levels appear to account for something other than resisting excessive trunk flexion. In control subjects, the ankle joints produce net positive work during gait (Winter, 1983), and reportedly contribute to body support, and forward and upward propulsion (Neptune et al., 2001). The same can be said of the knee joint (Winter, 1990). Given the lack of torque generation at the passive ankle and knee joints of the TFA participants in the current study, it is hypothesized that the ES partially contributes to the upward and forward propulsion of the prosthetic side of the body. During the majority of prosthetic leg stance, the prosthesis will act as a pendulum, but the ipsilateral ES activity was shown to be higher just prior to TO than IC (Figure 5.11). The swing phase braking muscles are reportedly active for longer in TFAs compared to control (Wentink et al., 2013), but the propulsive period of swing phase just prior and including TO was still shown to be more demanding on the ES at this time. This would appear to indicate the back's role in producing extension of the prosthesis that could ease the demand on the transverse rotational core musculature, no longer requiring them to move the prosthetic side forward.

Trunk rotational stiffness exhibited very similar patterns between the TFA and control groups in the F/E and AT directions. The TFA group's stiffness peaks later relative to the gait events, while the control group displays peaks corresponding to right and left IC. This is in accordance with the timing of the ES – a major contributor in both the sagittal and transverse planes – discussed previously. LB trunk stiffness also peaks following IC events, but this pattern differs most between the two groups. At each IC event, natural peaks in 3D trunk stiffness occur because of the previously-described function of the trunk muscles (Cromwell et al., 1989; Thorstensson eta l., 1982). These moments of restricting excessive trunk motion assist in maintaining a stable spinal column. To review, with the activation of the trunk musculature, the various muscles produce their own stiffness around the spine. In a 'guy wire' effect, the muscles' stiffness combines to increase the stiffness of any joints that the muscles cross (McGill, 2014). This stiffness restricts the movement of the vertebral joint and helps reduce the risk of excessive flexion or buckling. As such, the peaks of stiffness at the IC events allow for sufficient spinal stability to mitigate the risk of acute injury. Unlike the control group, however, the TFA participants demonstrated increases of stiffness between the IC peaks, suggesting a greater instability that must be countered with an altered stiffening strategy.

The concept of proximal stiffening details the function of the trunk musculature as it fixates the proximal attachments of a more distal muscle prior to its activation so that the mechanical effect of that muscle contraction is exhibited at the distal attachment (McGill, 2014). This allows for movement of the upper and lower extremities without a direct counter-movement, as the opposite force is transferred to the contact surface. During gait, this is partially mitigated by reciprocal arm swing, but trunk muscles still display anticipatory activation to IC and TO events; moments of increased distal segment deceleration and acceleration, respectively.

ES activity approximately coinciding with hip and knee extensors during gait was cited as constraining trunk motion at bilateral IC events (Thorstensson et al., 1982). Results from the current study show ES activity peaking after contralateral IC, but just prior to ipsilateral TO, for both control and TFA groups. This helps stiffen the torso which allows for productive distal movement of the lower limbs. With this said, average frontal plane trunk stiffness during prosthetic SS phase is consistently greater than during intact SS for the TFA group, but is relatively even between legs in the control group.

During a typical stance phase of gait, the pelvis must maintain a level position in order to achieve foot clearance for the swing phase of the contralateral leg. This is partially achieved by the hip abductor muscles on the stance leg side. It has been found that TFAs present prosthetic hip abductor torque 30% lower than that of the intact or control group legs (Ryser et al., 1988). It has also been reported that if hip abductors cannot meet the load demands during single support phases, the subject may employ lateral torso muscles on the swing side to assist with lifting and supporting the hip (McGill et al., 2009). This phenomenon is represented in Figure 6.1, as the contralateral QL, IO, and EO muscles compensate for reduced hip abductor torque.

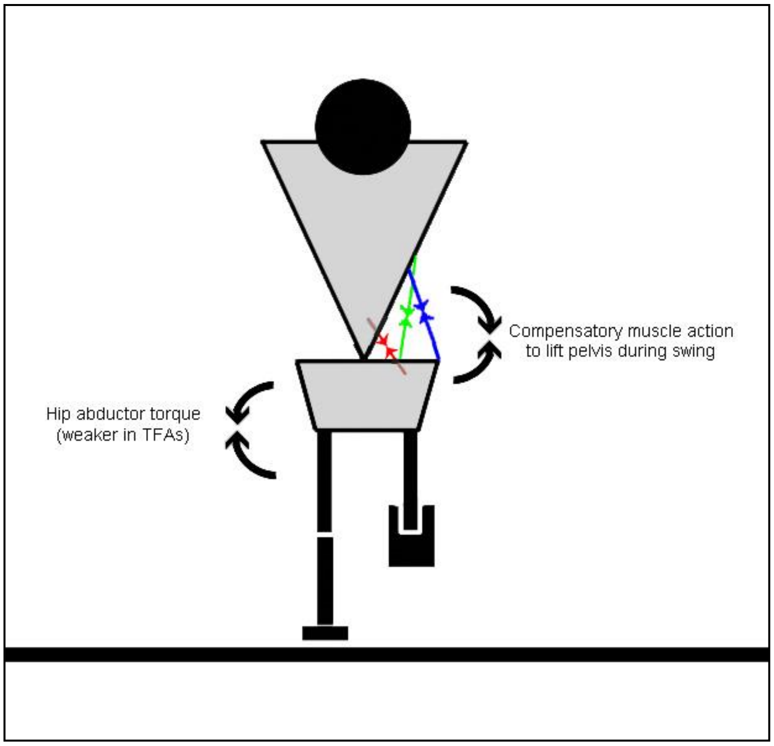

Figure 6.1: Anatomical representation of the quadratus lumborum, and internal and external obliques contracting to compensate for the lower hip abduction strength in the prosthetic limb during single support phase.

Strong prosthetic side hip abductors have been associated with increased prosthesis weight bearing and improved gait (Nadollek et al., 2002). During intact SS, the two subject groups did not show noticeable differences in stiffening strategy, but the TFA participants employed different methods to support the pelvis and trunk during prosthetic stance. Examining the percent contribution of muscles to LB trunk stiffness, TFA1 and TFA2 each displayed relatively similar right and left LTM (QL, IO, and EO) input to the control group. As touched upon in prior, the canes accept a significant portion of the upper body load and would help support the pelvis and trunk during prosthetic stance, not requiring further compensations on either side for lateral support. TFA3 and TFA4 do not have canes for support at any point during the cycle and, as such, they show a much greater contribution to LB stiffness from the RLTM during prosthetic SS. In a recent study, elevated forces from the LTM have been documented in transtibial amputees during gait tasks (Yoder et al., 2015). Specifically, it was found that the oblique muscles produce higher forces during the prosthetic SS phase, which is in line with the results of the current study. This compensatory strategy is made possible by more rapid, shorter steps (McGill et al., 2009), and the TFA group fits this criteria, with shorter single support times for the prosthetic limb.

Overall, the trunk compensations during gait seem to point to an attempt to remain stable. The contralateral cane support directly assists in maintaining stability, while the lateral trunk lean and LB trunk stiffness assist with foot clearance and vertebral joint stability. Some of these compensations, however, have a side-effect of increasing potential for injury. In particular, the ES is a dominant contributor to lumbar loading (Marras and Sommerich, 1994), and the increased asymmetrical activation and loading increases the risk of injury and LBP (Davis and Marras, 2000). Subjects experiencing LBP have themselves shown elevated ES activation during gait (Ghankhar and Kahlaee, 2015), though none of the TFA participants reported severe LBP so it can be assumed that this was not a factor.

6.2.2 – Step down

As has been reported previously (Protopapadaki et al., 2007; Riener et al., 2002), the control group maintained similar relative stance and swing phases to level gait. The TFA group, however, displayed asymmetry, favouring increased intact leg support, while decreasing single support time on the leading, prosthetic limb. This timing can be represented by a quick descent of the trail leg, reported as passively 'falling' down and experiencing greater GRF than control samples (Schmalz et al., 2007). At LfIC, TFA1 and TFA2 have a relatively similar sagittal trunk angle, while TFA3 is notably flexed. The difference in execution varies with the use of canes. The two cane using amputees begin the task at LfTO having already established a base of support with the canes on the lower surface. These two participants exhibit greater initial lead hip flexion than either the control group or TFA3, so the combined hip and trunk flexion angles of the TFA participants may be closer to one another than the reported relative trunk angle. TFA3 begins the task upright, and reaches peak flexion very close to LfIC. At this point, the trunk begins to extend for all three TFAs. While this increased flexion may seem like an unstable position to be in, comparison to the control group shows that non-amputees continue to flex their trunk following LfIC. The TFAs' resistance to further trunk flexion implies a desire to not stray from the established base of support.

Examining step down landing in lower limb amputees, Jones et al. (2006) found that the GRF vector during lead foot impact was positioned anterior to the prosthetic knee joint, working to resist passive knee flexion and potential falls. As such, they described the CoM as positioned directly over the landing limb. The current study has confirmed that this is achieved through increased trunk flexion prior to landing. It has also been reported that the prosthetic limb may display lower A/P GRF during descent than a control group (Schmalz et al., 2007; Jones et al., 2006). Preparing the trunk to begin extension at or just prior to the LfIC event would theoretically lessen this A/P GRF for the prosthetic limb. The TFAs' flexed landing and subsequent reduction in A/P GRF suggest that global stability is a foremost motivation behind the amputee participants' performance of the SD task, though this requires greater stiffness to compensate.

Also a factor during step descent is absorbing BW following the LfIC. In control populations, ankle plantar flexion and knee extension are the greatest energy absorbers during landing (McFayden and Winter, 1988). In recent studies examining drop landing, it was reported that decreases in ankle movement were associated with a more flexed landing in the hip joint, with little additional flexion (Begalle et al., 2015), and increased plantar flexion has been shown to decrease the hip's contribution to the peak support moment (Rowley and Richards, 2015). As the prosthetic ankles are passive joints and do not exhibit normal ROMs, this could be a factor in the landing strategy of the TFA participants. Additionally, only TFA1 had a microprocessor knee, while TFA3 and TFA4 had hydraulic knees, neither attaining the full function of an intact knee joint. The knee joint angular displacement during step down tasks has been reported as lower in unilateral TFA populations (Jones et al., 2006), and the current study demonstrated this. With the lack a fully functional knee joint, the restriction of movement is by design, as mentioned in terms of the sagittal flexion of the trunk. The forward flexion of the trunk is mainly seen in those TFAs not using canes, as the cane-using TFAs do not require as much absorbing knee flexion given their external support – though the early trunk extension is seen in all TFA participants. The flexion is also lowest in TFA1, as his MPK was shown to absorb some of the BW at LfIC, and the CoM need not be so far forward as to completely restrict knee flexion. Because of the restrictive knee rotation following LfIC and the lack of active torque at the ankle, the BW absorption responsibilities would theoretically be transferred to the hip and trunk.

Landing during TFA step descent is typically characterized by stiffness of the lower limb (Jones et al., 2006). Following impact, TFA1 with the MPK was the only amputee participant to demonstrate an additional period of flexion following LfIC, which even then was minimal. Because of the trunk extension necessary to maintain A/P stability, the hip joint would be the primary absorber of BW. Joint intersegmental reaction forces allow the hip extensors to become a knee extensor contributor (Arnold et al., 2005). As an absorbing flexion at either joint is characterized by eccentric contraction of the extensor muscle, resistance at the knee is allowed by the activation of the hip extensors. TFA1 and TFA2 utilized canes to assist their descent and did not show noticeable absorbing flexion at the hip joint, as it was not required given the BW support of the canes, and the absorbing qualities of TFA1's MPK joint. TFA3 and TFA4 each showed a small hip flexion following landing indicating that the BW absorption duties were transferred to the hip in amputees without the aid of a cane. The trunk segment of the control group also displays absorbing flexion after the LfIC event, but the TFA group does not, demonstrating more restricting rotational trunk stiffness in all three planes in an effort to prevent excessive movement and potential spine buckling. The slight flexion at the hip in TFA3 and TFA4, coupled with extension of the back necessary for global stability, results in a complex stiffening strategy to maintain stability of the amputee spine.

The stiffening pattern of the trunk exhibited by the control group shows a ramped increase, peaking at LfIC in all three planes. TFA1, already supported unilaterally with a cane and with minimal trunk flexion, did not show as substantial an increase in stiffness in the frontal plane following landing, but did show clear increases in sagittal and transverse stiffness. TFA2, with bilateral cane support, showed a slight increase in 3-dimensional stiffness over the course of the task, with no clear sign of landing. TFA3 and TFA4 each showed clear increases in stiffness at LfIC. TFA3 maintained a relative plateau while TFA4 showed decreased trunk stiffness after TfTO. Unlike the control group, both showed a sharp increase in stiffness at the end of the task cycle, potentially in aid of stability once recovering at rest.

A lack of stiffness during loading may contribute to the buckling of the spine (Cholewicki and McGill, 1996). The stiffening pattern exhibited by TFA3 and TFA4 showed lower relative stiffness than TFA1, TFA2, or control closer to the landing event. This apparent delay in active stiffening is potentially detrimental if not accomplished in time. During reflexive responses, the inherent stiffness of the spine is not sufficient to mitigate injury risk (Brown and

McGill, 2008). Though the increase in trunk stiffness does precede LfIC for both participants, it is a much sharper increase with less time to prepare for landing. The stiffening of the spine prior to impact allows for a more stable series of joints at the time of loading.

The absorbing, sagittal trunk stiffness displayed during this task also showed substantial TFA asymmetry, with the lead limb (prosthesis) side contributing much greater percent values from both abdominal and back musculature to the total F/E stiffness than the trail limb side. These values were closer to even in the control group. This appears to support the possibility that the prosthetic side is responsible for absorbing the landing impact. Previous research has shown that an increase in stiffness from co-contraction limits the need for complex muscular responses to sudden loading (Vera-Garcia et al., 2006). The CCI values of the TFA group are significantly higher than their control counterparts, and also higher relative to other tasks in the TFA group compared to the control group. Although an increase in bilateral co-contraction appears to contradict the asymmetry observed in trunk stiffness contribution, they both point to an increase in bracing for impact. The CCI values may be elevated due to higher values on the lead limb side.

Once again, the TFA participants' compensations are in pursuit of stability. The overall impact of the descent for TFAs is less, though it is characterized by greater stiffness, with the majority of the absorption now transferred to the hip and trunk. TFA1 and TFA2 utilized canes to support the body during descent, while TFA3 absorbed more BW at the hip joint than the control group or other TFA participants. All TFAs employed strategic trunk flexion and extension to limit A/P movement. The asymmetrical stiffness contributions and lead limb kinematics would appear to suggest that SD displays a greater demand on the prosthetic limb in TFAs. SD is, however, one of the tasks where risk of injury is more easily preventable, as a railing can be used as a substitute for a cane during descent in most stair situations; easing the impact and lessening the need for coordinated trunk stiffening.

6.2.3 – Step up

Trunk motion during the step up task displays similar patterns between control and TFA groups, with the exception of frontal plane motion during the prosthetic swing phase. The TFA group displays greater lean towards the intact leg side than the control group during this time. This is in aid of the compensatory movement employed by TFAs to create foot clearance for the prosthesis.

The most drastic difference in the performance of the step up task by TFA and control groups is the lack of prosthetic knee flexion, which would affect the foot clearance of that trailing limb. For TFAs unable to perform the task without tripping, it has been proposed that they could extend and then rapidly flex the hip joint in early swing phase to achieve ample clearance for the trail leg (Hobara et al., 2014). This method of extra hip extension during single support may put more strain on the low back, while the rapid flexion may create instability on an already uneven surface. The current study's TFA participants also demonstrated less sagittal hip extension than their control counterparts. Examining the lateral and axial hip angles, it would appear that the subjects of the current study – all able to perform the task – have already developed a compensatory strategy. An increase in hip abduction for the trailing prosthetic limb of all TFA participants, coupled with the afore-mentioned lateral trunk bend towards the intact side would suggest the use of circumduction to compensate for the low knee flexion. This compensation, however, puts the trunk segment in a potentially compromising position.

This task was the most similar between subject groups in terms of stiffening performance. After a slight increase in stiffness – especially laterally – during single support of the trail leg, a local peak is reached just prior to TfTO. This effect is diminished in TFA1, and virtually non-existent in TFA2. These participants began the task with canes already in place for support and, as such, the stiffness stays high from the start of the task to the LfIC. A lower relative increase in stiffness is needed to reach a peak value necessary to execute the task successfully. Following TfTO, the control group and TFA1 show a decrease in stiffness only to increase again – mostly in the frontal plane – during the trailing, prosthetic swing phase. TFA3 and TFA4 remain stiff following TO, and show a second increase subsequently. These second peaks correspond with local peaks in trail leg lateral hip angle, and lateral trunk lean towards the intact side. Previous research has reported that the lumbar ES displayed the largest response to added lateral bend loads, helping to initiate trunk stiffness (Chiang and Potvin, 2001). Though the opposite direction of the trunk and trail leg would work to balance loading, the RES – as well as the right LTM – still showed significantly greater contributions to LB trunk stiffness for the TFA group. While the F/E and AT dimensions act rather similarly in both the control and TFA groups, the LB stiffness of the TFAs demonstrates the larger muscle activation, reacting to the additional lateral load brought on by trunk excursion. The effect is diminished in the cane-using TFAs, but the lack of stiffness decrease followed by an additional increase in TFA3 and TFA4 show that there is a higher demand on the trunk musculature at this point in the SU cycle.

The stiffening pattern exhibited by TFA3 and TFA4 suggest that a higher stiffness is required to sufficiently stabilize the spine during the lead limb single support phase. As the trunk is bent laterally to assist in trail limb toe clearance, it is put in a compromising position if muscular activation does not act to prevent further, excessive trunk motion. As such, though the amputees predominantly demonstrated increased lateral bending, the higher stiffness may be acting as a guarding mechanism to mitigate injury risk. TFA1, who did not show much lateral bending of the trunk, had the highest lateral hip ROM, suggesting increased prosthetic circumduction. The remaining amputees, whose prosthetic abduction was only marginally higher than the control group, required the lateral trunk bend and, subsequently, the guarding LB stiffness of the trunk to prevent spine buckling.

With the majority of the SU task employing support from the intact leg, TFAs compensate not only for stability, but for safety. Circumduction, though potentially more efficient than hip hiking, is an inefficient but safer step movement, utilized for increased foot clearance when stepping over objects (Chen et al., 2005; Stanhope et al., 2014). Hip osteoarthritis is a prevalent concern in the TFA population (Benichou and Wirotius, 1982), so a repetitive, asymmetrical movement and stiffness compensation may be damaging to the health of both of the intact and residual hip joints (Lloyd et al., 2010). Raised trunk stiffness remains an effective method of stabilizing the spine, however, and as circumduction appears to be the safest compensation, the increased stiffness is required to prevent acute injury.

6.2.4 – Sit-to-stand

TFA participants universally displayed bilateral asymmetry in both trunk motion and muscle activation. Previous research has reported that the vertical GRF of the prosthesis was significantly less than a control group, while the intact hip moment was significantly greater (Highsmith et al., 2011). The four participants in the current study performed the task with high variability in frontal plane trunk motion (Figure 5.25). TFA1 transferred his cane to the prosthetic side and pushed off with that hand, performing a lateral bend towards the prosthesis. Virtually all of the upward motion was produced by the intact side, as the prosthesis was lifted from the ground and replaced when the task was complete. A base of support was maintained with the left cane. TFA2, supported by two canes, swayed from left to right through the task cycle, also alternating bilateral GMax and ES activity. TFA3 and TFA4 performed the task relatively similarly, both performing lateral leans towards the intact leg.

