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SEGMENTAL MOVEMENT COMPENSATIONS IN PATIENTS WITH
TRANSTIBIAL AMPUTATION IDENTIFIED USING ANGULAR MOMENTUM
SEPARATION

A Dissertation

Presented to

the Faculty of the Daniel Felix Ritchie School of Engineering and Computer Science
University of Denver

In Partial Fulfillment

of the Requirements for the Degree

Doctor of Philosophy

by

Brecca M. M. Gaffney

June 2017

Advisor: Bradley S. Davidson, PhD

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Author: Brecca M. M. Gaffney

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ABSTRACT

Patients with unilateral dysvascular transtibial amputation (TTA) adopt movement compensations that are required to maintain balance and achieve ambulation in the absence of ankle plantar flexion, and result in increased and asymmetric joint loading patterns. As a result, this population is at an increased risk of overuse injuries, such as low back pain (LBP). Clinical gait analysis is used to guide diagnostics in movement retraining following amputation, and is performed using instrumented (research based) or observational analyses (clinically based). However, instrumented analyses are currently impractical in most clinical settings due to expense and computational limitations. This dissertation presents the use of segmental angular momentum to describe movement compensations in patients with TTA, and assess their effects on the musculoskeletal system; which provides a potential platform applicable in both instrumented and observational settings.

Ten patients with unilateral dysvascular TTA and two cohorts (patients with diabetes mellitus and healthy controls) completed one experimental study in which whole-body kinematics and core muscle demand were collected during walking and bilateral stepping tasks. Specific Aim 1 described the foundations of the separation of angular momentum into two components, translational (TAM) and rotational angular momentum (RAM) to describe movement coordination during healthy walking. Euler's rotational laws were

used to calculate segmental translational and rotational moments, which provide insight into the effort required to generate and arrest momentum by their relation to external forces and moments. Specific Aim 2 described trunk and pelvis movement compensations in patients with TTA during walking using TAM and RAM. Specific Aim 3 described the trunk translational and rotational moments in patients with TTA during step ambulation. Finally, Specific Aim 4 described the demand from the core musculature that supports trunk movement compensations in patients with TTA during step ambulation.

The results from these Specific Aims indicate that patients with TTA generate larger amounts of TAM and RAM, which were caused by larger translational and rotational trunk moments and demand from core muscles, than healthy controls. These compensations alter the low back loading patterns, which may be reduced by targeted strengthening and retraining motor control compensations to better support trunk movements.

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CHAPTER 1: INTRODUCTION

Currently in the United States, over one million individuals live with a lower-limb amputation and this number is expected to more than double by 2050 (Ziegler-Graham et al., 2008). This marked increase is primarily attributed to an aging population with neurovascular pathologies (e.g. diabetes mellitus) (Dillingham et al., 2002). Although over 80% of all lower-extremity amputations are due to neurovascular pathologies (Dillingham et al., 2002), this population is vastly underrepresented in biomechanical research (Fortington et al., 2012). This is particularly problematic because patients with dysvascular amputation commonly suffer from additional comorbidities that affect physical function, such as low back pain (Ehde et al., 2001). Therefore, 40-50% have limited physical function one year following amputation, which negatively affects the overall quality of life (Davies & Datta, 2003).

The majority of biomechanical research that identifies movement compensations adopted by patients with transtibial amputation (TTA) combine patients with multiple etiologies of amputation (e.g. traumatic, dysvascular, and cancerous) (Fey et al., 2010; Silverman and Neptune, 2011; Ventura et al., 2011; Huang and Ferris, 2012; Schaarschmidt et al., 2012). However, patients with dysvascular amputation have distinct differences in age, BMI, and time frame of prosthetic use compared to patients with traumatic amputation (Davies & Datta, 2003; Nehler et al., 2003; van Velzen et al., 2006). In addition, most amputee research is focused on compensations adapted in the

lower extremities during gait, which do not encompass all movement compensations during high-demand tasks that are required for community ambulation. Therefore, this dissertation will focus on movement compensations adopted only by patients with dysvascular TTA during both over-ground walking and high-demand tasks.

Following amputation, movement compensations are identified using clinical gait analysis, which is defined as the systematic measurement and interpretation of quantities that characterize human locomotion, and is most commonly accomplished using instrumented and observational analyses (Krebs et al., 1985; Saleh & Murdoch, 1985). Instrumented gait analysis (kinematics, kinetics, and electromyography) is currently the gold standard to fully describe and accurately quantify human movement. In an effort to prevent a patient adopting a potential movement pattern that can negatively affect physical function, clinicians rely heavily on observation during rehabilitation following TTA to identify and correct potential consequential movement compensations adopted by patients with TTA that are used to compensate for the loss of the ankle plantar flexors and foot musculature. Although indispensable in providing immediate feedback to the patient, observational gait analysis often lacks the necessary sensitivity and reliability to detect and diagnose potential consequential movement patterns adopted by patients with TTA; and therefore, clinicians are left to use their intuition and training to hypothesize regarding how specific movement compensations alter muscle and joint demand. Because of this, many movement patterns are often prone to misidentification through observation (Shores, 1980; Robinson & Smidt, 1981; Holden et al., 1984; Krebs et al., 1985; Frigo et al., 1998; Coutts, 1999; Toro et al., 2003). In addition, although instrumented analysis is

sensitive and accurate at identifying specific movement compensations, specialized and expensive equipment and personnel are required, which results in it rarely being used in a clinical setting. Therefore, although both instrumented and observational analyses are used to identify compensations adopted by patients with TTA, a clear translational pathway between the two currently does not exist.

The separation of segmental angular momentum provides a potential translational pathway between instrumented and observational analyses because it is a foundation in Newton-Euler mechanics and is composed of observable and interpretable quantities. In order to develop feasible translation between instrumented and observational gait analyses, the relation between angular momentum and common forms of instrumented gait analyses must first be established. The overall goal of this project is to identify movement compensations in patients with unilateral dysvascular TTA during gait and high-demand tasks, and assess the effects of compensations on the musculoskeletal system, using the principle of separation of segmental angular momentum. To accomplish this, this project is divided into four specific aims:

Specific Aim 1: Describe the foundations of the separation of segmental angular momentum that can be used to translate between observational and instrumented gait analyses.