Muscle activation of a standard STS cycle utilize quadriceps, hamstrings, ES, and gluteal muscles activating at or prior to the seat off event (Roebroeck et al., 1994; Ashford and De Souza, 2000). As the quadriceps do not function as an active knee extensor on the prosthetic side for TFAs, the knee extension will then be produced by a different muscle. TFA1 transferred this assignment almost exclusively to the intact side. TFA2 utilized the canes for assistance, demonstrating consistently elevated activation of the LD muscles from 10% to 90% of the task cycle, while also favouring the intact limb. In both of these cases, prosthetic knee extension was mainly passive; allowed to extend with gravity as the rest of the body achieved an upright posture.

During leg extension, joint intersegmental reaction forces allow the hip extensors to become a knee extensor contributor (Arnold et al., 2005). As the participants' prostheses did not have powered knees, the hip extensors were the dominant knee extensor muscles on the prosthetic side. TFA3 and TFA4 both chose to use the intact limb as the dominant side, though with less asymmetry than the other TFA participants. Knee extension was achieved on the prosthetic side by increasing gluteal and ES activation. LGMax activated at approximately 20% of the task cycle for TFA3 and 25% for TFA4 and both remained active at 60%MVC for the remainder of the task. RES and LES activation for TFA3 and TFA4 (Figure 5.26) most clearly demonstrated the asymmetry compared to the control group. Both the right and left ES quickly peak just prior to seat off, but the RES activation then gradually declines. Between seat off and full hip extension, the LES is fully active before turning off during the stabilization phase. This suggests that the RES helps initiate the motion, but LES is a larger contributor to trunk extension overall. Knee extension begins with the activation of the ES and GMax muscles (Figure 5.25) and thus are assumed to be the major muscles contributing to upward motion.

The timing and sequencing of muscle activation can drastically affect the success of stiffness in stabilizing the spinal column. Altering the proximal to distal activation may create unwanted movement as opposed to restrictive, stabilizing stiffness. Muscle activation too late may result in loading an unstable joint, risking acute injury, as would activating and shutting off too early. If activation occurs early and sustains for longer than normal, it will result in a stable joint, but is not ideal for free movement. When performing the STS movement, rotational trunk stiffness in all three planes peaked at seat off. In the control group, stiffness then decreased to varying degrees with a brief influx after hip extension as the participants became stable. TFA1, TFA3, and TFA4 each reached a local maximum at seat off, but during hip and trunk extension stiffness in all dimensions remained high. Maintaining stiffness throughout the task ROM is inefficient and may be a result of a compensatory strategy.

Example trials from both control and TFA groups are presented in Appendix F. At the SO event, the knee extensors would typically be the dominant contributor to upward motion (Roebroeck et al., 1994). During this knee extension, the trunk muscles of the control group are not particularly active, though it corresponds with a period of great sagittal trunk extension. This means that proximal stiffening is not being employed, allowing for a proximal reaction to the distal movements. Therefore, it is hypothesized that, through intersegmental joint reaction forces, the extension of the leg would lead to a passive extension of the trunk. Stiffness of the trunk at this point would hinder the intended motion and, as it is not needed to increase stability for most control participants, is not employed. This example trial resulted in a stiffness pattern of Method

A (Figure 5.27), and slightly increased back muscle activation are present in the other stiffening methods.

Comparing this to the task performance of an example TFA participant (Appendix F), a few differences are clearly evident. The abdominal muscles activated first, followed by the back extensors and gluteals almost simultaneously. At this point in the control group, the abdominal muscle activation decreased, but the TFA4 example shows a local peak activation in both left oblique muscles. This corresponded with peaks in the back extensor and gluteal muscles to form a marked increase in 3-dimensional trunk stiffness and initiate the seat off action. In contrast to the control group, TFA trunk extension was not passive. Following seat off, the GMax muscles maintain their activation level. As discussed previously, the GMax, as a major hip extensor, is now a significant contributor to prosthetic knee extension. In order for the hip extension effects to be expressed at the more distal knee joint, the trunk muscles remain active to stiffen the proximal joints. At this point, the back muscles are also thought to actively contribute to trunk extension. Throughout full body extension, the sustained LES, LSLM, and LIO activity works to increase overall trunk stiffness, especially in the frontal plane.

Unlike in the control group, the TFA participants displayed higher F/E co-contraction during STS than other tasks. Compared to the controls, TFAs also show a relatively higher percent contribution from the abdominal muscles to trunk stiffness. Van Dieen et al. (2003) found that during lifting, co-contraction contributed to maintaining trunk stiffness and overall stability. During TFA STS, it has been hypothesized that extra muscular effort is required for stabilization in the frontal plane (Gao et al., 2011). If the STS movement is unstable for TFAs, the movement may be perceived as a high performance task, which is associated with increased joint stiffness (Butler et al., 2003). As such, the continuous trunk stiffness through the STS task is thought to be a strategy to control movement stability and perform the task safely.

In accordance with how the movement was performed, the muscle stiffness contributions from each of the TFA participants were different. As a main extensor for the STS movement, the ES was found to have asymmetrical contributions from TFA2, TFA3, and TFA4. TFA2 displayed a greater contribution from the right side, though ES activation showed alternating activation from the left and right sides. TFA3 and TFA4 each showed a bias towards the left side, with activation patterns having been covered previously. In terms of muscles responsible for indirect knee extension, this resulted in relatively less bilateral ES co-contraction than the GMax muscles. ES co-contraction was still larger for this task than the other gait-based tasks, as there was no alternating movement. As mentioned prior, the ES is a dominant contributor to lumbar loading. Given that the STS movement is performed frequently throughout the day, and is characterized by high, asymmetrical stiffness throughout trunk extension, it is believed that this task is a less ideal activity – with regards to LBP – when performed improperly.

6.2.5 – Door pull

The variability of the door opening task was most evident in the trunk angles. While the prosthetic knee and ankle displayed generally lower ROM than the control group, the trunk motion was most notably different in the lead up to the door opening. Following contact with the handle, both groups performed a right lateral bend to move away from the door, and an axial twist to assist with the pull. Though they were not significantly different, the amputees demonstrated a lesser axial twist movement. Anecdotally, TFA1 and TFA2 executed the door pull with a quick jerk; enough for the door to be held open as TFA2 replaced his hand on the cane to walk through.

Despite exhibiting increased stiffness during the opening of the door, most values were not significantly different from gait. The substantially greater co-contraction values during the DP task were mostly a function of time. When normalized for task length, DP was much closer to the remaining tasks (Table 6.1). Considering that steps were a major component of the DP task, this comparison seemed appropriate. The TFAs uniformly showed increased relative stiffness prior to the door opening event, while the control group displayed a significant relative spike in stiffness when opening the door. At door pull, one main difference between groups was the stiffening in the AT direction, with TFAs displaying a much less clear local peak. As previously mentioned, a 'jerking' movement was utilized by some amputees when opening the door. A heavier or more resistant door may be more difficult to open this way, and the difference in subject group stiffening may show more distinct compensations if this were the case.

Table 6.1: Co-contraction values represented as an average of 100 normalized data points to account for the relative length of tasks.

Subject group	Task	Bilateral CCI	Flexion/Extion CCI
Control	GT	8.6	6.7
	SD	8.3	6.6
	SU	9.3	7.2
	STS	10.8	5.8
	DP	8.6	7.1
TFA	GT	33.7	24.1
	SD	35.8	25.1
	SU	38.3	24.0
	STS	35.9	25.2
	DP	28.9	22.2

As mentioned in the prior discussion section, the DP task exhibited increased AT motion, with decreased AT stiffness in the TFAs without canes. This lack of guarding seems to imply that a freedom of movement – potentially multi-planar movement – is important in the successful completion of a door opening task. The use of canes was accompanied by relatively high AT trunk motion and stiffness, implying an active, controlled rotation of the trunk which was more in line with the control group. This variability in how the TFAs performed the task did not seem to have any reliable biological basis, and appeared to act counter to the idea that the canes negated the need for an increase in sufficient stiffness. Generally, the comparison between the TFA and control groups was found to be inconclusive.

6.3 – Low back pain & Injury

Common factors associated with LBP include excessive lumbar spine loading, asymmetrical movement (Davis and Marras, 2000), and repetitive low to moderate spine loading (Kumar, 2001; McGill, 2007). It has been hypothesized that LBP may be the cause of movement changes (Hodges and Moseley, 2003), and pharmaceutically-induced LBP has been shown to cause reduced muscle force and stiffness, theoretically in an effort to limit pain (Ross et al., 2015). LBP is a prevalent problem in lower limb amputee populations (Devan et al., 2012), with movement compensations employed well before the onset of LBP. Altered trunk motion and loading has recently been reported in TTAs and TFAs without LBP (Hendershot and Wolf, 2014; Yoder et al., 2015).

In addition to accumulated LBP, pain may be a result of an acute injury. In an effort to lower the risk of injury, the trunk muscles must produce stiffness sufficient to prevent buckling of the spine. Within each of the tasks executed by TFAs, certain events demanded a relatively greater stiffness to sufficiently stabilize the spinal column than their control counterparts. If a TFA experiences muscle weakness or dysfunction in muscle activation timing and sequencing, the elevated sufficient stiffness may not be met. As such, these events may be viewed as areas of potential injury and LBP. The current study had participants with various degrees of LBP.

The results of the Owestry Low Back Pain Disability Questionnaire completed by both TFA and control participants are shown in Figure 6.2. The control group ended up with a score of 2.6 (4.6)%, while the four TFA participants - in ascending numerical order – had scores of 32.5%, 10%, 0.5%, and 12.5%. With the exception of TFA1, who fell into the 'moderate' category, each participant displayed minimal disability within the 10% difference that reveals meaningful change. TFA1, with a contralateral ankle replacement, would have had to adapt to additionally altered movement which may have affected his overall LBP score. No participants displayed 'severe disability' or higher from LBP.

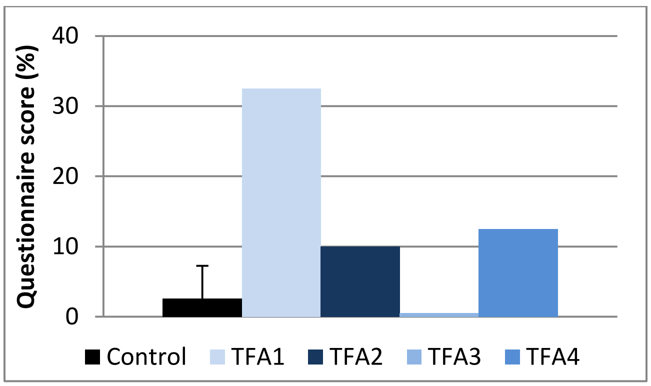

Figure 6.2: Owestry questionnaire scores for the control group and the four TFA participants.

During the GT task, the TFA group displayed increased frontal plane trunk stiffness during prosthetic stance, with asymmetrical LTM stiffness contributions. TFAs reportedly show no significant difference in internal and external oblique muscle thickness and function (Springer and Gill, 2007), but have been shown – along with the QL muscle – to exhibit elevated activation levels for TTAs during gait (Yoder et al, 2015). In the current study, TFA1 and TFA2 – with canes – displayed bilateral stiffness contributions from these muscles in line with those of the control group. TFA3 and TFA4 – without canes – showed a clearer asymmetry, with an overall greater activation of the LTM group. Scoliosis has a high prevalence rate among TFAs (Burke et al., 1978), and is associated with general LBP as well as specified pain in the QL, psoas, and EO muscles (Travell and Simons, 1992). Because the TFA participants without canes utilized these muscles as part of a compensatory technique to lift the pelvis and achieve toe clearance, a greater demand was place on TFA3 and TFA4's LTM relative to the control group and the cane-using TFAs. It is thought that the enhanced pelvis-supporting function taken on by the LTM is a factor in developing specified LBP as described with regards to scoliosis or other asymmetrical trunk changes.

Asymmetry was also a factor for TFAs during the SD, SU, and STS tasks. The compensations at the hip and trunk during landing of the SD task were more fully realized on the prosthetic side for all TFA participants, while upward force production on the prosthetic side during STS was transferred to the hip and trunk of TFA3 and TFA4. With increased prosthetic hip abduction and lateral trunk bend, the TFA group also compromised their medio-lateral stability during the SU task. To maintain stable movement, the participants compensated with increased rotational trunk stiffness. As mentioned prior, the force required for muscle stiffening is the main contributor to spine loading (McGill et al., 2009), though which muscle is the overall dominant contributor depends on body position (Kavcic et al., 2004). The asymmetrical movement and stiffening contributions during these tasks would contribute to asymmetric spine loading and, potentially, LBP.

Loading the spine repetitively at a low to moderate level has also been linked to LBP. All of the gait-based tasks are prone to repetition, and the STS task has the potential for high frequency (Dall and Kerr, 2010; Riley et al., 1991), though the relative effect of the repetition will depend on many internal and external factors. For each of these tasks, however, the TFAs also demonstrated increased asymmetrical movement and relatively prolonged loading, making the repetitive characteristics of the tasks an additional risk factor. The repetitive or frequent nature of these ADLs attaining moderate stiffening and loading may – as well as being associated with LBP – increase risk injury.

Repetitive loading has been shown to reduce stiffness (Graham and Brown, 2014), and a lack of sufficient stiffness has been shown to lead to buckling of the spine or potential soft tissue injury (Cholewicki and McGill, 1996). During TFA gait, increased, asymmetric forces and moments around the low back have been reported, particularly in the frontal plane (Hendershot and Wolf, 2014). This, along with the asymmetrical stiffening reported prior, the repetition in walking and stair ascent or descent tasks makes these potentially risky activities if performed excessively or with significantly altered form.

Specifically within the GT task, different phases of the cycle presented various levels of injury risk. During the double leg support phases, natural peaks in stiffness occur due to the afore-mentioned role of the trunk muscles in typical gait (Cromwell et al., 1989; Thorstensson eta l., 1982). Through the single support phases, however, 3D trunk stiffness decreases in the control group evenly on the right and left sides. For the TFA participants, the stiffness during the single support phase of the intact leg decreases as in the control group, but the prosthetic single support phase displays a maintenance or extra peak of LB stiffness. This elevation in required stiffness implies an inherent instability during this phase of the movement. Combined with the asymmetrical muscle contributions seen in the non-cane-using TFAs, the PSS phase is considered at greater risk of injury with dysfunctional stiffening of the spine. Because TFAs demonstrate a greater stiffness necessary to develop a stable spine than the control group, a lack of stiffness in the prior subject group would lead to an unstable spinal column and put the participant at risk of buckling and injury, specifically in the frontal plane. Though all TFA participants were able to achieve sufficient stiffness during the GT task, an amputee that is unable to consistently achieve this would be at risk; a concept that carries over to all tasks.

During SD, the TFA participants displayed a number of compensations to avoid risk of injury to the low back. An initial flexion of the trunk with no additional flexion following LfIC allows the amputee to maintain an extended knee without collapse. In the flexed position, the trunk is in a compromising position with risk of spine buckling increased if sufficient stiffness is not met. As a result of the TFAs' reported relative increase in necessary stiffness, however, there is little absorption of impact and the trailing, intact limb must follow quickly and produces a GRF greater than that of the control group. As such, the stiffness of the TFAs' landing is a requirement both in terms of avoiding risk of falling – global stability – and risk of spine buckling – lumbar stability. An inability to achieve this sufficient stiffness would make the LfIC event the most at risk of injury.

Once again, all TFA participants in this study were able to achieve sufficient stiffness for the SD task, but were able to mitigate risk in different ways. TFA1 and TFA2 both had canes to alleviate the trunk movement required to maintain global stability and, as such, lessened the stiffness required of trunk musculature to maintain spine stability, Additionally, TFA1's MPK assisted in absorbing some of the impact at LfIC, and lesser trunk compensations were required. Without these aids, however, TFA3 and TFA4 demonstrated sharp, asymmetrical increases in stiffness to account for the elevated instability. Should they have not been able to achieve this higher level of sufficient stiffness or else failed to time the stiffening correctly, this event would be most at risk of injury when compared to the control group execution of this task.

Another discrepancy between the sufficient stiffness required from the two subject groups is found during the SU task. In the control group, a relatively even double peak of stiffness is exhibited just prior and following the TfTO even, mirroring the initiation of upward force and mid-single support phase, respectively. TFA1 and TFA2, with canes for support, demonstrate less exaggerated, though still even, stiffness peaks. Both were able to achieve clearance of the trailing, prosthetic limb with LB trunk movement no larger than that of the control group, with TFA1 especially showing a notable increase in lateral prosthetic hip excursion without a contralateral trunk lean. TFA3 and TFA4, however, did not have the support of the canes, and compensated for the prosthesis circumduction with a larger LB trunk movement. This put the trunk in a position requiring stiffness relatively larger than that displayed by the control group to achieve a stable spine. The increase in sufficient stiffness was exemplified by uneven stiffness peaks in the non-cane-using TFAs. Following the initial spike in stiffness essential in achieving the upward movement, there was no decrease in stiffness. Instead, trunk stiffness was maintained until the mid-single support phase, at which point an additional increase was shown. As the excessive trunk movement was seen predominantly in the frontal plane, so the difference in stiffening strategy was unique in the LB direction. In this regard, risk of injury is greatest during the LfSS phase of the SU task in the presence of stiffening dysfunction. The risk is mostly seen in the frontal plane, and appears to be somewhat mitigated with the use of canes.

When performing the STS task, all TFA participants displayed more sustained elevated stiffness than their control counterparts. This would imply a more active trunk extension, as described earlier, but also a potential increased stiffness guarding to account for spine instability during the movement. As such, this indicates that the risk of acute back injury for the TFA participants was higher than that of the control group throughout the extension phase. Both groups showed a peak in stiffness just preceding the SO event, as the force required for upward movement was generated. In the majority of cases for the control participants, stiffness then decreased to near-baseline levels, while TFA1, TFA3, and TFA4 decreased 3D stiffness marginally, and maintained an elevated stiffness at SO to account for the instability inherent to the task. TFA2, who felt he could not perform the STS task from the standard seat height, was shown to have a drastically increased stiffness prior to SO – notably in the frontal plane – but it decreased by the SO event. In both the F/E and LB directions, TFA2's trunk stiffness demonstrated this drop, then a delayed increase during the extension phase, followed by another decrease and increase around HE. This suggested a muscle timing or sequencing dysfunction in the stiffening strategy of TFA2 to allow for sufficient stability to perform the STS task without injury. Unable to perform the task from the standard height, the apparent stiffening dysfunction continued to present a risk for TFA2 at an easier, higher level. The relatively high stiffness exhibited by the remaining TFA and control participants appeared to sufficiently protect against spine buckling. However, these TFAs still required an increase to sufficient stiffness where the control group did not and, consequently, the SO event was considered the most at risk of injury during STS, most notably in the frontal plane.

During the DP task, few definable areas of injury risk were observed. The more passive AT movement of the TFAs during this task described earlier may become a problem should the amputee encounter increased resistance from the door mechanism. Because this was not the case, any hypothesized increase in required AT sufficient stiffness – and subsequent risk of injury – cannot be confirmed. The TFA participants also fairly consistently displayed lower relative increases in 3D stiffness at the door pull event than the control group. As each participant was able to perform this task safely, any application of this information would need further measurement and observation. However, overall, across the other four ADL-based tasks, there existed areas of increased stiffening demand for the TFA participants, suggesting increased risk of injury if these sufficient stiffness levels are not met, and with acute injury may also follow LBP

Generally, stiffness should maintain an optimal level: sufficient to perform required tasks, but not so high as to overload the low back structure. As TFAs display elevated trunk stiffness to perform common activities, the repetition of these tasks may lead to the onset of LBP. Given the potential frequency of the more demanding STS task (Dall and Kerr, 2010; Riley et al., 1991), greater repeated loads may also lead to stiffening dysfunction. With a higher sufficient stiffness required, TFAs are at greater risk of acute injury in the presence of stiffening dysfunction, especially in the LB direction. Therefore, it would appear safe to assume that trunk stiffening function – especially with regards to the lateral trunk musculature – is of great importance when planning a course of rehabilitation for TFAs that may mitigate LBP.