Specific Aim 2: Identify segmental movement compensations in patients with TTA using the separation of angular momentum.

Specific Aim 3: Identify the trunk kinetic effort required during step ambulation in patients with TTA using segmental momenta.

Specific Aim 4: Identify the demands from core muscles that were used to support trunk movement compensations adopted by patients with TTA during step ambulation.

1.1 Dissertation Overview

Chapter 2 presents an overview of common tools used within instrumented and observational gait analyses. Within this review, the associated common dependent variables, and how they are used to identify movement compensations, are presented. Chapter 3 provides a literature review of movement compensations in patients with unilateral TTA that are identified via instrumented and observational gait analyses. For continuity in the cohort used in Chapters 5-7, this review encompasses movement compensations in patients with passive transtibial prostheses. Chapter 4 presents the foundations of the separation of segmental angular momentum based on Euler's rotational laws and demonstrates how this can be applied to describe movement coordination during walking in a cohort of healthy adults (Specific Aim 1). Total segmental angular momentum is separated into two independent components, translation (TAM) and rotational angular momentum (RAM), which provide different but complementary perspectives of the segmental dynamics needed to achieve forward progression.

Using these foundations, Chapters 5-7 are experimental studies that apply the separation of angular momentum to describe movement compensations, and their effects on the musculoskeletal system, in patients with unilateral dysvascular TTA. Chapter 5 is a study that describes trunk and pelvis movement patterns in patients with unilateral

dysvascular TTA by assessing patterns of generating and arresting segmental TAM and RAM during over-ground walking (Specific Aim 2). Chapter 6 is a study that assesses the required trunk kinetic effort, which is defined via the translational and rotational moments (time derivative of TAM and RAM, respectively) to ascend and descend a step (Specific Aim 3). Chapter 7 is a study that describes the demand from the core muscles that are used to support trunk movement compensations adopted by patients with TTA during step ascent and descent (Specific Aim 4). Chapter 8 is a summary and conclusions of the main findings of this project and provides recommendations of future research based on the present findings. Appendix A describes subject anthropometrics of all subjects used in Chapters 4-7. Appendix B describes the clinical measures of functional performance of each subject that was included in the analyses in Chapters 4-7. Appendix C presents the individual ensemble averaged curves of each patient with unilateral dysvascular amputation for all dependent variables presented in Chapters 5-7.

CHAPTER 2: CLINICAL GAIT ANALYSIS – A REVIEW OF METHODS AND OUTCOMES OF INSTRUMENTED AND OBSERVATIONAL ANALYSES

2.1 Introduction

Clinical gait analysis, which involves systematic measurement, identification, and interpretation of quantities that characterize human walking (Perry & Burnfield, 2010; Saleh & Murdoch, 1985), is widely used by researchers and clinicians to understand the human movement system. Two primary applications of clinical gait analysis are to identify movement dysfunction and assess outcomes of interventions that target movement quality (Shull et al., 2014). The methods of performing clinical gait analysis fall generally into two categories: instrumented and observational gait analysis (Saleh & Murdoch, 1985). Instrumented gait analysis, most commonly performed in a research setting consists of measurement of kinematics, kinetics, and electromyography (EMG) (Messenger & Bowker, 1987; Morton, 1999; Carollo & Matthews, 2009). However, although instrumented gait analysis remains the current gold standard of accurate quantification of pathologic movements, it is still considered impractical in the vast majority of clinical settings due to monetary and computational restrictions (Toro et al., 2003). In contrast, observational gait analysis is most often preferred by clinicians over instrumented gait analysis because it does not require specialized laboratory equipment and allows immediate feedback to be provided to the patient during rehabilitation (Krebs et al., 1985). Observational gait analysis consists of the observer assessing the body's

(segmental and joint) movements in two planes (sagittal and frontal) throughout the rapidly repeating gait cycle (Saleh & Murdoch, 1985). The objective of this review is to describe the most common methods, outcome variables, and clinical interpretation of instrumented and observational gait analyses, as well as to demonstrate how momentum is a potential link between these two forms of analyses.

2.2 Instrumented Gait Analyses

Instrumented gait analysis is a powerful technological tool that is used to quantify the characteristics of human movement using specialized equipment to measure kinematics, kinetics, and dynamic electromyography (EMG) during gait. While there are other useful forms of instrumentation and measurement (e.g. radiographic image, dynamic ultrasound, etc.), a comprehensive gait analysis laboratory must require these three fundamental components of measurement for accreditation (Kaufman et al., 2001); and therefore, this review will focus on the application and interpretation of measurements of these components.

Instrumented gait analysis is the current gold standard for accurate quantification of gait assessment because it is highly objective, sensitive, and reliable in comparison to observational gait analysis (Krebs et al., 1985; Eastlack et al., 1991; Keenan & Bach, 1996). When combined, the three foundational forms of instrumented gait analysis provide empirical evidence for understanding the underlying cause of a movement dysfunction and their potential consequential effects on the musculoskeletal system.

2.2.1 Gait Kinematics

Kinematics are defined as the branch of mechanics that describe the motion of an object without regard to the external forces and rotational torques that cause motion. Gait kinematics are therefore composed of the measurement of linear and angular displacements, velocities, and accelerations of the body segments throughout movement (Carollo & Matthews, 2009). The most common measurement system used to collect kinematics for instrumented gait analysis uses passive infrared motion capture systems to track the three-dimensional global position of reflective markers placed on skin above specific anatomic landmarks relative to an inertial reference frame. Each body segment is defined using at least three non-collinear markers and assumed to be rigid (Robertson et al., 2004). The primary sources of error in measurements using reflective motion captures systems are marker placement and skin movement artifact (Della Croce et al., 1999; Chiari et al., 2005; Della Croce et al., 2005; Leardini et al., 2005; Gao & Zheng, 2008).

Using the position and orientation of each segment relative to an inertial reference frame, each segment is modeled as a rigid body and the most common outcome variables consist of joint angles and segment angles (Kadaba et al., 1990). Joint angles are calculated between a distal and proximal segment about a fixed point at the center of the joint, and are most commonly quantified using Euler angles to describe the relative rotation of the distal segment relative to the proximal segment (Chao et al., 1983). Segment angles are calculated as the absolute angle of the segment orientation relative to a fixed global origin.

During gait, joint angles are most commonly used to describe lower extremity movements and segment angles are most commonly used to describe upper extremity movements. Movement pathologies using joint angles are most commonly identified based on deviations of the hip, knee, and ankle joints that are different (less or greater) than a reference group or contralateral limb, dependent upon the pathology (McGinley et al., 2009). For example, patients with unilateral transtibial amputation (TTA) demonstrate higher ranges of hip flexion angles on the amputated limb in comparison to the intact limb (Isakov et al., 2000; Bateni & Olney, 2002), which is typically interpreted to be an effect of a forward trunk lean that is adopted to be a forward progression and balance strategy to reduce the demand on the knee extensors (Winter & Sienko, 1988; Sanderson & Martin, 1997). Segment angles are most commonly used to describe the trunk and pelvis segment movements (McGinley et al., 2009). For example, pelvic obliquity is quantified using the angle of the pelvis relative to the global origin in the frontal plane, and is most commonly used to describe dysfunctional loading patterns during weight acceptance (Perry & Burnfield, 2010). Although segment kinematics are useful to accurately quantitatively describe segmental and whole-body motion during movement, they do not provide insight regarding mechanisms behind how the motion is achieved, and are therefore less common descriptors of human movement than joint angles.

2.2.2 *Gait Kinetics*

Kinetics are defined as the branch of mechanics that describe the underlying external forces and rotational torques that cause motion. External forces during walking consist of the ground reaction force (GRF) between the foot and the ground, and are measured

through embedded force platforms (Carollo & Matthews, 2009). Force platforms measure force by converting deformation (strain) that is due to a load into an electrical potential using two primary mechanisms: piezoelectric material or a strain gauge. The three-dimensional GRF (anterior-posterior shear, mediolateral shear, and vertical force) measured from the force platforms are used to calculate inverse dynamics (ID) to solve for joint kinetics (joint moments and joint powers).

ID analyses are the most common descriptors of gait kinetics and are dependent upon segment kinematics, external forces, and segment inertial parameters (segment mass, center of mass location, and moments of inertia) (Winter, 2009). Beginning with the foot, ID most commonly solves joint kinetics using a “bottom up approach” in which joint forces and moments are calculated using the linear and angular forms of Newton’s 2nd Law, and propagating up the kinetic chain represented in a link-segment model (iterative Newton-Euler method) (Robertson et al., 2004). The resultant moment is referred to as either internal (the moment generated by internal ligaments and muscles acting on the musculoskeletal system required to counteract the external load) or external (moments acting on the body that propagate through the musculoskeletal system from the external load); the only difference being the sign of the vector.

Using the ID solution, the most common descriptors of movement dysfunction are through joint moments and power. Joint kinetics are fundamental in understanding the underlying mechanisms behind movement patterns because they represent the net effect of all agonist and antagonist muscle activity, as well as absorption and generation of power in a joint, and therefore represent specific motor patterns at a joint (Winter, 1984).

Joint moments provides insight regarding the net joint torque generated about the joint, which is commonly used to quantify joint demand and describe compensatory movement patterns (Sanderson & Martin, 1997; Powers et al., 1998; Bateni & Olney, 2002). For example, patients with TTA achieve forward progression by increasing the hip extensor moment on the amputated limb, which is interpreted as increased demand of the hip extensor musculature (Winter & Sienko, 1988; Sanderson & Martin, 1997; Bateni & Olney, 2002). Joint power, which is calculated as the scalar product between a joint's moment and angular velocity, quantifies when a joint is generating (positive power indicates concentric (shortening) muscle contraction) or absorbing (negative power indicates eccentric (lengthening) muscle contraction) power (Winter, 1984; Carollo & Matthews, 2009). For example, patients with TTA absorb less power on the amputated limb knee joint, which is interpreted as a quadriceps avoidance strategy due to lack of eccentric quadriceps control, and is hypothesized to improve stability by avoiding concentric contraction of the knee extensors during weight acceptance (Sadeghi et al., 2001).

2.2.3 *Dynamic Electromyography (EMG)*

EMG is defined as the study of muscle electrical activity and is used to provide insight regarding the motor control behind voluntary movements by assessing the muscle and neurologic function (Robertson et al., 2004). Muscle activation is accomplished by the central nervous system sending an action potential along the motor neuron to innervate at the neuromuscular junction. Once innervated, a sequence of electro-chemical events occurs as the motor unit action potential propagates along the muscle fiber

membrane. These electro-chemical events, commonly referred to as the cross-bridge theory, create active muscle force generation (Lieber, 2002). Measurement of these electro-chemical events during a dynamic activity (e.g. gait) are most commonly performed using surface or indwelling wire electrodes (Perry & Burnfield, 2010).

The overall goal of dynamic EMG in an instrumented gait analysis setting is to identify atypical or unnecessary muscle activity magnitude and timing, and determine if this is responsible for the dysfunctional movement patterns identified using kinematics or kinetics. This is accomplished by identifying motor control compensations and neurologic function using: muscle demand/force generation, activation timing, and fatigue (Perry & Burnfield, 2010; Robertson et al., 2004). For example, patients with TTA adopt a motor control strategy to stiffen the amputated limb knee joint by increasing the co-contraction of the quadriceps and hamstrings, which is hypothesized to improve stability during single limb stance (Seyedali et al., 2012). Increased and prolonged generation of EMG signals compared to a healthy reference group are commonly linked to movement pathologies that may have consequential effects through the development of chronic pain (e.g. myalgia).

2.3 Observational Gait Analysis

Observational gait analysis is a primary tool used by clinicians to guide decision making in rehabilitation by evaluating pathologic gait patterns, identifying areas in need of targeted intervention, and observing and evaluating the longitudinal effects of treatment (Lord et al., 1998). The two primary forms of observational gait analysis are performed through real-time observation and video based observation, and patient

outcomes are then subsequently quantified using clinical measures of functional performance. The accuracy of observational gait analysis is assessed using four components: 1) validity (the degree to which the observation observed reflects the actual event), 2) reliability (the repeatability of the observation), 3) sensitivity (the ability of the observation to identify deviations from normal), and 4) specificity (the ability of the observation to identify no change in normal gait compared to a previous observation) (Toro et al., 2003).

2.3.1 Real-Time Observation

There are multiple observational gait analysis scales that are completed in real-time using naked eye observation that have wide ranges of validity, reliability, sensitivity, and specificity; thus, varying in their acceptance as everyday tools used by clinicians during rehabilitation.