6.4 – Rehabilitation

When designing rehabilitation for functional stability, the predominant approach is to begin locally, helping to stiffen specific muscles while restricting movement, and gradually move globally (Comerford and Mottram, 2001). The abdominal draw-in maneuver (ADIM) is a commonly used exercise which primarily activates the deep abdominal muscles and has been shown to increase lumbar stiffness when included in training programs (Puntumetakul et al., 2013). Reflexive contraction of the deep abdominals is sometimes not present in the presence of LBP (Miura et al., 2014), and therefore it must be considered when designing rehabilitation to avoid such pathology. It has been reported, however, that a core stiffening action without drawing in is preferable from both a stiffness and injury mitigation perspective (McGill, 2014). Indeed, core stiffening during gait has been shown to stabilize the pelvis and spine, and increase hip extension (Madokoro et al., 2014); a ROM lacking in the TFAs participating in the current study.

In this study, areas of instability for TFAs relative to the control group were found mostly in the frontal plane. In order to not overload the spine, active stiffening in the LB dimension without needlessly stiffening the other two dimensions would be preferable. A specified training program to isolate the muscles that predominantly contribute to the LB stiffness of the trunk would help alleviate some of this injury risk. Training the specific muscles to activate when needed would allow for additional active stiffness without overloading the spinal column in the F/E and AT dimensions. Recent research has shown that a long term isometric contraction training program – and, to a lesser extent, a dynamic program – has increased passive trunk stiffness (Lee and McGill, 2015). If directionally trained, this type of program may be beneficial in relieving the relatively increased active stiffening demand in the frontal plane for TFAs. The necessity of this training may be somewhat mitigated, however, as hip abductor strength training programs have resulted in higher scores for standard TFA rehabilitation tests that include gait and STS actions (Pauley et al., 2014). As previously noted, in the absence of a successful strengthening of the hip abductors, the trunk musculature compensates – specifically in the frontal plane. Supplementary strength training of the hip abductors would help ease the demand on this dimension of trunk stiffness.

In addition to repetitive moderate activation of the lateral trunk muscles, the current study also revealed greater, more sustained stiffness in the GT, SU, and STS tasks. As increased muscle stiffening remains a contributing factor to spine loading, some of these tasks that were found to be more demanding for TFAs may demonstrate excessive loading. In some of these cases, it is because of specific events during the tasks that indicate increased injury risk. With muscle strength training being a typical aspect of TFA rehabilitation, any dysfunction exhibited by TFAs – such as TFA2 at the SO event of STS – may be a product of problems with muscle activation timing and sequencing. Timing of stiffness production is important for all the tasks, perhaps mostly so at the areas of injury risk mentioned in the previous section. Stiffness training specific to individual tasks may help both alleviate stiffening demand and mitigate risk of injury. Training muscle activation timing would also help in lessening the need for sustained stiffness during these tasks, allowing the amputee to focus the energy on the moments of greatest instability.

With this information considered, it would suggest that a TFA rehabilitation program should include both localized and gross movement stiffness training. Locally, specific training of the lateral trunk muscles would be most beneficial, while learning to incorporate stiffening techniques within the ADLs is equally important in preventing areas of injury risk and LBP. Gait has been reported as exhibiting decreased tissue loading compared to other rehabilitation exercises (Callaghan et al., 1999), and should remain a main activity used to help with transferring local stiffening techniques to gross movement. Because the TFA population is not as homogeneous as the control population, each amputee's areas of difficulty will be unique. Carrying tasks have also been reported as requiring trunk stiffness to maintain torso and pelvis stability, and are integrated movement activities that can be used to expose weak links that would not be revealed during lifting or isolated contraction tasks (McGill et al., 2009). Increasing load during gait linearly increases muscular stiffness (Caron et al., 2015), so a stepped increase in task difficulty would potentially be of benefit for stiffness training. Because of the variability and unique challenges present in the TFA population, rehabilitation strategy must be specially suited to each amputee in order to be effective.

CHAPTER 7 – Conclusion

Over the course of the five ADL-based tasks, the movements of the TFA participants helped inform the definition of conclusive compensatory strategies affecting the low back. Throughout gait-based movement, the role of LB stiffness indicated the strategies that each participant utilized to achieve safe ambulation. During the prosthetic single support phase, all TFA participants displayed increased stiffness compared to the intact single support, which was not the case in the control group. However, only TFA1 and TFA2, with the external support of their canes, demonstrated relatively increased LB trunk motion during the PSS phase. TFA3 and TFA4, without canes, utilized increased LB stiffness to restrict trunk motion and increased contribution from the LTM to allow for toe clearance during PSS phase. Further compensations were specific to the individual tasks.

During the GT task, increased AT stiffness and decreased transverse plane trunk movement has been hypothesized as a guarding mechanism against LBP (Selles et al., 2001). For the TFA participants without canes, this appeared to not be the case, as the relative decline in AT ROM was not accompanied by a change in AT trunk stiffness. This apparent passive decrease in AT movement is thought to be a factor of the reduced step length exhibited by TFAs.

For the SD task, it was concluded that, in the absence of a cane or other such supporting device (eg. railing), TFAs transferred a portion of the absorption following LfIC from the knee to the hip. This was observed in the increase in reactive hip flexion and decrease in knee flexion in TFA3 and TFA4 compared to the control group.

With increased LB trunk movement during the SU activity, TFA participants without canes displayed increased relative LB stiffness as an apparent guarding mechanism. While the control group and TFA1 exhibited a reduction in LB trunk stiffness following LfIC, TFA3 and TFA4 maintained an elevated stiffness and increased again following TfTO.

As with the SD task, part of the extension power was transferred to the intact leg and the hip joint of the prosthetic limb in TFA participants during the STS protocol. This was exhibited through increased LB trunk motion and increased asymmetry in GMax activity in TFA participants. These differences – particularly the asymmetrical GMax activity – were especially noticeable in the non-cane-using TFA participants.

The DP task, with coordinated movement in multiple planes, was the most varying. It was decided, however, that the use of canes allowed for TFA movement more closely resembling that of the control participants. TFA3 and TFA4, unlike the cane users or control group, displayed decreased F/E and LB trunk motion, with increased AT motion and decreased relative AT stiffness. This comparison to the control group did not appear to have any biological significance, as more information would be needed.

Generally, across the five tasks, it was concluded that STS had the largest increase in required effort for the TFA participants, with specific increased demand on the abdominal muscles. With asymmetrical gluteal and trunk muscle activity, and greater LB trunk excursion among TFAs, the STS task also appeared to require the largest effort to maintain stability and, therefore, was considered the most inherently destabilizing compared to the control group. However, stiffness sustained throughout the movement and less passive trunk extension allowed the TFA participants to effectively guard against injury. The stiffness levels achieved during the GT and SU tasks also appeared to be sufficient in resisting instability. During SD, muscles on the leading, prosthetic side contributed much more to trunk stiffness than they had during the other tasks, and the trunk does not flex following landing. This implies that a high demand on the prosthetic side trunk musculature is present for TFAs, while the control group maintains even challenges across tasks. These compensations present altered demands on the trunk structure, which may affect the probability of successful pain-free mobility.

7.1 – Low back pain & Injury

In order to mitigate risk of injury, the trunk muscles must produce sufficient stiffness to prevent buckling of the spine. For each of the ADL-based tasks there existed certain events for TFA participants that demanded a relatively greater stiffness to sufficiently stabilize the spinal column. These events can be viewed as areas of potential injury risk if the elevated sufficient stiffness is not met.

During the GT task, PSS phase was considered most at risk of injury, specifically in the frontal plane. For SD, the LfIC event resulted in greater stiffness demand for both F/E and LB. The LfSS phase of SU also displayed an increased stiffness – notably in the frontal plane – to account for the instability of the movement. In all three of these cases, the use of canes alleviated some of the stiffness demand, as did TFA1's MPK specifically during the SD task.

For the STS task, all TFA participants displayed more sustained elevated stiffness than their control counterparts. Both groups, however, showed a local peak stiffness just preceding the SO event. TFA2, who felt he could not perform the STS task from the standard seat height, was shown to have a drastically increased stiffness prior to SO, though it decreased by the SO event. The relatively high stiffness exhibited by the remaining TFA and control participants appeared to sufficiently protect against spine buckling. In the case of TFA2, however, an apparent dysfunction in stabilizing strategy did not allow him to safely perform the standard STS task.

Consequently, the SO event was considered the most at risk of injury during STS, most notably seen in the frontal plane.

Along with the potential acute injuries, common factors associated with LBP include excessive, sustained lumbar spine loading, asymmetrical movement (Davis and Marras, 2000), and repetitive low to moderate spine loading (Kumar, 2001; McGill, 2007). Comparing the TFA participants to the control group, these factors were displayed in several of the tasks.

Among the five tasks, the TFA participants' increasingly asymmetrical movement was most readily observed during the GT, SD, and STS tasks. During the GT task, the LTM demonstrated the most asymmetry during PSS phase in the TFA participants without canes. This compensation has already been hypothesized as being linked to LBP associated with scoliosis (Yoder et al., 2015). SD and STS each displayed asymmetrical contributions to trunk stiffness in order to either absorb impact or generate full body extension.

Sustained trunk stiffness directly leads to a sustention of loading on the lumbar spine, and a predisposition to LBP. Compared to the control group, the TFA participants demonstrated consistently sustained stiffness in both the SU and STS tasks. The SU task change was most clearly represented by those without canes to compensate for the increased LB of the trunk following TfTO. During STS, the stiffness was maintained to account for an apparent frontal plane instability.

Repetitive loading has been shown to reduce stiffness (Graham and Brown, 2014), with a lack of sufficient stiffness having been shown to lead to buckling of the spine or potential soft tissue injury. Each of the gait-based tasks demonstrated potential for repetitive stiffness. In each of these cases for TFAs, the repetition was also coupled with additional predisposing LBP factors. This included the asymmetry found in GT and SD, and the more sustained stiffness during SU.

Overall, the TFA participants demonstrated a propensity for increased risk of both acute spinal injury, and gradually-developed LBP. The most at risk activities for TFAs were determined to be the SD task – specifically at LfIC – and the STS task – specifically at the SO event. Though TFAs demonstrated LBP determinants in several of the tasks, the SU and STS tasks were decidedly less than optimal. Given the potential frequency of the more demanding STS task (Dall and Kerr, 2010; Riley et al., 1991), the repetition, asymmetry, and sustained stiffness exhibited by TFAs in these two tasks would indicate a higher level of risk of LBP.

7.2 – Rehabilitation

Typical lower limb amputee rehabilitation involves muscle strengthening and gait training. However, while training and evaluation may focus of task performance, assessment of trunk stiffness is difficult and, in some cases, unreliable (Latimer et al., 1996). This means that, though a TFA patient may score well on mobility or isolated strength tests, the transferability of sufficient stiffness generation to ADLs may exhibit dysfunctional qualities that are not discovered until initial rehabilitation is complete. In order to prevent this, it is suggested that TFA rehabilitation include strategies both specific and gross motor movements designed to encourage sufficient trunk stiffness.

Generally, the frontal plane appeared to be most prone to spine instability for the TFA participants. A specified training program to isolate the muscles that predominantly contribute to the LB stiffness of the trunk would help alleviate some of this risk without overloading the spinal column in the F/E and AT dimensions. Recent research has shown that a long term isometric contraction training program – and, to a lesser extent, a dynamic program – has increased passive trunk stiffness (Lee and McGill, 2015). If directionally trained, this type of program may be beneficial in relieving the relatively increased active stiffening demand in the frontal plane for TFAs. The necessity of this training may be somewhat mitigated, however, as hip abductor strength training programs have resulted in higher scores for standard TFA rehabilitation tests that include gait and STS actions (Pauley et al., 2014). As previously noted, in the absence of a successful strengthening of the hip abductors, the trunk musculature compensates – specifically in the frontal plane. Supplementary strength training of the hip abductors would help ease the demand on this dimension of trunk stiffness.

Additionally, stiffness training specific to individual tasks may help both alleviate stiffening demand and mitigate risk of injury. An example of this would be the STS task. The dysfunction exhibited by TFA2 did not appear to be an inability to produce the sufficient level of stiffness, but was rather an issue of timing and sequence. Similarly, to avoid injury during the SD task, the timing of stiffening plays a large role. As such, training muscle activation timing would also help in lessening the need for sustained stiffness during these tasks.

With this information considered, it would suggest that a TFA rehabilitation program should include local stiffness training, specifically of the lateral trunk muscles, as well as learning to incorporate stiffening techniques within ADLs. Carrying tasks have been shown to require trunk stiffness to maintain torso and pelvis stability, and are an integrated movement activity that can be used to expose weak links that would not be revealed during lifting or isolated contraction tasks (McGill et al., 2009). Given the variability of TFAs, however, each rehabilitation program must be individually specialized to be effective.

7.3 – Limitations

This study included several limitations. One such limitation was the participant group design. The TFA population is inherently variable, and there is no guarantee that a larger sample would have helped with homogeneity. The four TFAs recorded represented a diverse sample of the population, accounting for two types of prosthesis suspension, three types of knee joint, and one, two, and no-cane conditions. With this said, some variable factor of TFAs – notably residuum properties – were not taken into account. Given the potential differences in performance of something like prosthetic hip abduction with changes in residuum length or distal muscle attachments, this should be taken into account in the future when examining trunk compensations. The control group was chosen to represent a homogeneous population, so each unique characteristic of the TFAs would show differences. Confirmation of some concepts presented in the discussion would require direct comparison to a conditionally-matched control group.

In terms of data collection, MVC values were obtained through isometric contractions, with greater than 100%MVC being a common occurrence during dynamic movements. Pathological populations have also shown an inability to properly perform MVC contractions. Although all participants were able to perform the MVC tasks, it is possible that reactive effort will far exceed the recorded voluntary effort. This may then affect the accuracy of the trunk stiffness model, which has its own set of limitations.

As a modeling of the second derivative of muscles' stored elastic potential energy, the model takes into account active muscle force production and passive muscle force based on its relative geometry. Not taken into account are other passive structures of the trunk or active forces from other deep trunk musculature that contribute to overall stiffness. As previously mentioned, Lee and McGill (2015) presented a training program that increases passive trunk stiffness. Given the range of mobility in the TFA participants, it is possible that they also demonstrated varying levels of passive stiffness. Also a limitation of this iteration of the model was that it was only focused around L4-L5, and not the full range of lumbar joints.

All of these limitations affect the applicability of the study results. More directly, the designs of the tasks are also not fully representative of ADLs in everyday life. Though gait was performed across the length of the room, data from only one stride was used. During step up and step down, a single step was used. This was rationalized based on the inability of TFAs to perform step-over-step stair ambulation, and may be representative of other obstacle navigation, such as curbs. However, no railings were used, and the task began from a stationary standing position; an unlikely situation in daily life. The door pull task did allow for a walk forward aspect, but the door frame was not restricted in width, giving the participants an excess of options for navigating around the handle and through the doorway. Sit-to-stand, though more representative of the traditional movement, was performed in a seat with no armrests.

7.4 – Future directions

Though increased trunk stiffness is an indication of higher spine loading, it remains an indirect measurement. Some papers have recently begun to look at joint forces and loads in the low back during lower extremity amputee gait (Hendershot and Wolf, 2014; Yoder et al., 2015), but it appears that no research has examined other ADLs. This would be the next step it characterizing influences on TFA LBP.

Further task-specific subjects may also be of interest. For example, a deeper, more direct examination of the relationship between prosthetic hip abductors and trunk musculature – especially LTM – would give more information on the specific strategies that TFAs employ to maintain stability during gait. The door structure used in the current study was also quite easily moveable compared to some doors in everyday life. Regulating or restricting the methods by which TFAs open doors may help maintain overall stability.

The rotational trunk stiffness modelled in this document was a representation of participants not experiencing debilitating LBP. The cited work on different rehabilitation programs designed to impact trunk stiffness would give good context for developing a program that can be introduced to TFAs suffering and not suffering from LBP. A longitudinal study on the effects of a trunk stiffening rehabilitation regime may provide some perspective necessary in identifying characteristics of TFA LBP.

REFERENCES

Arjmand N, Shirazi-Adl A. Role of intra-abdominal pressure in the unloading and stabilization of the human spine during static lifting tasks. Eur Spine J 2006; 15: 1265-1275.

Arnold AS, Anderson FC, Pandy MG, Delp SL. Muscular contributions to hip and knee extension during the single limb stance phase of normal gait: a framework for investigating the causes of crouch gait. J Biomech 2005; 38: 2181-2189.

Ashford S, De Souza L. A comparison of the timing of muscle activity during sitting down compared to standing up. Physiotherapy Research Int 2000; 5(2):111-128.

Ayis S, Gooberman-Hill R, Bowling A, Ebrahim S. Predicting catastrophic decline in mobility among older people. Age Aging 2006; 35:382-387.

Bae TS, Choi K, Hong D, Mun M. Dynamic analysis of above-knee amputee gait. Clinic Biomech 2007; 22:557-566.

Bae TS, Choi K, Mun M. Level walking and stair climbing gait in above-knee amputees. J Med Eng Tech 2009; 33(2): 130-135.

Baum BS, Schnall BL, Tis JE, Lipton JS. Correlation of residual limb length and gait parameters in amputees. Int. J. Care Injured 2008; 39: 728-733.

Begalle RL, Walsh MC, McGrath ML, Boling MC, Blackburn JT, Padua DA. Ankle dorsiflexion displacement during landing is associated with initial contact kinematics but not joint displacement. J Appl Biomech 2015.

Bell JC, Wolf EJ, Schnall BL, Tis JE, Potter BK. Transfemoral amputations: is there an effect of residual limb length and orientation on energy expenditure? Clin Orthop Relat Res 2014; 472: 3055-3061.

Bell JC, Wolf EJ, Schnall BL, Tis JE, Tis LL, Potter BK. Transfemoral amputations: the effect of residual limb length and orientation on gait analysis outcome measures. J Bone Joint Surg Am 2013; 95: 408-414.

Benichou C, Wirotius JM. Articuluar cartilage atrophy in lower limb amputees. Arthritis Rheum 1982; 25: 80-82.

Berger N. Analysis of amputee gait. In Bowker JH, Michael JW (Eds.), Atlas of limb prosthetics: Surgical, prosthetic, and rehabilitation principles (371-379). St. Louis: Mosby-Year Book, Inc 1992.

Bergmark A. Stability of the lumbar spine: a study in mechanical engineering. Act Orthop Scand 1989; 60: 1-54.

Bhangu S, Devlin M, Pauley T. Outcomes of individuals with transferral and contralateral transtibial amputation due to dysvascular etiologies. 2009; Prosthet Orthot Int. 33: 33-40.

Bjerke J, Öhberg F, Nilsson KG, Stensdotter AK. Compensatory strategies for muscle weakness during stair ascent in subjects with total knee arthroplasty. J Arthoplasty 2014; 29: 1499-1502.

Boonstra AM, Schrama J, Fidler V, Eisma WH. The gait of unilateral transfemoral amputees. Scand J Rehab Med 1994; 26(4): 217-223.

Bouillon L, Wilhelm J, Eisel P, Wiesner J, Rachow M, Hatteberg L. Electromyographic assessment of muscle activity between genders during unilateral weight-bearing tasks using adjusted distances. Int J Sports Phys Ther 2012; 7: 595-605.

Bowker JH, Michael JW (Eds.). Atlas of limb prosthetics: Surgical, prosthetic, and rehabilitation principles. St. Louis: Mosby-Year Book, Inc 1992.

Brereton LC, McGill SM. Frequency response of spine extensors during rapid isometric contractions: effects of muscle length and tension. J Electromyogr Kinesiol 1998; 8: 227-232.

Brown SH, McGill SM. How the inherent stiffness of the in vivo human trunk varies with changing magnitudes of muscular activation. Clin Biomech 2008; 23: 15-22.