The Rancho Los Amigos National Rehabilitation Center developed what is widely considered the most well-formed and well-known approach to real-time observational gait analysis (Perry & Burnfield, 2010). This systematic approach to observational gait analysis discretizes the gait cycle into eight phases: 1) initial contact, 2) loading response, 3) mid stance, 4) terminal stance, 5) pre-swing, 6) initial swing, 7) mid swing, and 8) terminal swing (Figure 2.1). The form is based on the clinician identifying deviations (minor or major) from normal gait in rotations of the trunk, pelvis, hip, knee, ankle, and toes in all three planes throughout the eight phases of the gait cycle. However, the level of deviation is highly subjective across therapists and does not provide a sensitive scale to detect subtle postural changes throughout the gait cycle, which may have a large impact

throughout the musculoskeletal system (Coutts, 1999). In addition to this approach, there are a variety of other naked eye observational scales including the Waterloo Gait Profile Form (Winter, 1985), Benesh Movement Notation (Harrison et al., 1992), the Rivermead Visual Gait Assessment scale (Lord et al., 1998), the Physician's Rating Scale (Koman et al., 1994), the Observational Gait Scale (Boyd & Graham, 1999).

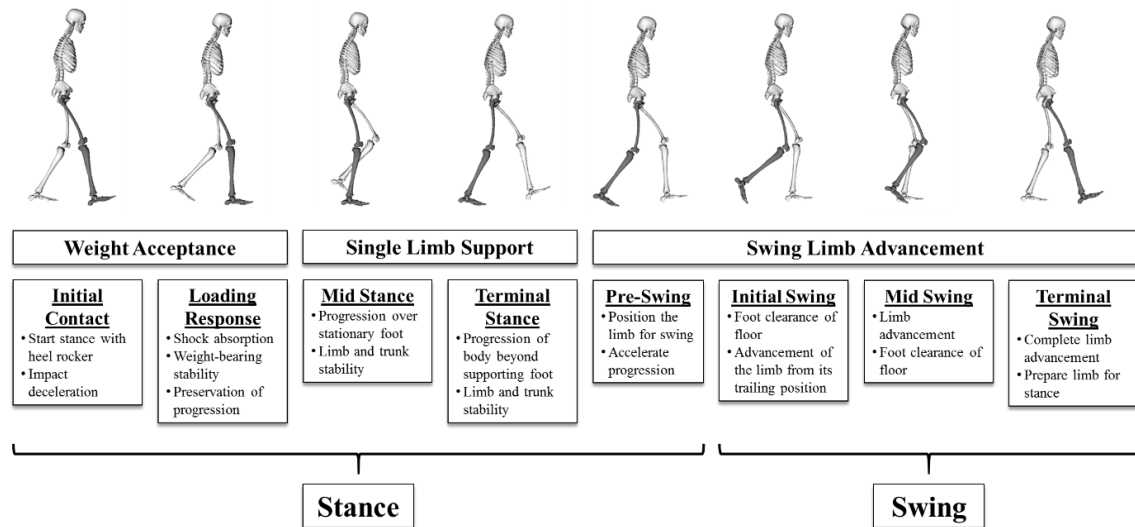


Figure 2.1. Eight phases of the gait cycle, with the objective of each phase listed, defined by Perry & Burnfield (2010).

Although these scales are designed to diagnose and document pathologic gait patterns, the foundational shared limitation of all real-time observational gait analyses are low sensitivity and reliability, due to the inability to discriminate high frequency accelerations, as well as the various levels of training and expertise across observers. For example, late toe rocker during terminal stance or inadequate toe clearance during the swing phase are used by prosthetist to identify prosthetic misalignment, which have shown to be identified differently between novice and trained observers (Saleh &

Murdoch, 1985). Because the human gait cycle is rapidly repeating, video based observation is often used in conjunction with real-time observational analyses to improve the sensitivity of gait assessment (Toro et al., 2003).

2.3.2 *Video-Based Observation*

Video-based observational gait analyses tools have become more common in a clinical setting because they allow for cost efficient technology and do not require the specialized personal that instrumented gait analysis does. The primary advantages of video-based observation over real-time observations are: clinicians can view the same gait cycle a repeated number of times, the gait cycle can be viewed in slow motion, and the recordings can be stored for longitudinal analyses to document effects of interventions to track patient progress (Keenan & Bach, 1996).

In addition to body segment position and postures, video-based observational gait analyses are also capable of calculating gait parameters that are commonly calculated with traditional instrumented gait analyses tools (e.g. spatiotemporal parameters and kinematics). Slow-motion video recordings have been used to divide the gait cycle into stance and swing periods (Wall & Crosbie, 1997) as well as quantify spatiotemporal gait patterns (Stillman & McMeeken, 1996). In addition, the Rivermead Video-Based Clinical Gait Analysis Method comprises video recordings and a computer program to determine hip, knee, and ankle sagittal plane joint angles. Due to the recent technologic advances, integrating visual-based recordings with simple code allows researchers and clinicians to quickly, efficiently, and accurately determine kinematics using personal laptops and tablets. For example, Dartfish (Alpharetta, GA) is a program designed to record and

identify movement patterns that can be altered or improved to enhance performance, and has been validated against a Vicon Motion Capture system to have less than a 5 mm difference in marker trajectories (Eltoukhy et al., 2012).

Findings from observational analyses (both real-time and video-based) alone do not provide clinicians a complete understanding of the severity of the movement dysfunction that affects physical function. Therefore, these are often used in conjunction with additional clinical tests and measures from the patient evaluation session (muscle strength testing, range of motion, etc.) to make a diagnosis and design a treatment plan.

2.3.3 Functional Performance Measures

Clinical measures of functional performance are commonly used to quantify functional movement performance (Steffen & Hacker, 2002). Clinicians use results from functional performance measures as a foundational tool to assess the severity of the movement dysfunction, which guides diagnostic and rehabilitation planning to assess the effects of intervention.

Measured walking velocity (self-selected and fast) assesses the time it takes a patient to walk a set distance (<50 meters). A clinician will instruct the patient to vary their gait speed which allows the assessment of their ability to increase or decrease walking speed to be made, which is required to adapt for external and varying environments and task demands that the patient will be required to complete for community ambulation (Steffen & Hacker, 2002; Bennell et al., 2011).

Timed walk tests (2-minute, 5-minute, and 12-minute walk) are reliable and valid tests to measure function in pathologic patients (Brooks et al., 2001; Datta et al., 1996;

Resnik & Borgia, 2011; Simpson et al., 1982). These tests measure the maximum distance that a patient can walk in the allotted time. These tests are clinically valuable because they are easily administered, represent an activity of daily living, and are generally not extremely strenuous to the patient (Steffen & Hacker, 2002).

The timed up and go test (TUG) measures the time a patients takes to rise from a seated position, walk 3-meters, turn 180°, and return to the original seated position without physical assistance (Mathias et al., 1986). Clinicians rate the patient on a scale based on their perception of the risk of patient fall throughout the TUG. The TUG has been modified to also be a measure of basic mobility skills by documenting the completion time of the task (Podsiadlo & Richardson, 1991) and has been shown to be a reliable test for assessing basic mobility, strength, balance, and agility (Shumway-Cook & Brauer, 2000; Ng & Hui-Chan, 2005; Bennell et al., 2011).

The Stair Climb Test (SCT) is a clinical test that assesses the ability to ascend and descend a flight of stairs, as well as assessing lower extremity strength, power, and balance (Powers et al., 1997; Schmalz et al., 2007; Bean et al., 2007; Bennell et al., 2011). The SCT is considered to be clinically relevant as it relates a patients leg power and mobility in a high-demand activity of daily living (Bean et al., 2007). Common scales of assessment for the SCT are: the number of steps taken, ability to complete task requirement (ascent/descent only or ascent/descent combined, or completion time (Bennell et al., 2011).

Clinical measures of functional performance are designed to relate to the ability of a patient to perform activities of daily living (Terwee et al., 2006) and are most commonly

scored through timing, observational counting, or distance measures. Although these measures have greater reliability than observational analyses alone, and are easily implemented in a clinical setting, they do not provide insight regarding the mechanisms behind the movement pattern observed.

2.4 Lack of Translation between Instrumented and Observational Analyses

Currently, the translation between instrumented and observational gait analyses is not well defined, which is primarily attributed to cost limitations (both equipment and personal). The equipment required for a comprehensive gait analysis laboratory consists of: infrared motion capture system (with accompanying software), force platforms, and an EMG system, as well as the corresponding supplies (e.g. reflective markers, electrodes, etc.); thus exceeding costs of upwards of hundreds of thousands of dollars, which is rarely reimbursed through insurance (Wren et al., 2011). Additionally, instrumented gait analysis is also computationally expensive, requiring processing times ranging from 8 to 12 hours for one patient (Carollo & Matthews, 2009). Clinicians commonly argue that while instrumented gait analysis is a valuable tool for research, it adds unnecessary costs that do not provide any proven benefits to individual patients (Wren et al., 2011). Therefore, instrumented gait analysis is currently impractical in the vast majority of clinical settings.

Current advancements in technology provide the opportunity to introduce instrumentation in an observational setting using cost efficient wearable sensors. Quantitative data when paired with a clinician's intuition can help improve the efficacy of diagnostics and rehabilitation planning by improving observer sensitivity, identifying

targets for intervention, and tracking of patient progress. However, a feasible method to quantify movement that is applicable in both instrumented and observational settings currently does not exist.

2.5 Momentum as a Link between Instrumented and Observational Gait Analyses

The potential to use momentum as a measurement to link instrumented and observational movement analyses arises from its role as a foundational variable in Newton-Euler mechanics and its intuitive nature in qualitative observational analysis. During observational gait analysis, a clinician observes segmental motion, and combines observation with an intuitive sense of the segment masses and mass distributions of the person. For example, consider two able-bodied people walking across the room at the same speed, one small statured with a low body mass index (BMI), the other large statured with a high BMI. Each person can have similar segmental kinematics (which are observable) due to the same gait speed; however, it's clear that the differing statures and masses require different levels of muscle effort and joint demand to achieve forward progression between the two subjects.

Likewise, momentum is a foundational concept and quantity on which Newton-Euler mechanics is based. Motion of an object is fully described by its position and velocity, which is defined as the state of the object. By connecting the state with the object's mass and inertia, momentum is obtained. The most common statement of Newton's Law of translational motion relates the applied forces to motion as $F=ma$; however, this law is properly stated in terms of momentum:

“The sum of all forces applied to an object are proportional to the change in *translational momentum* of the object.”

where translational momentum is the velocity of an object scaled by its mass. Applying this statement to an observational clinical analysis, the clinician observes generation (increasing change) or arresting (decreasing change) of translational momentum, and then interprets these in light of the unobservable forces that cause motion (muscle effort) or restrain motion (joint demand).

Although the aggregate effect of walking is translational (moving from point A to point B), we must consider segment rotation caused by joint moments, and how this is observed and interpreted in light of biomechanics. Humans create all movement by coordinating the rotational motion of segments relative to other segments about joints (Kadaba et al., 1990). When a segment with mass rotates and translates, angular momentum occurs, and can be used to indirectly infer joint moments, which can be estimated using instrumented analysis, but are inherently unobservable in observational analysis.

Similar to Newton’s Law, Euler’s Law of rotational motion relates the applied forces and moments to motion through a statement of momentum:

“The sum of all moments applied to an object are proportional to the change in the *total angular momentum* of the object.”

Although total angular momentum is not a term commonly used outside of physics and engineering, it is easily related to observable motion, and encompasses both translation and rotational motion.

Total angular momentum of an object is composed of two individual components of angular momentum that result from the rotation of the object relative to a reference point as well as the rotation about its center of mass (COM) (Kasdin & Paley, 2011), which is defined as translational angular momentum (TAM) and rotational angular momentum (RAM), respectively. When mathematically modeling the human body, it is common to model each segment as a rigid body object (Hanavan, 1964). In biomechanics, the TAM of a segment is the angular momentum of a segment COM relative to a chosen reference point, and the RAM of a segment is the angular momentum about its own COM.

Considering the changes of TAM and RAM, which are independent components of angular momentum, provides insight into the unobservable biomechanics of a segment. The change in TAM over time of a segment is roughly proportional to net force applied to the segment at the joints and by gravity (referred to as Newton's Law in angular momentum form). The change in RAM over time is roughly proportional to the net moment provided by the muscles and connective tissue at each end of the segments. These relationships are implicitly applied in observational analysis, and explicitly measured in instrumented analysis.