Brown SH, McGill SM. Muscle force-stiffness characteristics influence joint stability: a spine example. Clin Biomech 2005; 20: 917-922.

Brown SH, McGill SM. The intrinsic stiffness of the in vivo lumbar spine in response to quick releases: implications for reflexive requirements. J Electromyogr Kinesiol 2009; 19: 727-736.

Buckley JG, De Asha AR, Johnson L, Beggs CB. Understanding adaptive gait in lower-limb amputees: insights from multivariate analyses. J Neuroeng Rehabil 2013; 10: 98.

Burke M, Roman V, Wright V. Bone and joint changes in lower limb amputees. Ann Rheum Dis 1978; 37: 252–254.

Burnfield JM, Eberly VJ, Gronely JK, Perry J, Yule WJ, Mulroy SJ. Impact of stance phase microprocessor-controlled knee prosthesis on ramp negotiation and community walking function in K2 level transfemoral amputees. Prosthet Orthot Int 2012; 36:95-104.

Butler K, Bowen C, Hughes AM, Torah R, Ayala I, Tudor J, Metcalf CD. A systematic review of the key factors affecting tissue viability and rehabilitation outcomes of the residual limb in lower extremity traumatic amputees. J Tissue Viability 2014; 23: 81-93.

Butler RJ, Crowell HP 3rd, Davis IM. Lower extremity stiffness: implications for performance and injury. Clin Biomech 2003; 18: 511-517.

Callaghan JP, Patla AE, McGill SM. Low back three-dimensional joint forces, kinematics, and kinetics during walking. Clin Biomech 1999; 14: 203-216.

Caron RR, Lewis CL, Saltzman E, Wagenaar RC, Holt KG. Musculoskeletal stiffness changes linearly in response to increasing load during walking gait. J Biomech 2015; 48: 1165-1171.

Chen G, Patten C, Kothari DH, Zajac FE. Gait differences between individuals with post-stroke hemiparesis and non-disabled controls at matched speeds. Gait Posture 2005; 22: 51-56.

Chiang J, Potvin JR. The *in vivo* dynamic response of the human spine to rapid lateral bend peturbation: effects of preload and step input magnitude. Spine 2001; 26: 1457-1464.

Cholewicki J, Juluru K, McGill SM. Intra-abdominal pressure mechanism for stabilizing the lumbar spine. J Biomech 1999; 32: 13-17.

Cholewicki J, McGill SM. Mechanical stability of the *in vivo* lumbar spine: implications for injury and chronic low back pain. Clin Biomech 1996; 11: 1-15.

Cholewicki J, McGill SM. Relationship between muscle force and stiffness in the whole mammalian muscle: a simulation study. J Biomech Eng 1995; 117: 339-342.

Comerford MJ, Mottram SL. Functional stability re-training: principles and strategies for managing mechanical dysfuntion. Man Ther 2001; 6: 3-14.

Cort JA, Dickey JP, Potvin JR. Trunk muscle contributions of to L4-5 joint rotational stiffness following sudden trunk lateral bend perturbations. J Electromyogr Kinesiol 2013; 23: 1334-1342.

Cowan SM, Crossley KM. Does gender influence neuromotor control of the knee and hip? J Electromyogr Kinesiol 2009; 19: 276-282.

Crisco JJ, Panjabi MM. The intersegmental and multisegmental muscles of the lumbar spine. A biomechanical model comparing lateral stabilizing potential. Spine 1991; 16(7): 793-799.

Cromwell R, Schultz AB, Beck R, Warwick D. Loads on the lumbar trunk during level walking. J Orthop Res 1989; 7: 1-7.

Dall PM, Kerr A. Frequency of the sit to stand task: An observational study of free-living adults. App Erg 2010; 41:58-61.

Dankaerts W, O'Sullivan P, Burnett A, Straker L. Altered patterns of superficial trunk muscle activation during sitting in nonspecific chronic low back pain patients: importance of subclassification. Spine 2006; 31(17): 2017-2023.

Davis KG, Marras WS. The effects of motion on trunk biomechanics. Clin Biomech 2000; 15: 703-717.

Devan H, Tumilty S, Smith C. Physical activity and lower-back pain in persons with traumatic transfemoral amputation: A national crosssectional survey. J Rehab Res Dev 2012; 49(10):1457-1466.

Dillingham TR, Pezzln LE, MacKenzie V. Limb amputation and limb deficiency: Epidemiology and recent trends in the United States. South Med J 2002; 95: 875-883.

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

Drake JDM, Fischer SL, Brown SHM, Callaghan JP. Do exercise balls provide a training advantage for trunk extensor exercises? A biomechanical evaluation. J Manipul Phys Therapeut 2006; 29(5): 354-362.

Dwyer MK, Boudreau SN, Mattacola CG, Uhl TL, Lattermann C. Comparison of lower extremity kinematics and hip muscle activation during rehabilitation tasks between sexes. J Athl Train 2010; 45: 181-190.

Feipel V, De Mesmaeker T, Klein P, Rooze M. Three-dimensional kinematics of the lumbar spine during treadmill walking at different speeds. Eur Spine J 2001; 10:16-22.

Gailey RS, Clark CR. Physical therapy management of adult lower limb amputees. In Bowker JH, Michael JW (Eds.), Atlas of limb prosthetics: Surgical, prosthetic, and rehabilitation principles. St. Louis: Mosby-Year Book, Inc 1992.

Gailey RS. Roach KE, Applegate EB, Cho B, Cunniffe B, Licht S, Maguire M, Nash MS. The amputee mobility predictor: an instrument to assess determinants of the lower-limb amputee's ability to ambulate. Arch Phys Med Rehabil 2002; 83: 613-627.

Gao F, Zhang F, Huang H. Investigation of sit-to-stand and stand-to-sit in an above knee amputee. Conf Proc IEEE Eng Med Biol Soc 2011; 7340-7343.

Gardner-Morse MG, Stokes IA. Trunk stiffness increases with steady-state effort. J Biomech 2001; 34: 457-463.

Gaunaurd I, Gailey R, Hafner BJ, Gomez-Marin O, Kirk-Sanchez N. Postural asymmetries in transfemoral amputees. Prosthet Orthot Int 2011; 35:171-180.

Ghamkhar L, Kahlaee AH. Trunk muscles activation pattern during walking in subjects with and without chronic low back pain: a systematic review. PM R 2015.

Goh JC, Toh SL, Bose K. Biomechanical study on axillary crutches during single-leg swing through gait. Prosthet Orthot Int 1986; 10: 89-95.

Gottschalk F. Transfemoral amputation: Surgical procedures. In Bowker JH, Michael JW (Eds.), Atlas of limb prosthetics: Surgical, prosthetic, and rehabilitation principles (501-507). St. Louis: Mosby-Year Book, Inc 1992.

Goujon-Pillet H, Sapin E, Fodé P, Lavaste F. Three-dimensional motions of trunk and pelvis during transfemoral amputee gait. Arch Phys Med Rehabil 2008; 89: 87-94.

Graham RB, Brown SHM. Local dynamic stability of spine muscle activation and stiffness patterns during repetitive lifting. J Biomech Eng 2014; 136: 121006.

Granata KP, England SA. Stability of dynamic trunk movement. Spine 2006; 31: 271-276.

Granata KP, Orishimo KF. Response of trunk muscle coactivation to changes in spinal stability. J Biomech 2001; 34: 1117-1123.

Grenier SG, McGill SM. Quantification of lumbar stability by using 2 different abdominal activation strategies. Arch Phys Med Rehabil 2007; 88: 54-62.

Grieve AC, Lankhorst GJ. Functional outcome of lower-limb amputees: a prospective descriptive study in a general hospital. Prosthet Orthot Int. 1996; 20: 79-87.

Hafner BJ, Willingham LL, Buell NC, Allyn KJ, Smith DG. Evaluation of function, performance, and preference as transfemoral amputees transition from mechanical to microprocessor control of the prosthetic knee. Arch Phys Med Rehabil 2007; 88: 207-217.

Hamamura S, Chin T, Kuroda R, Akisue T, Iguchi T, Kohono H, Kitagawa A, Tsumura N, Kurosaka M. Factors affecting prosthetic rehabilitation outcomes in amputees of age 60 years and over. J Int Med Res 2009; 37: 1921-1927.

Hamamura S, et al. Factors affecting prosthetic rehabilitation outcomes in amputees of age 60 years and over. J Intern Med Res 2009; 37: 1921-1927.

Harness N, Pinzur MS. Health related quality of life in patients with dysvascular transtibial amputation. Clin Orthop Relat Res 2001; 383:204–207.

Hebert JS, Payne MWC, Wolfe DL, Deathe AB, Devlin M. Comorbidities in amputation: a systematic review of hemiplegia and lower limb amputation. Disabil Rehabil 2012; 34: 1943-1949.

Hendershot BD, Wolf EJ. Three-dimensional joint reaction forces and moments at the low back during over-ground walking in persons with unilateral lower-extremity amputation. Clin Biomech 2014; 29: 235-242.

Hermens HJ, Freriks B, Disselhorst-Klug C, Rauu G. Development of recommendations for SEMG sensors and sensor placement procedures. J Electromyogr Kinesiol 2000; 10:361-374.

Highsmith MJ, et al. Kinetic asymmetry in transfemoral amputees while performing sit to stand and stand to sit movements. Gait Posture 2011; 34:86-91.

Highsmith MJ, Lura DJ, Carey SL, Mengelkoch LJ, Kim SH, Quillen WS, Kahle JT, Miro RM. Correlations between residual limb length and joint moments during sitting and standing movements in transfemoral amputees. Prosthet Orthot Int. 2015.

Himmelreich H, Vogt L, Banzer W. Gluteal muscle recruitment during level, incline and stair ambulation in healthy subjects and chronic low back pain patients. J back Musculoskel rehab 2008; 21:193-199.

Hobara H, Kobayashi Y, Nakamura T, Yamasaki N, Ogata T. Foot clearance strategy for step-over-step stair climbing in transfemoral amputees. Prostate Orthot Int 2014; 38: 332-335.

Hobara H, Kobayashi Y, Tominaga S, Nakamura T, Yamasaki N, Ogata T. Factors affecting stair-ascent patterns in unilateral transfemoral amputees. Prosthet Orthot Int. 2012; 37: 222-226.

Hodges PW, Moseley GL. Pain and motor control of the lumbopelvic region: the effect and possible mechanisms. J Electromyogr Kinesiol 2003; 13: 361-370.

Hong JH, Mun MS. Relationship between socket pressure and EMG of two muscles in trans-femoral stumps during gait. Prosthet Orthot Int. 2005; 29: 59-72.

Ip, D. Orthopedic rehabilitation, assessment, and enablement. Berlin; New York: Springer 2007.

Ivanenko YP, Wright WG, St George RJ, Gurfinkel VS. Trunk Orientation, stability, and quadrupedalism. Front Neurol 2013; 4: 1-6.

Jaegers SMHJ, Hans Arendzen J, de Jongh HJ. Prosthetic gait of unilateral transfemoral emputees: A kinematic study. Arch Phys Med Rehab 1995; 76:736-743.

Jones SF, Twigg PC, Scally AJ, Buckley JG. The mechanics of landing when stepping down in unilateral lower-limb amputees. Clin Biomech 2006; 21: 184-193.

Kahle JT, Highsmith MJ, Hubbard SL. Comparison of nonmicroprocessor knee mechanism versus C-Leg on prosthesis evaluation questionnaire, stumbles, falls, walking tests, stair descent, and knee preference. J Rehab Research Develop 2008; 45(1):1-14.

Kapp S, Cummings D. Transtibial amputation: Prosthetic management. In Bowker JH, Michael JW (Eds.), Atlas of limb prosthetics: Surgical, prosthetic, and rehabilitation principles (453-478). St. Louis: Mosby-Year Book, Inc 1992.

Karacoloff LA, Hammersley CS, Schneider FJ. Lower extremity amputation: a guide to functional outcomes in phusical therepy management. Washington; National Rehabilitation Hospital 1992.

Kaufman KR, Frittoli S, Frigo CA. Gait asymmetry of transfemoral amputees using mechanical and microprocessor-controlled prosthetic knees. Clinic Biomech 2012; 27:460-465.

Kavcic N, Grenier S, McGill SM. Determining the stabilizing role of individual torso muscles during rehabilitation exercises. Spine 2004; 29: 1254-1265.

Keun Lee J, Desmoulin GT, Khan AH, Park EJ. Comparison of 3D spinal motions during stair-climbing between individuals with and without low back pain. Gait & Posture 2011; 34: 222-226.

Krebs DE, Wong D, Jevsevar D, Riley PO, Hodge WA. Trunk kinematics during locomotor activities. Phys Ther 1992; 72: 505-514.

Kulkarni J, Gaine WJ, Buckley JG, Rankine JJ, Adams J.Chronic low back pain traumatic lower limb amputees. Clin Rehabil 2005; 19: 81-86.

Kumar S. Theories of musculoskeletal injury causation. Ergonomics 2001; 15: 17-47.

Lamoth CJ, Ainsworth E, Polomski W, Houdijk H. Variability and stability analysis of walking of transfemoral amputees. Med Eng Phys 2010; 32: 1009-1014.

Latimer J, Lee M, Adams R. The effect of training with feedback on physiotherapy students' ability judge lumbar stiffness. Man Ther 1996; 1: 266-270.

Lawson BE, Varol HA, Goldfarb M. Standing stability enhancement with an intelligent powered transfemoral prosthesis. IEEE Trans Biomed Eng 2011; 58: 2617-2624.

Lee B, McGill S. The effect of long term isometric training on core/torso stiffness. J Strength Cond Res 2015.

Lee JK, Desmoulin GT, Khan AH, Park EJ. Comparison of 3D spinal motions during stair-climbing between individuals with and without low back pain. Gait Posture 2011; 34: 222-236.

Lee PJ, Rogers EL, Granata KP. Active trunk stiffness increases with co-contraction. J Electromyogr Kinesiol 2006; 16: 51-57.

Li K, Ackland DC, McClelland JA, Webster KE, Feller JA, de Steiger R, Pandy MG. Trunk muscle action compensates for reduced quadriceps force during walking after total knee arthroplasty. Gait Posture 2013; 38: 79-85.

Lindemann U, Muche R, Stuber M, Zijlstra W, Hauer K, Becker C. Coordination of strength exertion during the chair-rise movment in very old people. J Gerontolog 2007; 62A(6):636-640.

Liu K, Heise GD. The effect of jump-landing directions on dynamic stability. J Appl Biomech 2013; 29: 634-638.

Lloyd CH, Stanhope SJ, Davis IS, Royer TD. Strength asymmetry and osteoarthritis risk factors in unilateral trans-tibial, amputee gait. Gait Posture 2010; 32: 296-300.

Madokoro S, Miaki H, Yamazaki T. The effect of the abdominal drawing-in manoeuvre during forward steps. J Phys Ther Sci 2014; 26: 889-893.

Marras WS, Sommerich CM. A three-dimensional motion model of loads on the lumbar spine: II. Model validation. Hum Factors 1991; 33(2): 139-149.

McDonough AL, Razza-Doherty M. Some biomechanical aspects of crutch and cane walking: the relationship between forward rate of progression, symmetry, and efficiency--a case report. Clin Podiatr Med Surg 1988; 5(3): 677-693.

McFadyen BJ, Winter DA. An integrated biomechanical analysis of normal stair ascent and descent. J Biomech 1988; 21(9):733-744.

McGill SM, McDermott A, Fenwick CM. Comparison of different strongman events: trunk muscle activation and lumbar spine motion, load, and stiffness. J Strength Cond Res 2009; 23: 1148-1161.

McGill SM. Electromyographic activity of the abdominal and low back musculature during the generation of isometric and dynamic axial trunk torque: implications for lumbar mechanics. J Orthop Res 1991; 9(1):91-103.

McGill SM. Low Back Disorders: Evidence-Based Prevention and Rehabilitation. Human Kinetics 2007.

McGill SM. Ultimate back fitness and performance. Back Fit Pro, 2014.

Melchiorre PJ, Findley T, Boda W. Functional outcome and comorbidity indexes in the rehabilitation of the traumatic versus the vascular unilateral lower limb amputee. Am J Phys Med Rehabil 1996; 75: 9-14.

Mensch G, Ellis PM. Physical therapy management of lower extremity amputations. Rockville, Maryland: Aspen Publishers 1986.

Michael JW. Modern prosthetic knee mechanisms. Clinic Orthop Relat Res 1999; 361:39-47.

Michaud SB, Gard SA, Childress DS. A preliminary investigation of pelvic obliquity patterns during gait in persons with transtibial and transfemoral amputation. J Rehabil Res Dev 2000; 37: 1-10.

Miura T, Yamanaka M, Ukishiro K, Tohyama H, Saito H, Samukawa M, Kobayashi T, Ino T,

Takeda N. Individuals with chronic low back pain do not modulate the level of transversus abdominis muscle contraction across different postures. Man Ther 2014; 19: 534-540.

Nadollek H, Brauer S, Isles R. Outcomes after trans-tibial amputation: the relationship between quiet stance ability, strength of hip abductor muscles and gait. Physiother Res Int 2007; 7: 203-214.

Nelson-Wong E, Callaghan JP. Is muscle co-activation a predisposing factor for low back pain development during standing? A multifactorial approach for early identification of at-risk individuals. J Electromyogr Kinesiol 2010; 20: 256-263.

Neptune RR, Kautz SA, Zajac FE. Contributions of the individual ankle plantar flexors to support, forward progression and swing initiation during walking. J Biomech 2001; 34: 1387-1389.

Nolan L, Lees A. The functional demands on the intact limb during walking for active transfemoral and transtibial amputees. Prosthet Orthot Int 2000; 24:117-125.

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

Nuzik S, Lamb R, VanSant A, Hirt S. Sit-to-stand movment pattern: A kinematic study. Phys Therap 1986; 66:1708-1713.

O’Connor CM, Thorpe SK, O’Malley MJ, Vaughan CL. Automatic detection of gait events using kinematic data. Gait Posture 2007; 25:469-474.

O’Sullivan PB, et al. Effect of different upright sitting postures on spinal-pelvic curvature and trunk muscle activation in a pain-free population. Spine 2006; 31(19): 707-712.

Panjabi MM. The stabilizing system of the spine. Part I. Function, dysfunction, adaptation, and enhancement. J Spinal Discord 1992; 5: 383-389.

Pantall A, Durham S, Ewins D. Surface electromyographic activity of five residual limb muscles recorded during isometric contraction in transfemoral amputees with osseointegrated prostheses. Clin Biomech 2011; 26: 760-765.

Pauley T, Devlin M, Madan-Sharma P. A single-blind, cross-over trial of hip abductor strength training to improve timed up & go performance in patients with unilateral, transfemoral amputation. J Rehabil Med 2014; 46: 264-270.

Pitkin MR. Biomechanics of lower limb prosthetics. London; New York: Springer-Verlag Berlin Heidelberg 2010.

Potvin JR, Brown SH. An equation to calculate individual muscle contributions to joint stability. J Biomech 2005; 38: 973-980.

Protopapadaki A, Drechsler WI, Cramp MC, Coutts FJ, Scott OM. Hip, knee, ankle kinematics and kinetics during stair ascent and descent in healthy young individuals. Clinic Miomech 2007; 22:203-210.

Puntumetakul R, Areeudomwong P, Emasithi A, Yamaguchi J. Effect of 10-week core stabilization exercise training and detraining on pain-related outcomes in patients with clinical lumbar instability. Patient Prefer Adherence 2013; 7: 1189-1199.

Raya MA, Gailey RS, Fiebert IM, Roach KE. Impairment variables predicting activity limitation in individuals with lower limb amputation. Prosthet Orthot Int. 2010; 34: 73-84.