2.6 Conclusion

Instrumented gait analysis combines multiple sources of measurement of kinematics, kinetics, and EMG which provides insights regarding the mechanisms behind the observable movement that is occurring. However, it requires expensive equipment, computational time, and trained personnel; and therefore, is rarely used in a clinical setting. Observational gait analysis is efficient and practical to provide immediate

feedback from the clinician to the patient during movement retraining. Using one or more of the observational gait analysis tools previously described, the clinician relies largely on their intuition to identify the area in need of targeted intervention, which creates low validity, reliability, and sensitivity (Coutts, 1999; Eastlack et al., 1991; Krebs et al., 1985). Developing a method that bridges these approaches has vast clinical implications, beginning with improving the overall efficacy of clinical diagnoses in patients with movement pathologies.

CHAPTER 3: A REVIEW OF MOVEMENT COMPENSATIONS AND THE ASSOCIATED SECONDARY PAIN CONDITIONS IN PATIENTS WITH TRANSTIBIAL AMPUTATION

3.1 Introduction

The number of patients with amputation in the United States is expected to exceed 3 million by the year 2050, which is primarily a result of an aging population with dysvascular pathologies (e.g. diabetes mellitus) (Ziegler-Graham et al., 2008). Specifically, over 80% of all lower-limb amputations are dysvascular (Dillingham et al., 2002) and are more common in adults over the age of 65 (Margolis et al., 2011). In addition to the amputation, patients with dysvascular amputation have multiple comorbidities that lower physical function, resulting in 40-50% of all patients to have limited physical function one year following amputation (Davies & Datta, 2003)

Patients with unilateral transtibial amputation (TTA) must adopt movement compensations to overcome the functional loss of the ankle plantar flexors, which are the primary muscles that contribute to forward propulsion, support, and swing initiation (Zajac et al., 2003). Evaluation of movement compensations adopted by patients with TTA is accomplished through instrumented (spatiotemporal parameters, inverse kinematics, inverse dynamics, electromyography) and observational (timing and events of postures) movement analyses (Smith et al., 2004). Movement compensations are

associated with asymmetric kinematic and loading patterns, which predispose patients with amputation to additional comorbidities that are associated with poor physical function, such as osteoarthritis (Norvell et al., 2005; Morgenroth et al., 2011) and low back pain (LBP) (Ehde et al., 2001). The objective of this chapter is to describe common movement compensations adopted by patients with TTA that are identified through instrumented and observational analyses. In addition, this chapter also describes common secondary pain conditions as a result of movement compensations adopted by patients with TTA that affect physical function. For continuity with the patient population described in Chapters 5-7, the movement compensations described in this review are of patients with unilateral TTA amputation who have passive or dynamic response prostheses (powered prostheses are excluded from the analyses).

3.2 Movement Compensations Identified through Instrumented Analysis

3.2.1 Spatiotemporal Movement Compensations

Spatiotemporal gait analysis quantifies the spatial and temporal patterns of gait, primarily through walking velocity, cadence, step length, step width, and stance time. It is well documented that patients with lower-limb amputation have less efficient gait patterns than their able-bodied peers (Gitter et al., 1995; Hoffman et al., 1997), which are consistently characterized by asymmetric spatiotemporal gait parameters (Schulz et al., 2010). Walking velocity is a common metric used to measure functional performance and can be measured using instrumented walkways, optical gates, a treadmill, or motion capture systems. Patients with unilateral TTA often do not demonstrate significantly different walking speeds when compared to healthy controls (Hafner et al., 2002), which

is likely explained by the wide variability of speeds achieved by patients with TTA, ranging from 0.75 m/s (Barth et al., 1992) to 1.21 m/s (Schulz et al., 2010). This large variability is most likely due to very different etiologies among patients (type of amputation, additional comorbidities, age, etc.) (Nielsen et al., 1988; Snyder et al., 1995). Patients with unilateral TTA have greater stance time on their intact limb (Barth et al., 1992; Sanderson & Martin, 1997; Isakov et al., 2000; Mattes et al., 2000; Sadeghi et al., 2001; Grumillier et al., 2008), which has been suggested to be a result of a protective motor strategy due to a feeling of instability on the amputated limb (Powers et al., 1998; Nolan et al., 2003). In order to increase levels of stability, patients with TTA take wider steps (Hak et al., 2013; Highsmith et al., 2010), thus increasing the hip abduction angles on the amputated limb (Molina-Rueda et al., 2014) which has shown to increase levels of patient-reported stability during walking (Hof et al., 2005). Patients with TTA also demonstrate shorter step length on the amputated limb, which is likely explained by the lack of propulsion from the ankle plantar flexors (Sadeghi et al., 2001; Schulz et al., 2010). The loss of ankle function reduces neuromuscular control required for correct foot placement on the amputated limb (IJmker et al., 2014). Although these compensations are a necessary result of the loss of ankle function, it has previously been shown that asymmetric spatiotemporal gait parameters are linked to adverse effects on the musculoskeletal system. For example, increased step width increases the metabolic cost of walking (Donelan et al., 2001), and alters the loading patterns at the ipsilateral knee and contralateral hip (Simic et al., 2011). Therefore, increased muscle demand is required

in order to shift the body center of mass greater mediolateral distances (Wezenberg et al., 2011).