Richardson CA, Snijders CJ, Hides JA, Damen L, Pas MS, Storm J. The relation between the transversus abdominis muscles, sacroiliac joint mechanics, and low back pain. Spine 2002; 27: 399-405.

Riener R, Rabuffetti M, Frigo C. Stair ascent and descent at different inclinations. Gait Posture 2002; 15:32-44.

Riley PO, Schenkman ML, Mann RW, Hodge WA. Mechanics of a constrained chair-rise. J Biomech 1991; 24: 77-85.

Roebroeck ME, Doorenbosch CAM, Harlaar JM, Jacobs R, Lankhorst GJ. Biomechanics and muscular activity during sit-to-stand transfer. Clinic Biomech 1994; 9:235-244.

Ross GB, Mavor M, Brown SH, Graham RB. The effects of experimentally induced low back pain on spine rotational stiffness and local dynamic stability. Ann Biomen Eng 2015.

Rowley KM, Richards JG. Increasing plantarflexion angle during landing reduces vertical ground reaction forces, loading rates and the hip’s contribution to support moment within participants. J Sports Sci 2015; 16:1-10.

Rueda FM, Diego IM, Sánchez AM, Tejada MC, Montero FM, Page JC. Knee and hip internal moments and upper-body kinematics in the frontal plane in unilateral transtibial amputees. Gait Posture 2013; 37: 436-439.

Ryser DK, Erickson RP, Cahalan T. Isometric and isokinetic hip abductor strength in persons with above-knee amputations. Arch Phus Med Rehabil 1988; 69(10): 840-845.

Sapin E, Goujon H, de Almeida F, Fode P, Lavaste F. Functional gait analysis of transfemoral amputees using two different single-axis prosthetic knees with hydraulic swing-phase control: Kinematic and kinetic comparison of two prosthetic knees. Prosthet Orthot Int 2008; 32:201-218.

Schaarschmidt M, Lipfert SW, Meier-Gratz C, Scholle H-C, Seufarth A. Functional gait asymmetry of unilateral transfemoral amputees. Human Movem Sci 2012; 31: 907-917.

Schinel-Ivy A, Nairn BC, Drake JD. Investigation of trunk muscle co-contraction and its association with low back pain development during prolonged sitting. J Electromyogr Kinesiol 2013; 23: 778-786.

Schmalz T, Blumentritt S, Marx B. Biomechanical analysis of stair ambulation in lower limb amputees. Gait Posture 2007; 25:267-278.

Schofield JS, Parent EC, Lewicke J, Carey JP, El-Rich M, Adeeb S. Characterizing asymmetry across the whole sit to stand movement in healthy participants. J Biomech 2013; 46:2730-2735.

Schoppen T, Boonstra A, Groothoff JW, de Vries J, Goeken LN, Eisma WH. Physical, mental, and social predictors of functional outcome in unilateral lower-limb amputees. Arch Phys Med Rehabil 2003; 84: 803-811.

Schuch CM. Transfemoral amputation: Prosthetic management. In Bowker JH, Michael JW (Eds.), Atlas of limb prosthetics: Surgical, prosthetic, and rehabilitation principles (509-533). St. Louis: Mosby-Year Book, Inc 1992.

Segal AD, et al. Kinematic and kinetic comparisons of transfemoral amputee gait using C-Leg and Mauch SNS prosthetic knees. J Rehab Research Developm 2006; 43(7):857-870.

Selles RW, Wagenaar RC, Smit TH, Wuisman PI. Disorders in trunk rotation during walking in patients with low back pain: a dynamical systems approach. Clin Biomech 2001; 16: 175-181.

Seroussi RE, Gitter A, Czerniecki JM, Weaver K. Mechanical work adaptations of above-knee amputee ambulation. Arch Phys Med Rehab 1996; 77:1209-1214.

Shirley D, Hodges PW, Eriksson AE, Gandevia SC. Spinal stiffness changes throughout the respiratory cycle. J Appl Physical 2003; 95: 1467-1475.

Springer BA, Gill NW. Characterization of lateral abdominal muscle thickness in persons with lower extremity amputations. J Orthopaed Sport Phys Therap 2007; 37(10):635-643.

Stanhope VA, Knarr BA, Reisman DS, Higginson JS. Frontal plane compensatory strategies associated with self-selected walking speed in individuals post-stroke. Clin Biomech 2014; 29: 518-522.

Stokes IA, Gardner-Morse MG, Henry SM. Abdominal muscle activation increases lumbar spinal stability: analysis of contributions of different muscle groups. Clin Biomech 2011; 26: 797-803.

Stokes IA, Gardner-Morse MG. Spinal stiffness increases with axial load: another stabilizing consequence of muscle action. J Electromyogr Kinesiol 2003; 13: 397-402.

Struyf PA, van Heugten CM, Hitters MW, Smeets RJ. The prevalence of osteoarthritis of the intact hip and knee among traumatic leg amputees. Arch Phys Med Rehabil 2009; 90: 440-446.

Tesio L, Lanzi D, Detrembleur C. The 3D motion of the centre of gravity of the human body during level walking. II. Lower limb amputees. Clinic Biomech 1998; 13(2):83-90.

Thorstensson A, Carlson H, Zomlefer MR, Nilsson J. Lumbar back muscle activity in relation to trunk movements during locomotion in man. Act Physiol Scand 1982; 116: 13-20.

Travell JG, Simons DG. Myofascial pain and dysfunction: the trigger point manual; the lower extremities. Philadelphia, PA: Lippincott Williams & Wilkins 1992.

Tura A, Raggi M, Rocchi L, Cutti AG, Chiari L. Gait symmetry and regularity in transfemoral amputees assessed by trunk accelerations. J Neuroeng Rehabil 2010; 7: 1-10.

van Dieën JH, Kingma I, van der Bug P. Evidence for a role of antagonistic cocontraction in controlling trunk stiffness during lifting. J Biomech 2003; 36: 1829-1836.

van Sint Jan S. Color atlas of skeletal landmark definitions: Guidelines for reproducible manual and virtual palpations. Edinburgh; New York: Churchill Livingstone/Elsevier 2007.

van Soest AJ, Haenen WP, Rozendaal LA. Stability of bipedal stance: the contribution of cocontraction and spindle feedback. Biol. Cybern. 2003; 88: 293-301.

Vera-Garcia FJ, Brown SH, Gray JR, McGill SM. Effects of different levels of torso coactivation on trunk muscular and kinematic responses to posteriorly applied sudden loads. Clin Biomech 2006; 21: 443-455.

Wentink EC, Prinsen EC, Rietman JS, Veltink PH. Comparison of muscle activity patterns of transfemoral amputees and control subjects during walking. J NeuroEng Rehab 2013; 10:87-97.

Winter DA. Energy generation and absorption at the ankle and knee during fast, natural, and slow cadences. Clin Orthop Relat Res 1983; May (175): 147-154.

Winter DA. The biomechanics and motor control of human gait: Normal, elderly and pathological. Waterloo: University of Waterloo Press 1991.

Wolf EJ, Everding VQ, Linberg AL, Schnall BL, Czerniecki JM, Gambel JM. Assessment of transfemoral amputees using C-Leg and Power Knee for ascending and descending inclines and steps. J Rehabil Res Dev 2012; 49: 831-842.

Yoder AJ, Petrella AJ, Silverman AK. Trunk-pelvis motion, joint loads, and muscle forces during walking with a transtibial amputation. Gait Posture 2015; 41: 757-762.

Zahalak GI. Modeling muscle mechanics (and energetics). In Winter JM, Woo, SL-Y (Eds.), Multiple muscle systems. New York: Springer 1990.

APPENDIX A – Oswestry Low Back Pain Disability Questionnaire

Scoring instructions

For each section the total possible score is 5. If all 10 sections are completed the score is calculated as follows:

Example: 16 (total scored)
50 (total possible score) x 100 = 32%

If one section is missed or not applicable the score is calculated:
16 (total scored)
45 (total possible score) x 100 = 35.5%

Minimum detectable change (90% confidence): 10% points (change of less than this may be attributable to error in the measurement)

Interpretation of scores

0% to 20%: minimal disability:	The patient can cope with most living activities. Usually no treatment is indicated apart from advice on lifting sitting and exercise.
21%-40%: moderate disability:	The patient experiences more pain and difficulty with sitting, lifting and standing. Travel and social life are more difficult and they may be disabled from work. Personal care, sexual activity and sleeping are not grossly affected and the patient can usually be managed by conservative means.
41%-60%: severe disability:	Pain remains the main problem in this group but activities of daily living are affected. These patients require a detailed investigation.
61%-80%: crippled:	Back pain impinges on all aspects of the patient's life. Positive intervention is required.
81%-100%:	These patients are either bed-bound or exaggerating their symptoms.

Section 1 – Pain intensity
0 – I have no pain at the moment
1 – The pain is very mild at the moment
2 – The pain is moderate at the moment
3 – The pain is fairly severe at the moment
4 – The pain is very severe at the moment
5 – The pain is the worst imaginable at the moment
Section 2 – Personal care (washing, dressing etc)
0 – I can look after myself normally without causing extra pain
1 – I can look after myself normally but it causes extra pain
2 – It is painful to look after myself and I am slow and careful
3 – I need some help but manage most of my personal care
4 – I need help every day in most aspects of self-care
5 – I do not get dressed, I wash with difficulty and stay in bed
Section 3 – Lifting
0 – I can lift heavy weights without extra pain

1 – I can lift heavy weights but it gives extra pain
2 – Pain prevents me from lifting heavy weights off the floor, but I can manage if they are conveniently placed eg. on a table
3 – Pain prevents me from lifting heavy weights, but I can manage light to medium weights if they are conveniently positioned
4 – I can lift very light weights
5 – I cannot lift or carry anything at all
Section 4 – Walking*
0 – Pain does not prevent me walking any distance
1 – Pain prevents me from walking more than 2 kilometres
2 – Pain prevents me from walking more than 1 kilometre
3 – Pain prevents me from walking more than 500 metres
4 – I can only walk using a stick or crutches
5 – I am in bed most of the time
Section 5 – Sitting
0 – I can sit in any chair as long as I like
1 – I can only sit in my favourite chair as long as I like

2 – Pain prevents me sitting more than one hour
3 – Pain prevents me from sitting more than 30 minutes
4 – Pain prevents me from sitting more than 10 minutes
5 – Pain prevents me from sitting at all
Section 6 – Standing
0 – I can stand as long as I want without extra pain
1 – I can stand as long as I want but it gives me extra pain
2 – Pain prevents me from standing for more than 1 hour
3 – Pain prevents me from standing for more than 3 minutes
4 – Pain prevents me from standing for more than 10 minutes
5 – Pain prevents me from standing at all
Section 7 – Sleeping
0 – My sleep is never disturbed by pain
1 – My sleep is occasionally disturbed by pain
2 – Because of pain I have less than 6 hours sleep
3 – Because of pain I have less than 4 hours sleep

4 – Because of pain I have less than 2 hours sleep
5 – Pain prevents me from sleeping at all
Section 8 – Sex life (if applicable)
0 – My sex life is normal and causes no extra pain
1 – My sex life is normal but causes some extra pain
2 – My sex life is nearly normal but is very painful
3 – My sex life is severely restricted by pain
4 – My sex life is nearly absent because of pain
5 – Pain prevents any sex life at all
Section 9 – Social life
0 – My social life is normal and gives me no extra pain
1 – My social life is normal but increases the degree of pain
2 – Pain has no significant effect on my social life apart from limiting my more energetic interests eg, sport
3 – Pain has restricted my social life and I do not go out as often
4 – Pain has restricted my social life to my home
5 – I have no social life because of pain

Section 10 – Travelling
0 – I can travel anywhere without pain
1 – I can travel anywhere but it gives me extra pain
2 – Pain is bad but I manage journeys over two hours
3 – Pain restricts me to journeys of less than one hour
4 – Pain restricts me to short necessary journeys under 30 minutes
5 – Pain prevents me from travelling except to receive treatment

APPENDIX B – Trunk stiffness model example equations

Total rotational trunk stiffness in 3 planes:

$$S_x = \sum_{m=1}^{14} F_m * \left[\frac{A_y B_y + A_z B_z - r_x^2}{l} + \frac{q * r_x^2}{L}\right]$$

$$S_y = \sum_{m=1}^{14} F_m * \left[\frac{A_x B_x + A_z B_z - r_y^2}{l} + \frac{q * r_y^2}{L}\right]$$

$$S_z = \sum_{m=1}^{14} F_m * \left[\frac{A_x B_x + A_y B_y - r_z^2}{l} + \frac{q * r_z^2}{L}\right]$$

S_x = total trunk stiffness in frontal plane (N*m/rad)
S_y = total trunk stiffness in transverse plane (N*m/rad)
S_z = total trunk stiffness in sagittal plane (N*m/rad)

$\langle A_x \quad A_y \quad A_z \rangle$ = relative muscle origin coordinates (Cholewicki & McGill, 1996)
$\langle B_x \quad B_y \quad B_z \rangle$ = relative muscle insertion/node coordinates (Cholewicki & McGill, 1996)
l = muscle length (m) from origin to insertion/node
L = muscle length (m) from origin to insertion
F = muscle force (N)
r = muscle moment arm (m)
q = dimensionless multiplier for muscle stiffness (Brown & McGill, 2005)

$$l_{rest} = \sqrt{(B_x - A_x)^2 + \left(B_y - A_y\right)^2 + (B_z - A_z)^2}$$

$$l_{2_{z-example}} = \sqrt{\left(B_x * \cos\theta_z - B_y * \sin\theta_z - A_x\right)^2 + \left(B_x * \sin\theta_z + B_y * \cos\theta_z - A_y\right)^2 + (B_z - A_z)^2}$$

$$r_{z-example} = \frac{B_x A_y - A_x B_y}{l}$$

$$F_m = PCSA_m * \delta * MVC_m * Lm * Gain$$

$$q = 6.4 - 3.6(x + 1)(x - 1)$$

θ = planar trunk angle of rotation from quiet stance (rad)
$PCSA_m$ = physiological cross-sectional area (m^2) (Cholewicki & McGill, 1996)
δ = max muscle stress (35 N/m2) (Reid & Costigan, 1987)
MVC_m = maximum voluntary contraction
Lm = dimensionless muscle length coefficient

Gain = dimensionless multiplier to balance net muscle moment from static trial
x = normalized muscle force, with 0.0 representing minimum muscle force, and 1.0 representing muscle force at maximal activation
NOTE: for calculations about the *x*-axis, substitute *x* for *z*, *y* for *x*, and *z* for *y*; about the *y*-axis, substitute *x* for *y*, *y* for *z*, and *z* for *x*.

L_r = resting muscle length (Potvin et al., 1996)

If $L>L_r$ …

$$Lm = \left(-1.092 * \frac{L}{L_r}\right) + 2.15$$

If $L<L_r$ …

$$Lm = sin\left[\pi\left(\frac{L}{L_r} - 0.5\right)\right]$$

APPENDIX C – Subject group comparisons of average 3-dimensional trunk stiffness

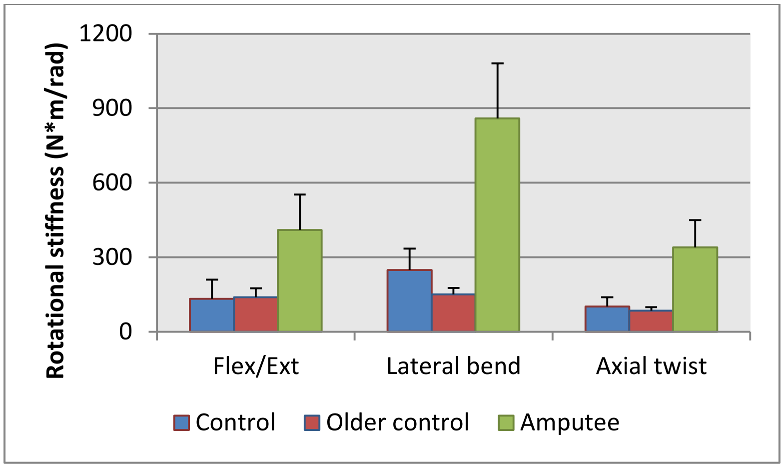

Figure C-1: Average rotational trunk stiffness in 3-dimensions for the young and old control groups, and the TFA group, during the gait task.

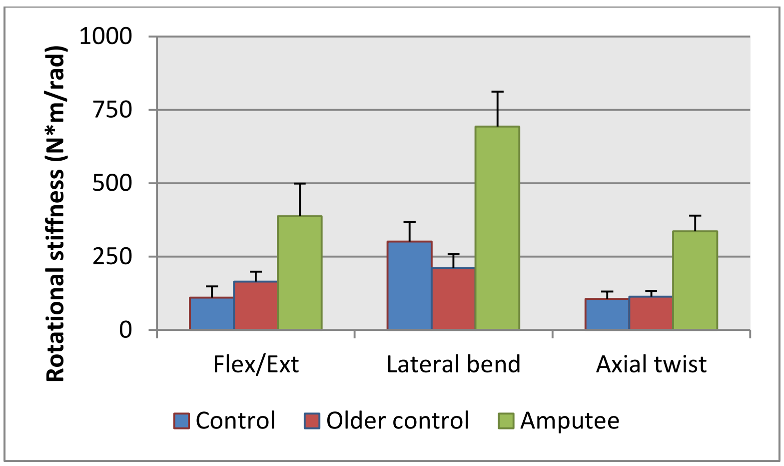

Figure C-2: Average rotational trunk stiffness in 3-dimensions for the young and old control groups, and the TFA group, during the step down task.

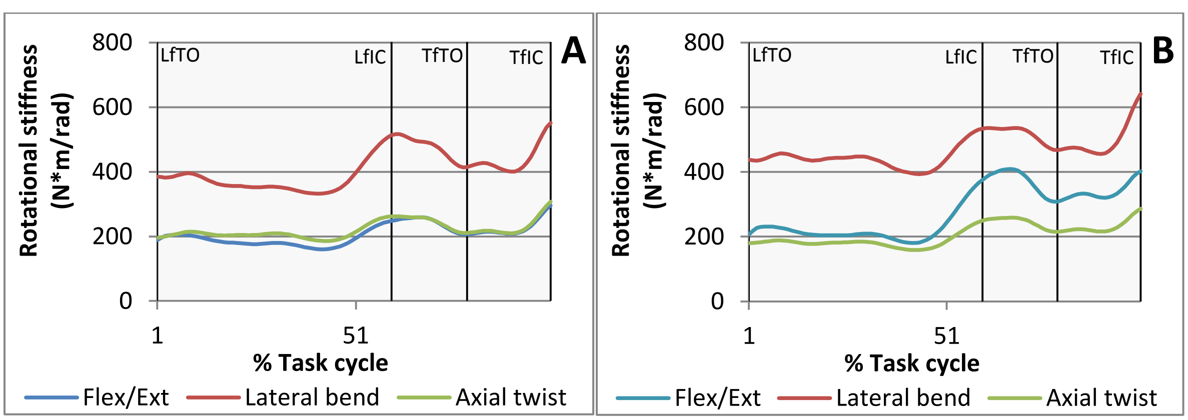

Figure C-3: Rotational stiffness patterns for TFA4 using the kinematic data from TFA3 (A) or leaving the model's trunk angle input at 0º in all dimensions (B).