3.2.2 Kinematic Movement Compensations

Patients with unilateral TTA adopt common segmental and joint kinematic movement compensations due to the loss of the ankle function that range from the knee joint to the trunk segment, that are used to assist forward progression in the sagittal plane and maintain lateral balance in the frontal plane. During walking, patients with TTA demonstrate higher knee flexion angles throughout the gait cycle on the amputated limb in comparison to the intact limb (Isakov et al., 2000; Bateni & Olney, 2002). Due to increased knee flexion, patients with TTA must also increase hip flexion angles on the ipsilateral limb to maintain an upright posture (Bateni & Olney, 2002). Increased hip flexion angles on the amputated limb is also coupled with a forward trunk lean, which is a common segmental posture adopted by patients with TTA to place the body COM beneath the stance foot and reduce the demand placed on the knee extensors by redirecting the ground reaction force vector through the knee joint (Michaud et al., 2000; Powers et al., 1998; Torburn et al., 1990). Therefore, this is commonly suggested to be a quadriceps avoidance strategy adopted to reduce the demand on the knee extensors (Sanderson & Martin, 1997; Winter & Sienko, 1988). In the frontal plane, patients with TTA adopt asymmetric pelvic obliquity patterns during loading (Michaud et al., 2000). It has previously been suggested that pelvic obliquity during loading functions as a loading response during healthy walking, and that an increase in loading may result in an increase in pelvic obliquity, which is defined as a rise of the pelvis opposite the stance limb (Gard

& Childress, 1997; Perry & Burnfield, 2010). Therefore, because patients with TTA walk with reduced loading on the amputated limb (Sanderson & Martin, 1997; Nolan et al., 2003), pelvic obliquity is decreased during loading response of the amputated limb (Michaud et al., 2000; Molina-Rueda et al., 2014). In addition, during single limb stance, patients with TTA increase pelvic obliquity in order to achieve swing limb foot clearance in absence of ankle function, which is a strategy commonly referred to as “hip hiking” (Michaud et al., 2000). During weight acceptance on the amputated limb, patients with TTA increase the lateral bending of the trunk segment towards the amputated limb, which is consistent with a compensated Trendelenburg gait pattern (Molina-Rueda et al., 2014). This movement compensation has been hypothesized to occur to reduce the lever arm and compensate for dysfunctional or weakened hip abductors, as similarly seen in patients with transfemoral amputation or hip osteoarthritis (Watelain et al., 2001; Goujon-Pillet et al., 2008).

3.2.3 Kinetic Movement Compensations

The investigation of joint demand, often quantified by joint kinetics (joint moments and power), is a common tool used to investigate movement in both healthy and pathologic gait because asymmetric joint kinetics have been linked to the development of a variety of overuse injuries, including low back pain (LBP) (Kumar, 2001) and osteoarthritis (Miyazaki et al., 2002). Because joint moments represent the net effect of all agonist and antagonist muscle activity that span a joint, joint kinetics are commonly interpreted as the demand placed at the joint and describe movement compensations (Bateni & Olney, 2002; Powers et al., 1998; Sanderson & Martin, 1997). Patients with

TTA reduce the load placed on the amputated limb by reducing the ground reaction force (GRFs) throughout loading (Arya et al., 1995; Nolan & Lees, 2000; Sanderson & Martin, 1997). This alteration in GRF propagates through the kinetic chain altering joint demand in patients with TTA. In the sagittal plane, patients with TTA increase hip extensor moments during the stance period (Bateni & Olney, 2002; Sadeghi et al., 2001; Silverman et al., 2008) in order to generate forward propulsion in absence of ankle function. In addition, patients with TTA reduce the first knee extensor moment during loading, which has been hypothesized to reduce the demand placed on the knee extensors (quadriceps avoidance strategy) (Winter & Sienko, 1988; Czerniecki et al., 1991; Gitter et al., 1995; Powers et al., 1998; Bateni & Olney, 2002). In the frontal plane, patients with TTA reduce hip abductor moments during the stance period of the amputated limb (Underwood et al., 2004; Royer & Wasilewski, 2006; Rueda et al., 2013; Molina-Rueda et al., 2014). In both the frontal and sagittal planes, patients with TTA have greater low back moments in comparison to healthy controls, which was hypothesized to be a result of compensated Trendelenburg gait posture and weakened hip abductors (Hendershot & Wolf, 2014).

3.2.4 Neuromuscular Movement Compensations

Electromyography (EMG) is a common clinical and biomechanical tool used to investigate the motor control patterns adopted during movement. Lower extremity muscle demand has been shown to be consistent across healthy adults (Winter & Yack, 1987; Kadaba et al., 1990; Neptune et al., 2008; Perry & Burnfield, 2010). Although different than healthy controls, patients with TTA also demonstrate consistent EMG patterns

during walking (Winter & Sienko, 1988; Beyaert et al., 2008; Fey et al., 2010). Patients with TTA increase the demand of muscles surrounding the amputated limb knee joint by increasing the magnitude and duration of the uniarticular knee extensors (Culham et al., 1986; Winter & Sienko, 1988; Torburn et al., 1990; Pinzur et al., 1991; Powers et al., 1998; Rietman et al., 2002) and biarticular hamstrings (Culham et al., 1986; Winter & Sienko, 1988; Torburn et al., 1990; Pinzur et al., 1991; Powers et al., 1998; Isakov et al., 2000; Isakov et al., 2001; Rietman et al., 2002; Schmalz et al., 2007). Increased co-contraction of muscles surrounding the knee joint is thought to be a limb-stiffening compensation strategy to increase levels of stability on the amputated limb (Seyedali et al., 2012). In addition, patients with TTA also increase the demand of the gluteus maximus on the amputated limb (Winter & Sienko, 1988; Torburn et al., 1990). It has also been shown that muscle demand on the intact limb of patients with TTA has been shown to be similar to healthy controls (Culham et al., 1986; Czerniecki, 1996; Rietman et al., 2002)

3.2.5 Angular Momentum Movement Compensations

In recent years, the use of whole-body angular momentum (WBAM) has become a common metric to describe dynamic balance because it has been shown to be regulated (generated and arrested) differently during normal and pathologic movements (Herr & Popovic, 2008; Bruijn et al., 2011). WBAM is dependent upon the mass, inertia, and linear and rotational velocities of all body segments with respect to the body center of mass. Herr & Popovic (2008) proposed that WBAM is “highly regulated” (i.e. minimized) by the central nervous system (CNS) in the sagittal, frontal, and transverse

planes during walking. Therefore, if WBAM is minimized during healthy movements, deviations from zero (positive and negative) and the range between the deviations may provide insight into the risk of falling and dynamic stability during walking. As a result, patients with TTA are less able to regulate WBAM compared to able-bodied individuals because ankle muscles are the primary mechanism of regulating WBAM (Pijnappels et al., 2004; Neptune & McGowan, 2011; Neptune & McGowan, 2016). This can be observed in a variety of different dynamic tasks including: over ground level walking (Silverman & Neptune, 2011; D'Andrea et al., 2014), sloped walking (Silverman et al., 2012; Pickle et al., 2016) and stair climbing (Pickle et al., 2014). Although WBAM provides quantitative insight into levels of dynamic balance in patients with TTA, it doesn't not provide insight regarding the segmental strategies that are used to regulate WBAM. Information regarding patterns of generating and arresting segmental angular momentum could be used to inform diagnostics and guide movement retraining following amputation, thus improving the overall efficacy of rehabilitation.