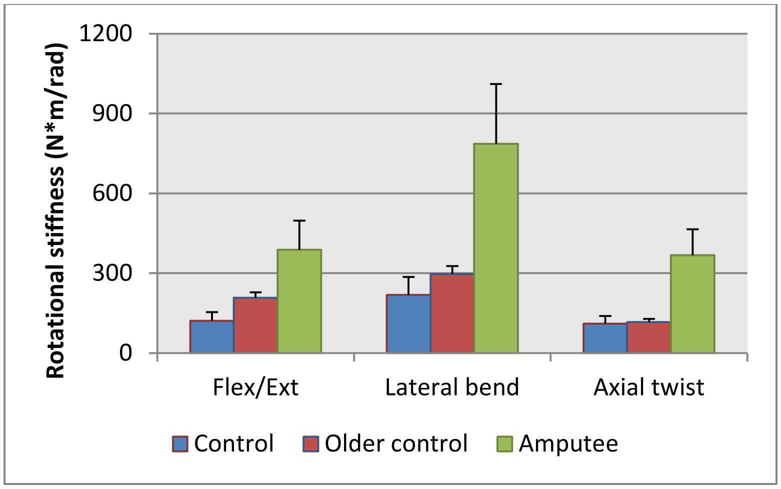

Figure C-4: Average rotational trunk stiffness in 3-dimensions for the young and old control groups, and the TFA group, during the step up task.

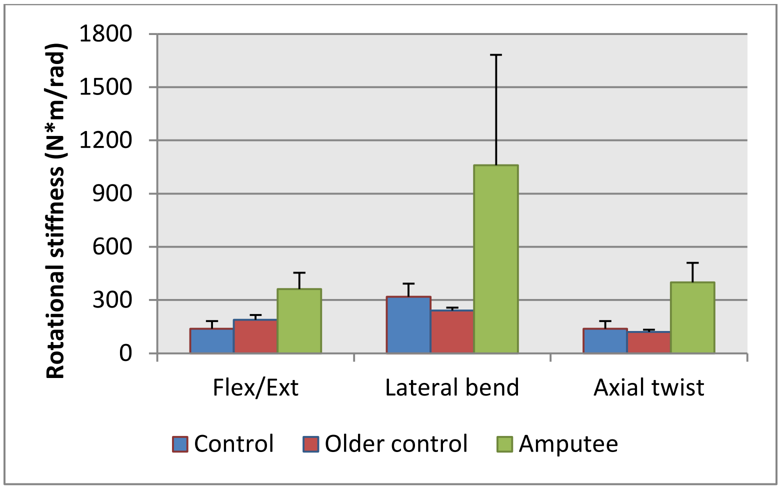

Figure C-5: Average rotational trunk stiffness in 3-dimensions for the young and old control groups, and the TFA group, during the sit-to-stand task.

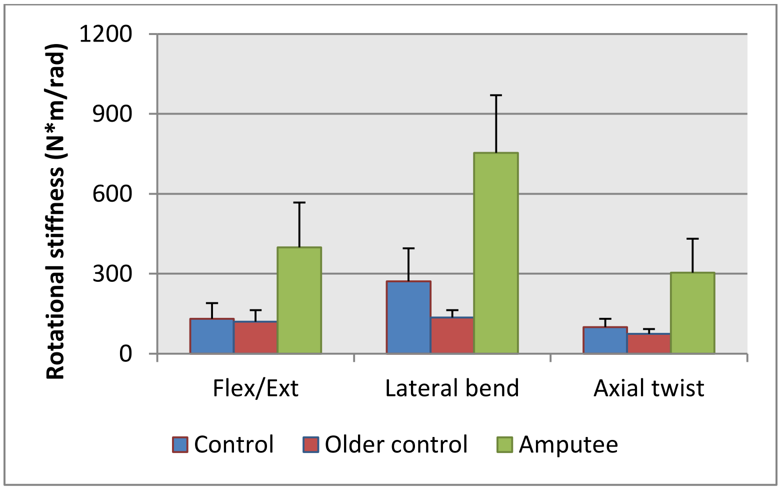

Figure C-6: Average rotational trunk stiffness in 3-dimensions for the young and old control groups, and the TFA group, during the door pull task.

APPENDIX D – Lower limb and trunk angles and EMG

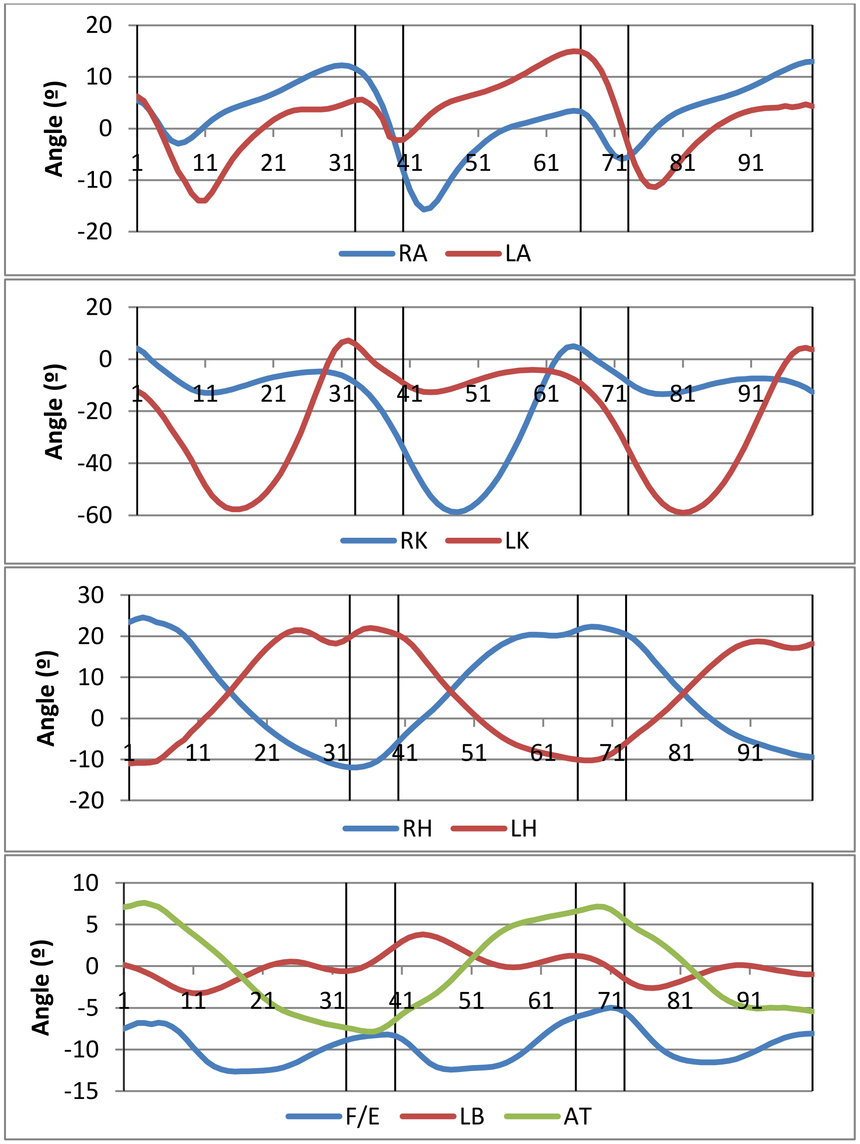

Figure D-1: Control group joint angles during the gait task.

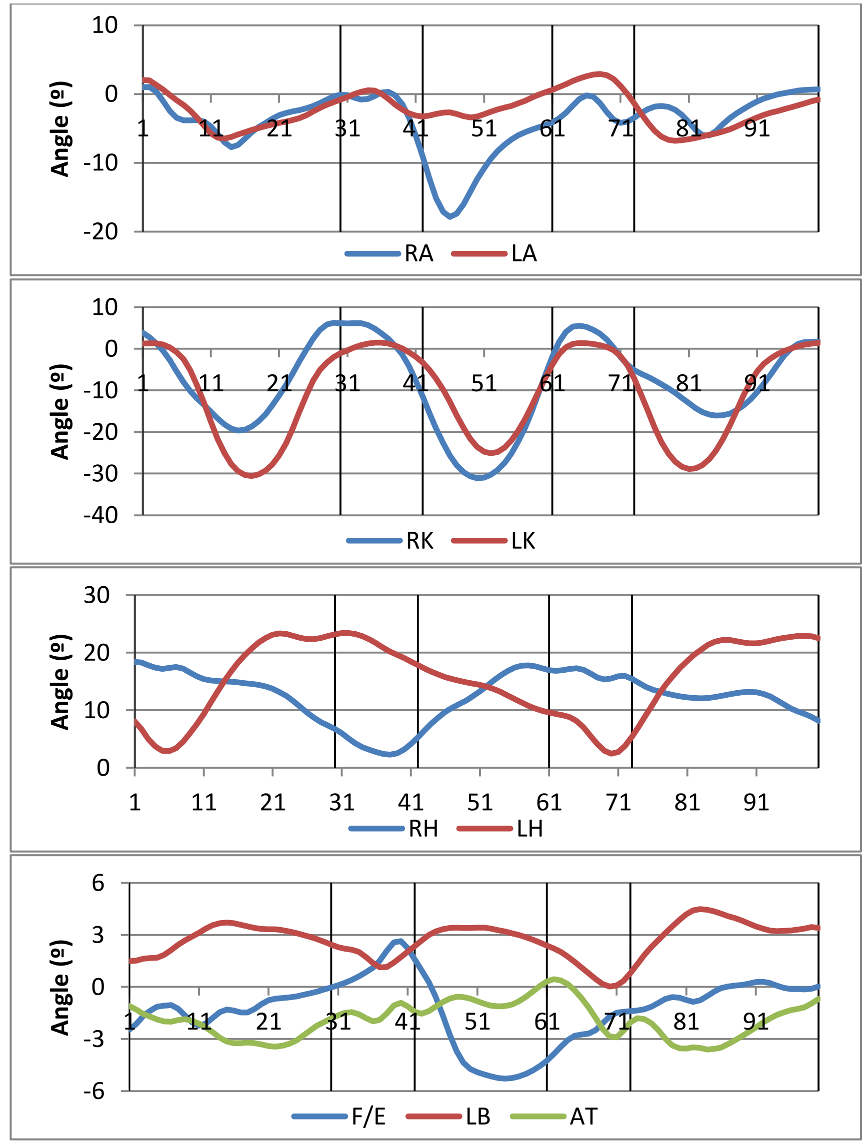

Figure D-2: Amputee group joint angles during the gait task.

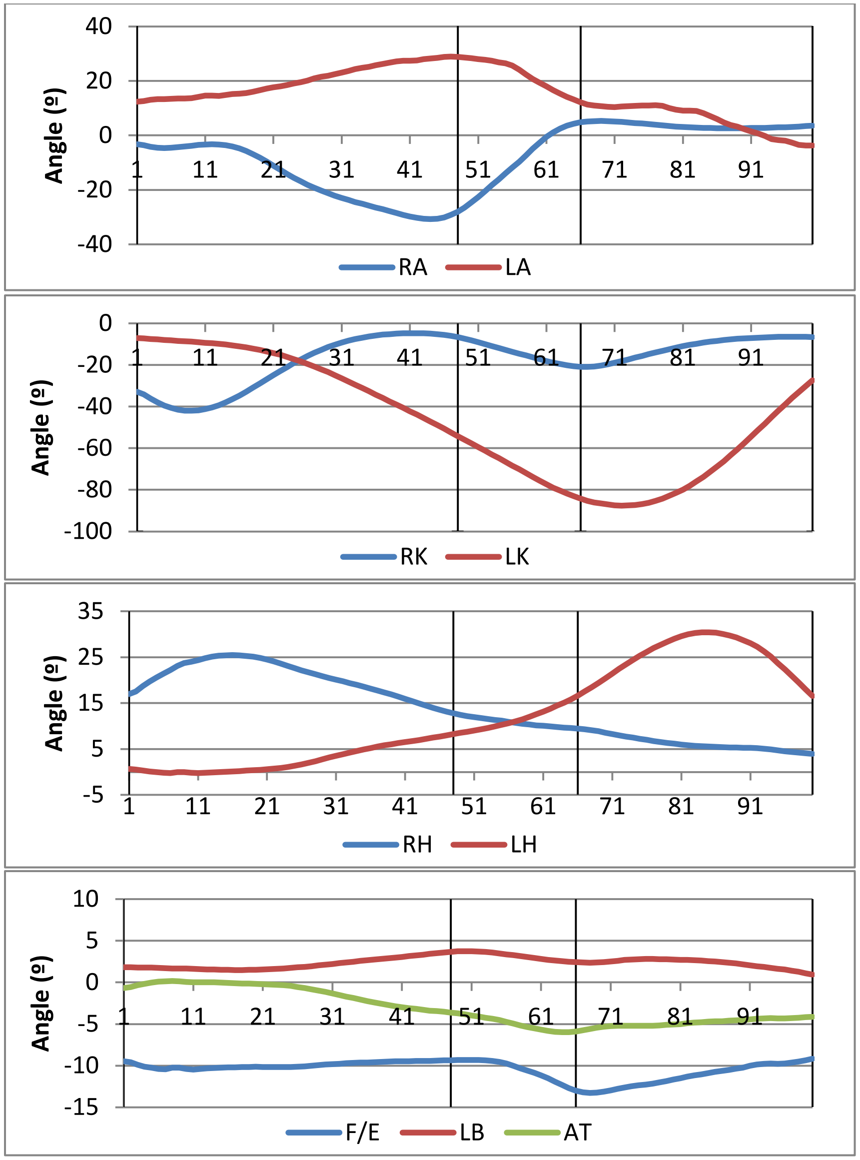

Figure D-3: Control group joint angles during the step down task.

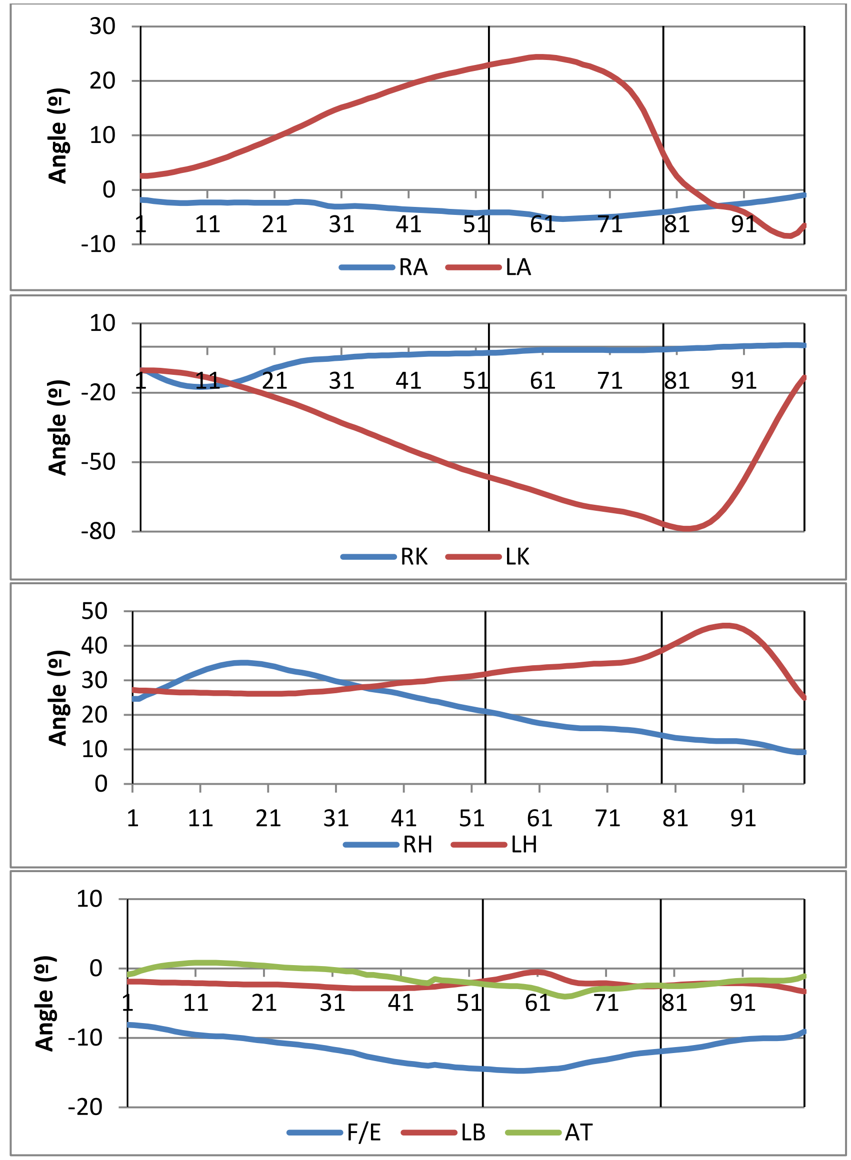

Figure D-4: Amputee group joint angles during the step down task.

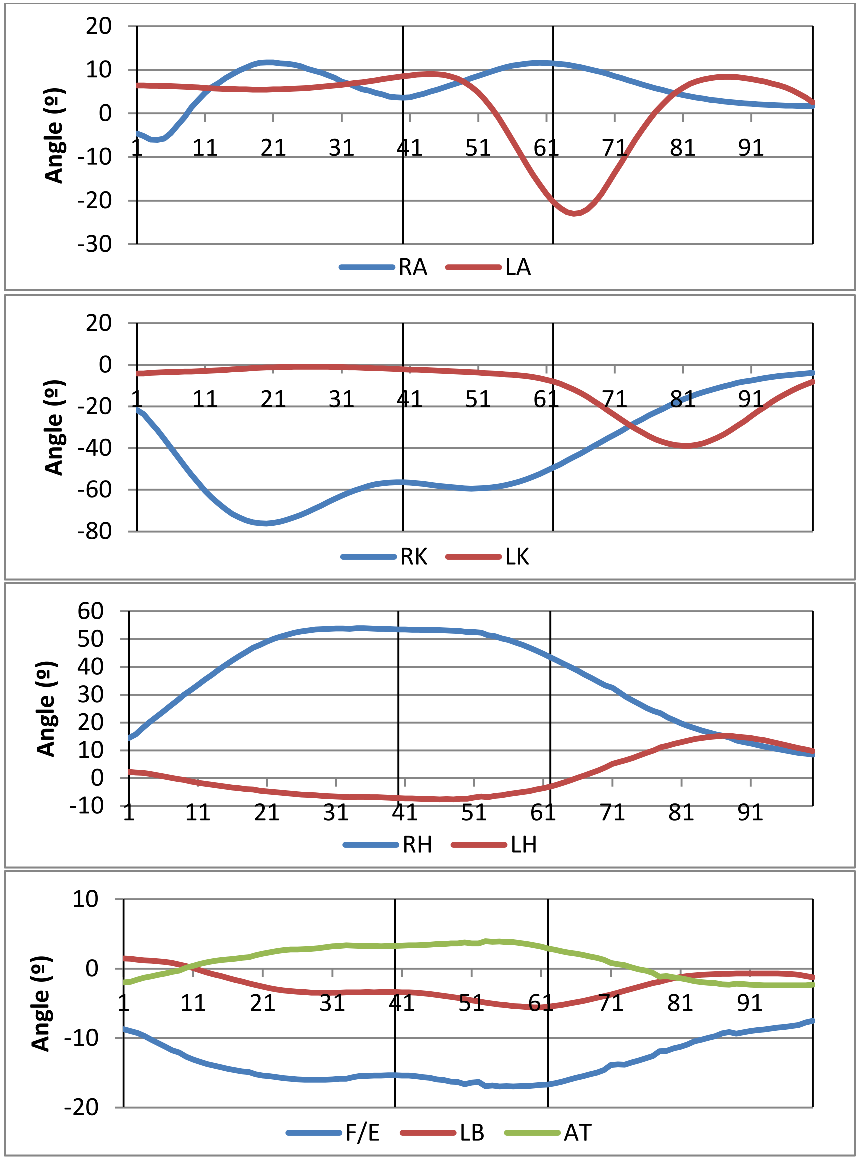

Figure D-5: Control group joint angles during the step up task.

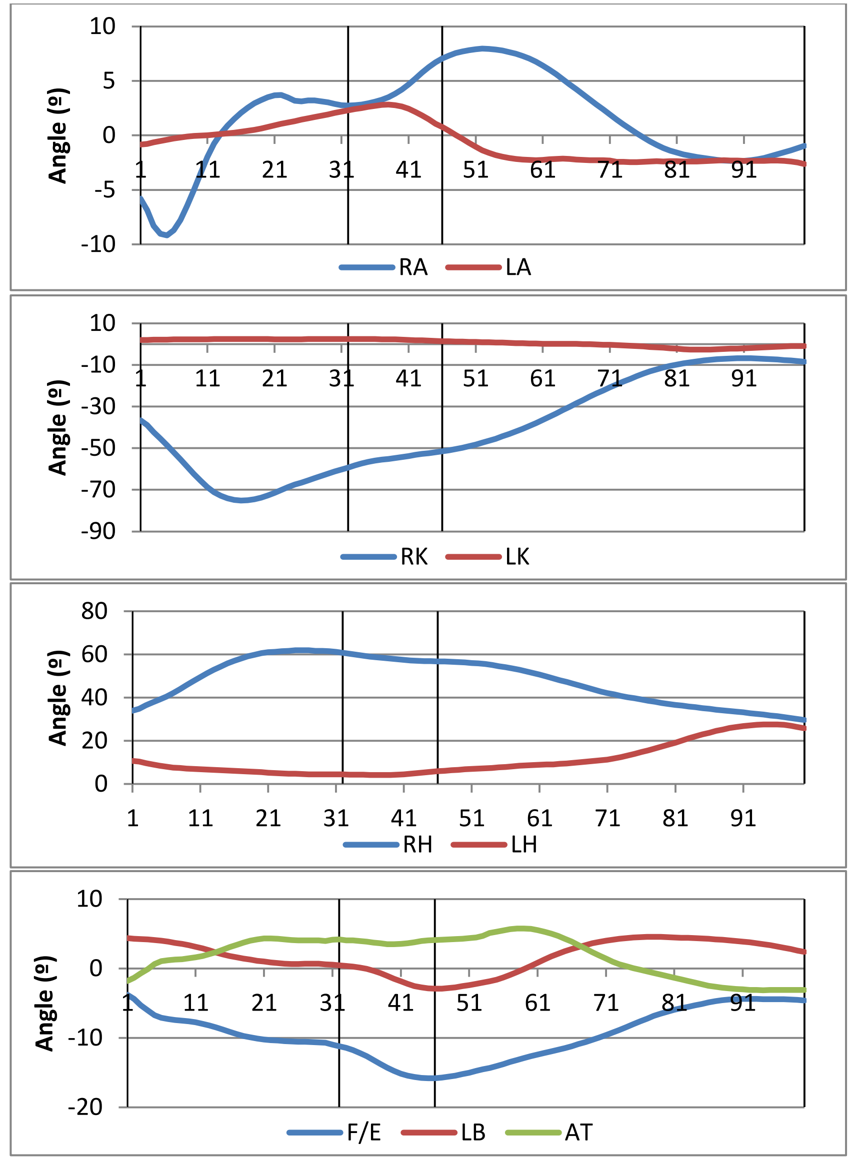

Figure D-6: Amputee group joint angles during the step up task.

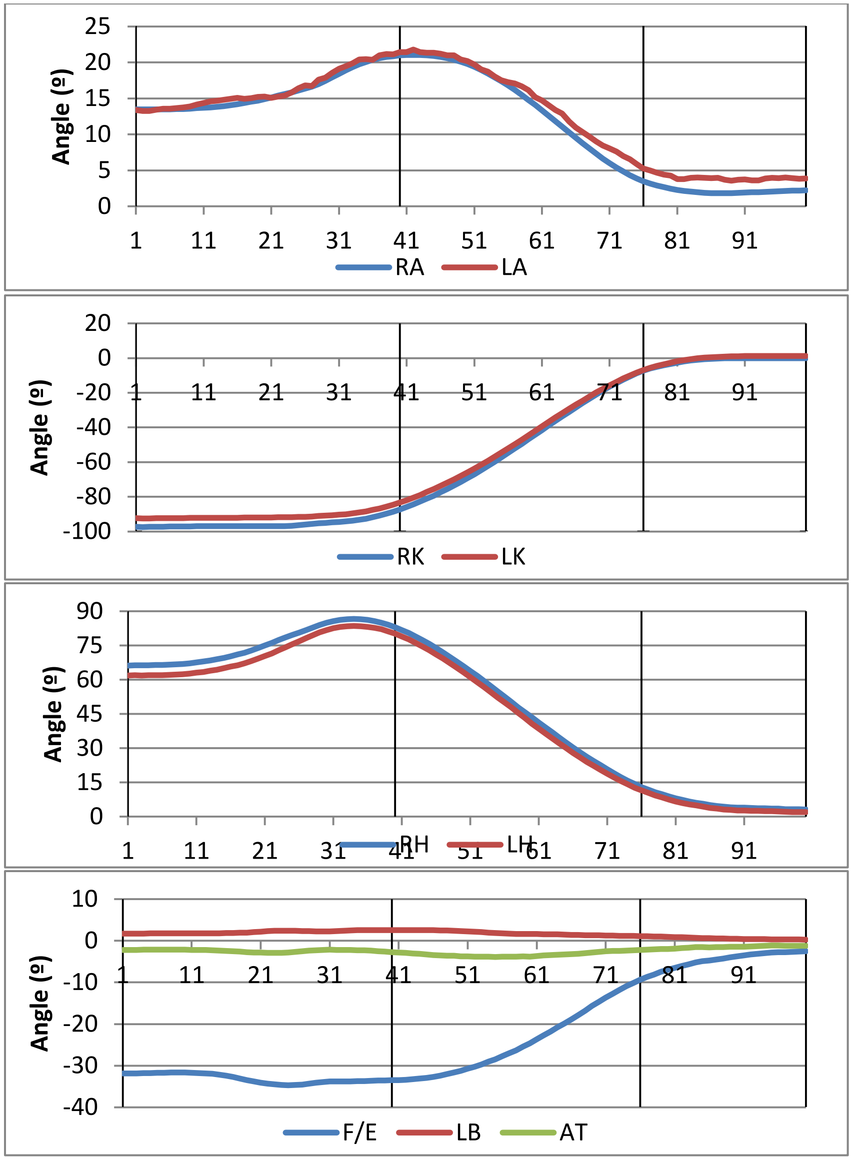

Figure D-7: Control group joint angles during the sit-to-stand task.

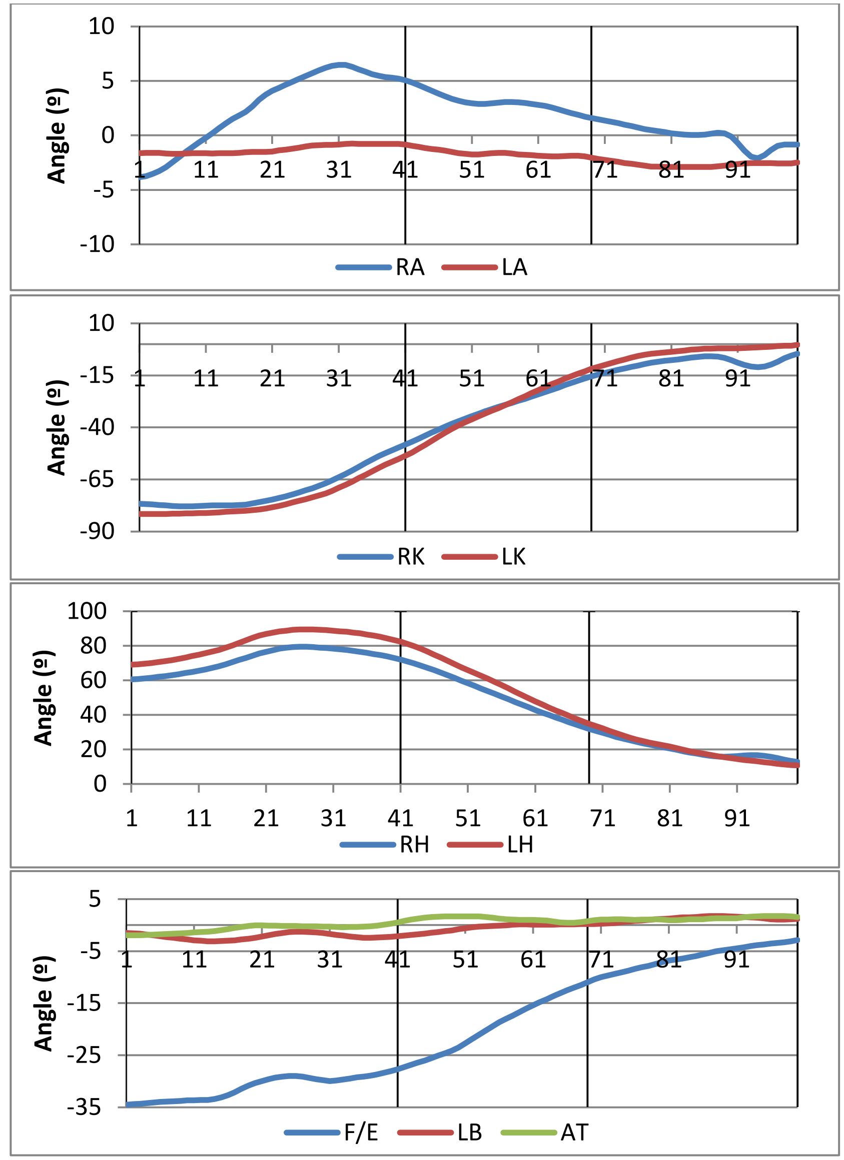

Figure D-8: Amputee group joint angles during the sit-to-stand task.

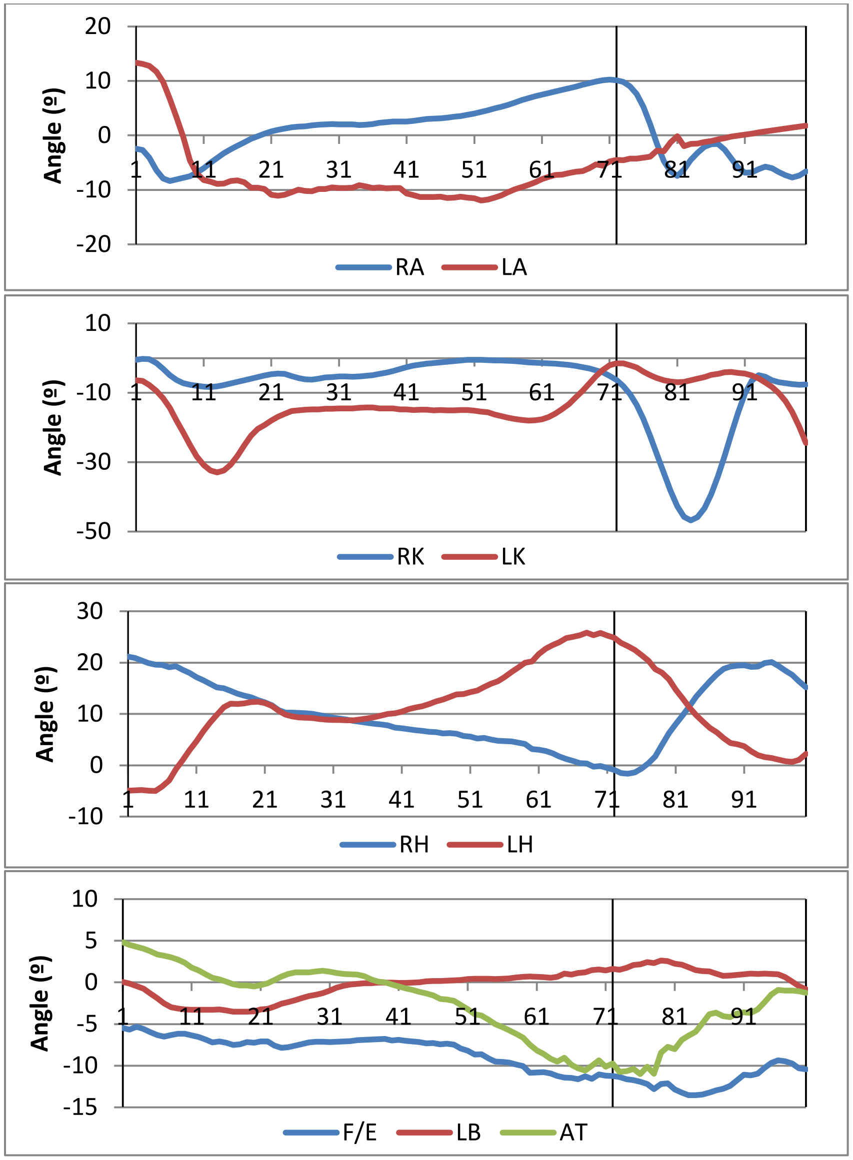

Figure D-9: Control group joint angles during the door pull task.

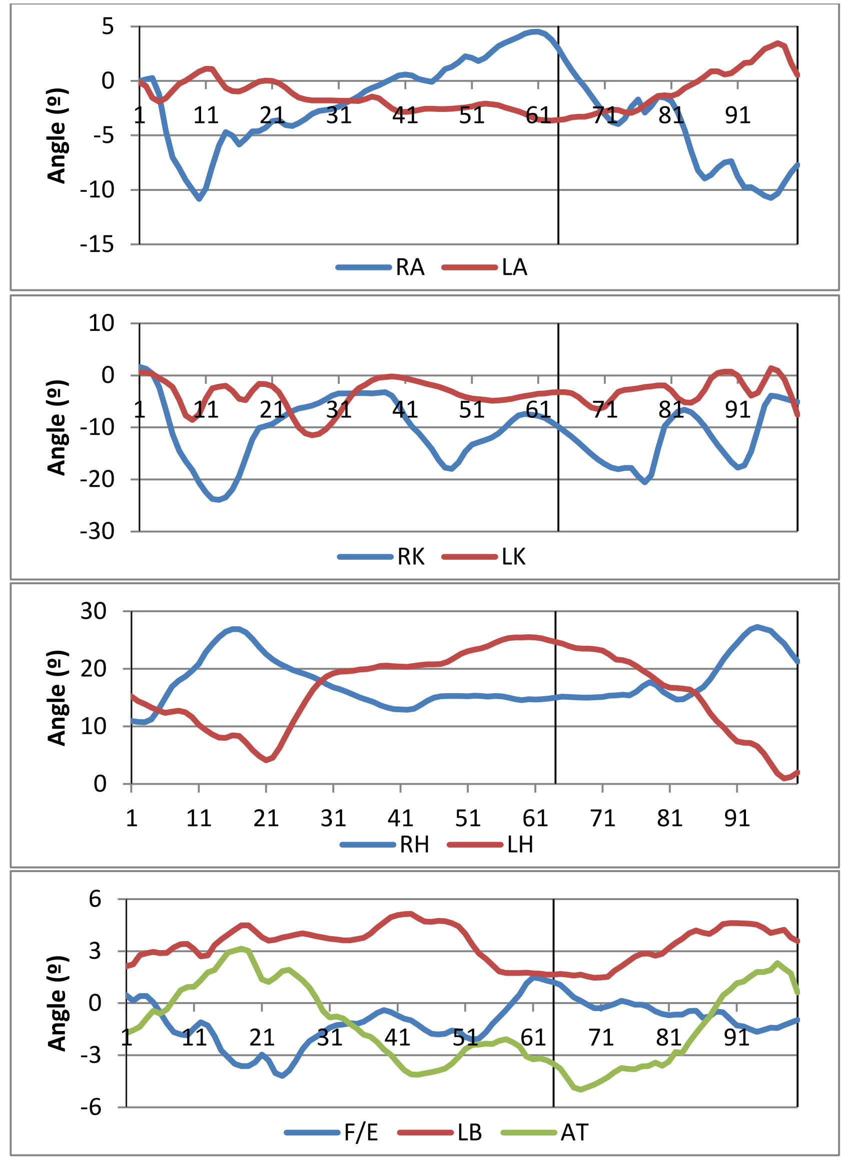

Figure D-10: Amputee group joint angles during the door pull task.

Table D-1: Subject group comparison of the average peak muscle activation during the gait task.

Muscle	Measure	CONTROL	AMPUTEE	Significant difference (p<0.05)
RES	Mean	11.54	19.27	***
	SD	8.53	4.65	
LES	Mean	14.32	56.79	***
	SD	9.1	10.2	
RSLM	Mean	21.57	67.56	***
	SD	9.96	45.47	
LSLM	Mean	19.1	63.66	***
	SD	9.67	28.45	
RLD	Mean	9.13	52.41	***
	SD	3.09	17.82	
LLD	Mean	10.92	42.81	***
	SD	3.34	19.68	
RRA	Mean	5.84	26.26	***
	SD	2.14	15.93	
LRA	Mean	5.72	19.34	***
	SD	2.24	10	
RIO	Mean	13.63	45.24	***
	SD	5.58	27.26	
LIO	Mean	13.25	52.34	***
	SD	6.03	16.08	
REO	Mean	10.34	21.19	***
	SD	5.12	10.67	
LEO	Mean	10.26	71.6	***
	SD	5.55	71.39	
RGMax	Mean	15.25	170.44	***
	SD	10.24	124.67	
LGMax	Mean	22.69	42.35	***
	SD	16.9	15.19	

Table D-2: Subject group comparison of the average peak muscle activation during the step down task.

Muscle	Measure	CONTROL	AMPUTEE	Significant difference (p<0.05)
RES	Mean	7.02	47.62	***
	SD	5.81	19.18	
LES	Mean	15.99	58.42	***
	SD	6.66	19.63	
RSLM	Mean	12.45	84.12	***
	SD	4.93	19.53	
LSLM	Mean	16.71	58.86	***
	SD	6.6	32.64	
RLD	Mean	7.76	46.23	***
	SD	3.91	14.11	
LLD	Mean	10.25	81.12	***
	SD	4.44	49.82	
RRA	Mean	7.96	28.1	***
	SD	2.15	10.35	
LRA	Mean	6.89	16.74	***
	SD	2.39	6.89	
RIO	Mean	17.87	40.95	***
	SD	7.99	14.42	
LIO	Mean	14.29	36.38	***
	SD	4.07	14.26	
REO	Mean	13.03	33.03	***
	SD	5.69	13.39	
LEO	Mean	11.55	60.31	***
	SD	4.23	81.93	
RGMax	Mean	9.43	204.2	***
	SD	4.16	132.83	
LGMax	Mean	10.24	54.81	***
	SD	4.69	29.3	

Table D-3: Subject group comparison of the average peak muscle activation during the step up task.

Muscle	Measure	CONTROL	AMPUTEE	Significant difference (p<0.05)
RES	Mean	13.79	23.44	***
	SD	6.73	5.25	
LES	Mean	20.95	67.02	***
	SD	7.88	7.18	
RSLM	Mean	25.91	90.47	***
	SD	4.78	60.58	
LSLM	Mean	21.89	73.5	***
	SD	7.07	25.53	
RLD	Mean	9.4	46.41	***
	SD	2.92	13.55	
LLD	Mean	14.5	102.82	***
	SD	3.62	74.82	
RRA	Mean	6.12	32.94	***
	SD	2.64	19.7	
LRA	Mean	5.43	25.02	***
	SD	2.83	15.14	
RIO	Mean	14.24	25.86	***
	SD	6.72	15.21	
LIO	Mean	12.74	59.26	***
	SD	6.05	25.43	
REO	Mean	9.75	21.7	***
	SD	4.11	7.91	
LEO	Mean	8.09	76.64	***
	SD	2.98	59.6	
RGMax	Mean	25.15	167.25	***
	SD	14.2	93.6	
LGMax	Mean	16.45	76.58	***
	SD	8.91	34.22	

Table D-4: Subject group comparison of the average peak muscle activation during the sit-to-stand task.

Muscle	Measure	CONTROL	AMPUTEE	Significant difference (p<0.05)
RES	Mean	22.68	29.81	
	SD	13.35	6.28	
LES	Mean	26.53	57.05	***
	SD	13.67	12.85	
RSLM	Mean	32.74	73.69	***
	SD	9.86	26.09	
LSLM	Mean	28.74	71.06	***
	SD	8.64	13.47	
RLD	Mean	13.82	57.99	***
	SD	6.98	16.18	
LLD	Mean	11.69	45.04	***
	SD	6.25	16.65	
RRA	Mean	5.21	32.22	***
	SD	2.91	9.92	
LRA	Mean	4.69	22.56	***
	SD	3.16	10.48	
RIO	Mean	9.23	58.18	***
	SD	3.52	51.56	
LIO	Mean	11.26	50.43	***
	SD	4.05	10.74	
REO	Mean	8.04	68.55	***
	SD	3.4	64.25	
LEO	Mean	8.93	83.5	***
	SD	3.98	31.71	
RGMax	Mean	19.47	84.09	***
	SD	12.77	39.25	
LGMax	Mean	21.23	68.51	***
	SD	14.81	20.52	

Table D-5: Subject group comparison of the average peak muscle activation during the door pull task.

Muscle	Measure	CONTROL	AMPUTEE	Significant difference (p<0.05)
RES	Mean	12.9	20.98	***
	SD	7.28	5.39	
LES	Mean	17.6	65.27	***
	SD	9.2	15.27	
RSLM	Mean	18.07	62.59	***
	SD	5.41	35.68	
LSLM	Mean	20	66.5	***
	SD	8.92	28.13	
RLD	Mean	8.83	48.73	***
	SD	2.38	31.56	
LLD	Mean	17.41	52.57	***
	SD	12.83	13.64	
RRA	Mean	5.89	27.08	***
	SD	2.31	15.44	
LRA	Mean	6.67	23.02	***
	SD	3.32	15.13	
RIO	Mean	13.09	30.47	***
	SD	5.14	17.11	
LIO	Mean	14.73	45.55	***
	SD	7.84	16.18	
REO	Mean	9.36	22.63	***
	SD	3.73	15.02	
LEO	Mean	11.01	64.41	***
	SD	5.16	56.52	
RGMax	Mean	14.72	115.5	***
	SD	7.79	58.54	
LGMax	Mean	12.06	44.59	***
	SD	3.18	14.23	

APPENDIX E – Subject group comparison of muscles' percent stiffness contribution

Table E-1: Comparison of individual muscle percent contribution to 3-dimensional rotational stiffness across subject groups during the gait task.

Muscle	Measure	Flex/Ext		Significant difference (p<0.05)	Lateral bend		Significant difference (p<0.05)	Axial twist		Significant difference (p<0.05)
		Control	Amputee		Control	Amputee		Control	Amputee	
RES	Mean	16.05	24.74	***	8.08	9.79		14.03	17.79	
	SD	7.06	15.26		4.21	5.99		6.76	10.61	
LES	Mean	12.29	13.32		6.38	5.57		10.52	9.55	
	SD	4.53	5.96		2.68	2.81		3.92	4.71	
RSLM	Mean	14.28	11.38		1.4	0.8	***	3.73	3.75	
	SD	3.41	8.05		0.43	0.86		1.11	1.44	
LSLM	Mean	11.25	9.78		1.04	0.68	***	2.92	2.08	***
	SD	3.95	3.89		0.39	0.33		1.11	0.9	
RLD	Mean	1.06	1.19		0.4	0.46		0.34	0.39	
	SD	0.39	0.22		0.11	0.2		0.09	0.13	
LLD	Mean	1.05	1.03		0.39	0.37		0.33	0.32	
	SD	0.53	0.2		0.16	0.11		0.12	0.04	
RRA	Mean	4.94	3.79	***	-0.02	-0.03	***	0.86	0.82	
	SD	0.99	0.84		0.01	0.03		0.2	0.34	
LRA	Mean	5.63	2.77		-0.01	-0.02	***	0.77	0.65	
	SD	7.92	1.38		0.01	0.03		0.17	0.49	
RIO	Mean	10.76	13.03		25.72	30.05		21.82	24.32	
	SD	5.04	7.88		8.49	12		8.39	11.29	
LIO	Mean	9.8	9.03		18.02	17.6		13.9	13.54	
	SD	5.11	4.4		6.7	5.16		5.85	4.85	
REO	Mean	4.62	2.05	***	13.25	7.15	***	13.61	7.72	***
	SD	1.63	0.83		4.12	3.41		4.09	3.65	
LEO	Mean	5.06	3.78		13.32	14.04		14	15.25	
	SD	3.5	0.88		7.34	5.75		7.91	5.43	
RQL	Mean	1.69	2.36	***	6.56	7.95		1.65	2.16	***
	SD	0.43	0.91		2.77	2.59		0.63	0.76	
LQL	Mean	1.52	1.76		5.46	5.58		1.51	1.66	
	SD	0.8	0.28		2.99	0.87		0.86	0.27	
Average task stiffness	Mean	84.99	287.67	***	166.75	628.28	***	69.27	256.54	***
	SD	57.67	121.87		56.52	149.97		24.69	107.6	

Table E-2: Comparison of individual muscle percent contribution to 3-dimensional rotational stiffness across subject groups during the step down task.

Muscle	Measure	Flex/Ext		Significant difference (p<0.05)	Lateral bend		Significant difference (p<0.05)	Axial twist		Significant difference (p<0.05)
		Control	Amputee		Control	Amputee		Control	Amputee	
RES	Mean	12.49	23.55	***	5.89	12.73	***	10.53	21.38	***
	SD	7.26	10.79		3.9	6.01		6.74	9.74	
LES	Mean	15.72	11.92	***	7.66	6.52		12.66	10.74	
	SD	2.36	7.05		1.88	3.95		2.92	6.6	
RSLM	Mean	10.6	15.92	***	0.95	1.66	***	2.7	4.68	***
	SD	2.12	2.43		0.2	0.12		0.47	0.39	
LSLM	Mean	11.82	7.51	***	1.02	0.68	***	3.13	1.98	***
	SD	2.77	4.58		0.36	0.3		1.14	1.11	
RLD	Mean	1.12	1.16		0.37	0.52	***	0.35	0.4	
	SD	0.48	0.29		0.14	0.15		0.14	0.13	
LLD	Mean	1.12	1.79	***	0.36	0.86	***	0.34	0.67	***
	SD	0.47	0.52		0.12	0.35		0.12	0.3	
RRA	Mean	5.8	4.86	***	-0.01	-0.04	***	0.95	0.91	
	SD	1.43	0.79		0.01	0.02		0.29	0.42	
LRA	Mean	4.76	2.65	***	-0.01	-0.02	***	0.78	0.5	***
	SD	1.42	0.57		0.01	0		0.28	0.26	
RIO	Mean	12.65	11.59		27.91	26.13		22.55	21.29	
	SD	4.5	6.1		6.74	7.12		6	7.33	
LIO	Mean	10.9	8.06		18.65	15.22		14.42	11.9	
	SD	4.26	4.19		5.21	4.9		4.41	4.6	
REO	Mean	5.14	4.71		13.74	12.91		14.94	13.6	
	SD	1.39	2.95		4.06	5.98		4.27	6.38	
LEO	Mean	4.75	1.54	***	12.5	6.19	***	13.7	7.18	***
	SD	2.03	1.88		5.06	6.06		5.69	6.76	
RQL	Mean	1.25	3.18	***	4.55	11.38	***	1.15	3.23	***
	SD	0.21	0.84		1.45	3.22		0.25	1.08	
LQL	Mean	1.88	1.56		6.42	5.24		1.8	1.53	
	SD	0.67	0.59		2.35	1.88		0.75	0.61	
Average task stiffness	Mean	74.16	285.82	***	186.79	529.55	***	69.56	248.72	***
	SD	28.71	93.36		57.06	126.27		21.29	38.62	

Table E-3: Comparison of individual muscle percent contribution to 3-dimensional rotational stiffness across subject groups during the step up task.

Muscle	Measure	Flex/Ext Control	Flex/Ext Amputee	Significant difference (p<0.05)	Lateral bend Control	Lateral bend Amputee	Significant difference (p<0.05)	Axial twist Control	Axial twist Amputee	Significant difference (p<0.05)
RES	Mean	15.43	26.32	***	9.01	13.78	***	15.42	24.8	***
	SD	6.29	13.81		4.21	6.62		6.46	11.84	
LES	Mean	17.16	14.84		10.36	8.73		16.6	15.29	
	SD	2.34	7.35		1.87	4.55		2.27	8.34	
RSLM	Mean	13.92	14.45		1.63	1.45		4.12	4.76	***
	SD	3.57	3.76		0.37	0.56		0.99	0.63	
LSLM	Mean	12.37	10.04	***	1.36	0.97	***	3.58	2.74	***
	SD	2.1	2.9		0.29	0.28		0.79	0.92	
RLD	Mean	1.04	1.19		0.46	0.52		0.35	0.42	
	SD	0.47	0.32		0.19	0.18		0.15	0.13	
LLD	Mean	1.1	1.41	***	0.49	0.76	***	0.38	0.64	***
	SD	0.4	0.49		0.18	0.47		0.12	0.38	
RRA	Mean	4.63	4.24		-0.03	-0.03		0.74	0.82	
	SD	1.74	0.95		0.01	0.03		0.26	0.31	
LRA	Mean	3.93	3.04		-0.02	-0.02		0.65	0.63	
	SD	1.56	1.52		0.01	0.03		0.3	0.42	
RIO	Mean	9.63	4.29	***	22.79	13.42	***	18.9	9.7	***
	SD	4.06	2.43		6.81	8.28		6.32	5.7	
LIO	Mean	9.28	9.05		18.45	19.58		14.11	14.21	
	SD	3.06	4.72		3.92	7.63		3.46	6.34	
REO	Mean	4.04	2.57	***	10.97	8.35	***	10.93	7.78	***
	SD	1.05	1.31		2.34	3.74		2.52	3.36	
LEO	Mean	3.63	3.56		10.3	13.18		10.39	13.18	
	SD	1.59	2.23		4.63	5.69		4.82	5.78	
RQL	Mean	1.75	2.96	***	6.63	11.51	***	1.74	2.92	***
	SD	0.17	0.75		0.69	2.32		0.2	0.58	
LQL	Mean	2.1	2.04		7.61	7.81		2.11	2.1	
	SD	0.58	0.6		2	2.23		0.64	0.72	
Average task stiffness	Mean	85.8	300.98	***	161.05	591.9	***	76.68	261.66	***
	SD	29.27	109.67		45.52	154.06		22.51	96.43	

Table E-4: Comparison of individual muscle percent contribution to 3-dimensional rotational stiffness across subject groups during the sit-to-stand task.

Muscle	Measure	Flex/Ext		Significant difference (p<0.05)	Lateral bend		Significant difference (p<0.05)	Axial twist		Significant difference (p<0.05)
		Control	Amputee		Control	Amputee		Control	Amputee	
RES	Mean	18.35	18.62		13.41	10.13	***	21.57	21.05	
	SD	5.71	11.24		4.76	4.59		6.59	12.43	
LES	Mean	17.45	13.09	***	14.24	9.75	***	22.46	17.25	***
	SD	3.37	6.44		3.4	5.29		4.53	9.43	
RSLM	Mean	13.66	11.87		1.94	1.32	***	5.08	4.18	***
	SD	4.45	2.57		0.55	0.3		1.23	0.87	
LSLM	Mean	12.2	7.77	***	1.71	0.98	***	4.8	2.75	***
	SD	3.72	1.59		0.52	0.29		1.67	0.98	
RLD	Mean	1.03	1.41	***	0.47	0.58	***	0.39	0.49	***
	SD	0.37	0.39		0.14	0.1		0.13	0.1	
LLD	Mean	0.92	1.05		0.42	0.44		0.35	0.38	
	SD	0.16	0.31		0.1	0.16		0.07	0.16	
RRA	Mean	5.1	7.74	***	-0.01	-0.01		0.63	0.8	
	SD	2.08	1.56		0.01	0.01		0.28	0.2	
LRA	Mean	4.69	5.4		-0.01	-0.01		0.58	0.61	
	SD	2.21	2.54		0.01	0.01		0.31	0.43	
RIO	Mean	6.55	8.24		14.8	18.51	***	11.33	14.58	***
	SD	3.07	2.12		5.54	5.03		4.94	3.62	
LIO	Mean	8.25	10.19		15.05	17.06		10.82	12.93	
	SD	3.56	4.41		5.21	6.33		4.33	5.38	
REO	Mean	3.41	3.36		7.54	8.08		7.84	6.73	
	SD	1.44	1.47		2.51	3.73		2.75	3.09	
LEO	Mean	3.64	6.76	***	8.22	15.81	***	8.51	13.74	***
	SD	0.84	1.74		2.25	4.82		2.02	3.35	
RQL	Mean	2.53	2.61		11.98	9.68	***	2.96	2.5	***
	SD	0.62	1.06		2.33	2.15		0.57	0.57	
LQL	Mean	2.21	1.89		10.24	7.69	***	2.68	2.01	
	SD	0.86	0.53		4.2	1.64		1.2	0.81	
Average task stiffness	Mean	83.66	261.2	***	172.7	650.6	***	73.96	264.5	***
	SD	32.36	81.24		50.75	303.23		25.13	104.68	

Table E-5: Comparison of individual muscle percent contribution to 3-dimensional rotational stiffness across subject groups during the door pull task.

Muscle	Measure	Flex/Ext		Significant difference (p<0.05)	Lateral bend		Significant difference (p<0.05)	Axial twist		Significant difference (p<0.05)
		Control	Amputee		Control	Amputee		Control	Amputee	
RES	Mean	17.95	26.3	***	9.43	12.04		15.55	21.89	***
	SD	7.21	12.77		4.12	6.68		6.11	12.2	
LES	Mean	16.37	14.13		8.92	6.45	***	14.21	11.2	
	SD	3.27	6.69		2.48	2.98		3.85	5.48	
RSLM	Mean	14.23	15.95		1.46	1.31		3.97	4.21	
	SD	3.47	2.53		0.29	0.3		0.87	0.94	
LSLM	Mean	11.91	9.77		1.17	0.74	***	3.35	2.24	***
	SD	4.05	3.24		0.51	0.22		1.44	0.73	
RLD	Mean	0.95	1		0.4	0.44		0.33	0.37	
	SD	0.36	0.51		0.13	0.3		0.1	0.24	
LLD	Mean	1.56	1.14		0.64	0.43	***	0.53	0.4	
	SD	0.91	0.28		0.32	0.06		0.26	0.08	
RRA	Mean	3.62	3.57		-0.01	-0.02		0.7	0.85	
	SD	0.9	0.69		0.01	0.02		0.29	0.31	
LRA	Mean	3.49	2.51	***	-0.01	-0.02		0.69	0.64	
	SD	1.02	1.19		0.01	0.02		0.34	0.44	
RIO	Mean	9.05	7.66		22.13	21.62		17.34	17.01	
	SD	4.35	5.3		6.75	11.13		5.44	9.71	
LIO	Mean	10.1	8.09		20.71	18.27		15.45	13.59	
	SD	4.32	4.32		5.65	6.98		4.89	5.86	
REO	Mean	3.6	2.17	***	10.61	8.36		11.85	8.73	***
	SD	1.42	1.02		2.99	3.88		3.75	4.16	
LEO	Mean	3.62	3.26		10.95	13.56		12.35	14.34	
	SD	1.43	1.73		3.52	6.36		4.05	6.68	
RQL	Mean	1.73	2.6	***	6.87	10.01	***	1.77	2.63	***
	SD	0.35	0.75		1.06	2.59		0.29	0.77	
LQL	Mean	1.82	1.86		6.73	6.8		1.92	1.9	
	SD	0.66	0.45		2.38	0.99		0.73	0.39	
Average task stiffness	Mean	83.96	288.54	***	161.26	558.76	***	67.76	230.67	***
	SD	37.72	145.24		54.69	173.58		21.38	117.44	

APPENDIX F – Example data from STS trials of a control participant and TFA4

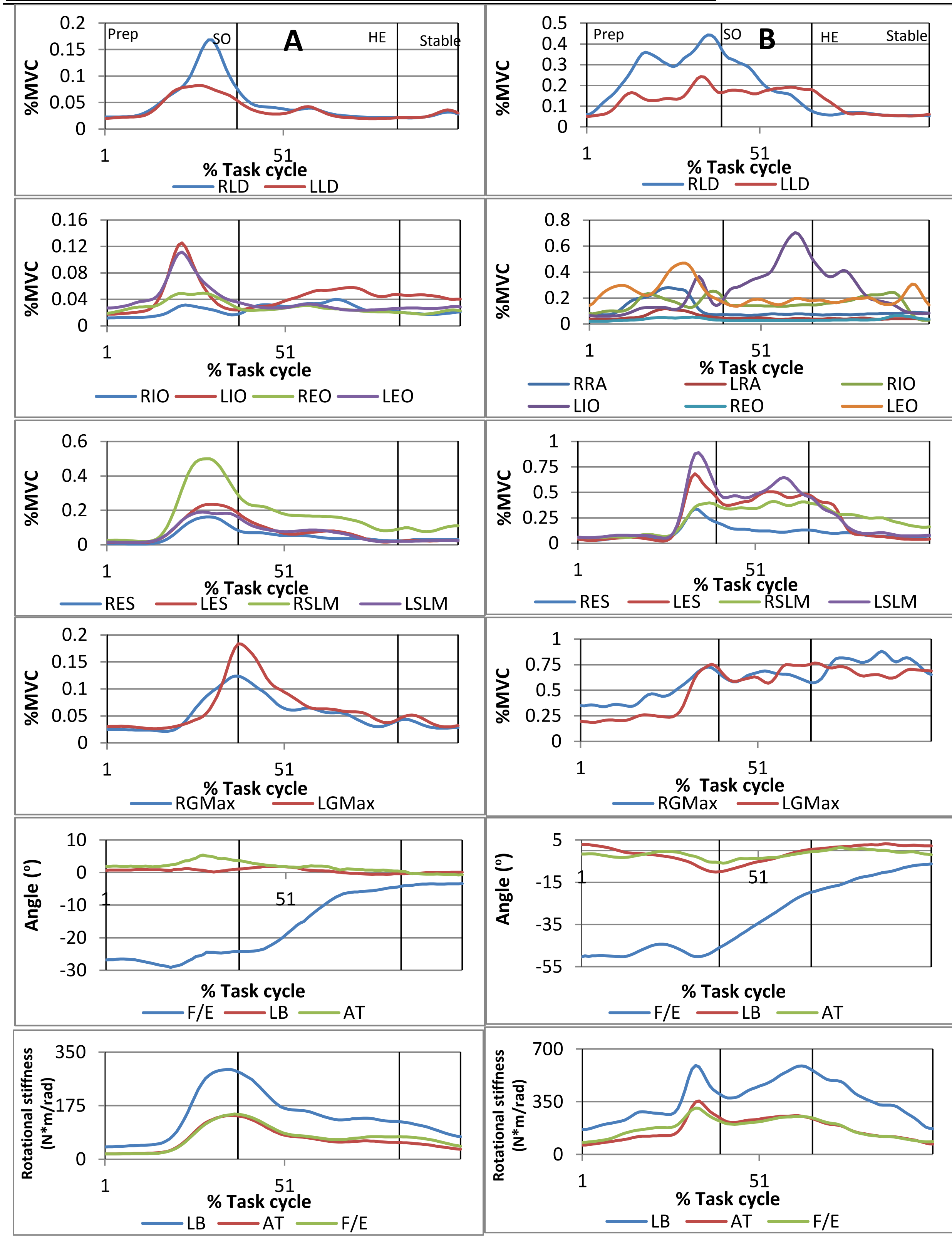

Figure F-1: Muscle activation patterns, trunk motion, and stiffness measurements for example trials from control (A) and TFA4 (B) participants, demonstrating sequential stiffening to initiate seat off, followed by increased trunk activation and stiffness from TFA4 to actively contribute to both knee and trunk extension.