3.3 Movement Compensations Identified through Observational Analysis

Observational gait analysis is a valuable tool used by clinicians and prosthetists to identify gait deviations and correct them via movement retraining or socket alignment adjustment (Perry & Burnfield, 2010). Clinicians most commonly use the systematic observational approach of dividing the gait cycle into eight phases that contain 13 critical events to identify gait deviations (Perry & Burnfield, 2010). Gait deviations, which are generally associated with a movement compensation, occur when one or more of these critical events deviate from normal or do not occur. Observational analysis does not

provide insight regarding the mechanism behind gait deviations; and therefore the clinician/prosthetist uses their intuition and training to diagnose the mechanism of gait deviation. The gait deviations described below are common observations of amputee gait that are hypothesized to be a result of a habitual movement pattern (movement pathology, muscle weakness, etc.) (Powers et al., 1996) or an error in prosthesis alignment (Fridman et al., 2003; Kobayashi et al., 2012; Boone et al., 2013).

During weight acceptance (initial contact and loading response), patients with TTA demonstrate atypical knee flexion (excessive or insufficient) (Berger, 2002). In healthy gait, knee flexion angles during weight acceptance are between 15-20° (Perry & Burnfield, 2010); and therefore flexion angles that are not within this range are identified as atypical. Excessive knee flexion angles during weight acceptance in patients with TTA is hypothesized to occur primarily due to prosthesis socket alignment: excessive anterior tilt of the socket, excessive anterior displacement of the prosthesis socket relative to the prosthesis foot, excessive prosthesis foot dorsiflexion, or excessively stiff heel cushion (Berger, 2002; Chow et al., 2006). Reduced or absent knee flexion during weight acceptance is hypothesized to occur due to: excessive posterior displacement of the prosthesis socket relative to the prosthesis foot, excessively soft heel cushion, insufficient socket flexion, excessive prosthesis foot plantarflexion, or quadriceps weakness (Berger, 2002; Chow et al., 2006; Powers et al., 1998). During weight acceptance and single limb stance, patients with TTA demonstrate excessive lateral trunk bending toward the stance (amputated) limb, which is hypothesized to occur due to a short prosthesis, insufficient socket support, weakened ipsilateral hip abductors, or residual limb stump pain (Hillman

et al., 2010). Patients with TTA also adopt a forward trunk lean throughout loading of the amputated limb, which is hypothesized to reduce the load placed on the quadriceps (Esquenazi, 2014).

During midstance, patients with TTA demonstrate excessive mediolateral thrust of the prosthesis, which is hypothesized to occur due to an abducted socket, excessive mediolateral dimensions of the prosthesis socket, or mediolateral prosthesis foot placement error (Berger, 2002). This prosthesis thrust therefore leads to an early increase of amputated limb knee flexion between midstance and toe-off, which is hypothesized to reduce the demand placed on the quadriceps (Saleh & Murdoch, 1985).

Movement compensations adopted by patients with amputation identified using observational analysis do not provide insight regarding the mechanisms behind the movements (Rose, 1983). In order for accurate diagnoses and rehabilitation planning to be made, a true understanding of the entire movement system must be made in order to permanently correct pathologic movement patterns (Coutts, 1999).

3.4 Comorbidities and Secondary Pain Conditions Associated with Dysvascular

TTA

Although movement compensations are necessary for patients with TTA to adopt in the absence of active ankle plantarflexion, they increase the risk of comorbidities through the development of secondary pain conditions that are attributed to overuse injuries. Specifically, movement compensations adopted by TTA increase the mechanical work at the proximal joints on the ipsilateral limb and contralateral limb (Beyaert et al., 2008; Grumillier et al., 2008; Sagawa et al., 2011; Winter & Sienko, 1988). Therefore,

comorbidities associated with increased and asymmetric joint loading patterns are a common result in this population, such as osteoarthritis in the hip and knee joints (Norvell et al., 2005; Struyf et al., 2009; Morgenroth et al., 2011), ulcerations of the intact foot (Johannesson et al., 2009), and LBP (Ehde et al., 2001). Of these over-use injuries, the prevalence of LBP is substantially higher compared to the general able-bodied population (52-71% compared to 6-33%) (Smith et al., 1999; Ehde et al., 2001). Although more common, the development of LBP within this population still remains idiopathic. Therefore, further understand of how movement compensations could potentially contribute to the development of LBP could help develop more targeted preventative movement retraining approaches following amputation.

In addition to these comorbidities associated with the amputation, patients with dysvascular TTA are generally older in age (Ziegler-Graham et al., 2008) and has a higher prevalence of obesity (Rosenberg et al., 2012), both of which lower levels of physical function, compared to younger patients with traumatic amputation. In addition to comorbidities associated with maladapted movement compensations, up to 85% of all patients with lower limb amputation suffer from phantom and residual limb pain, which has been associated with increased rates of morbidity (Sherman, 1997). Phantom limb pain is defined as pain that is perceived to originate from the missing limb; whereas residual limb pain is defined as pain originating in the residual portion of the amputated limb (Ehde et al., 2000). As a result of all comorbidities, it is estimated that 40-50% of all patients dysvascular amputation do not have physical function levels required for community ambulation after one year following the amputation (Davies & Datta, 2003).

This decline in physical function likely creates the increase in body mass within the first 1-3 years following amputation (Rosenberg et al., 2012), thus furthering the decline of physical function within this population.

3.5 Conclusion

Following dysvascular TTA, rehabilitation is primarily focused on interventions immediately following the amputation, that are centered on prosthetic function and gait/mobility function (Yiğiter et al., 2002; Klute et al., 2006; Gailey, 2008). However, this method does not take into account the underlying complexity of physical problems, such as obesity, that likely attributed to the amputation. In addition, patients with TTA are often not instructed on how to compensate in tasks other than walking during the acute care phase of rehabilitation, which is required for community ambulation. As a result, potential consequential movement compensations adopted by patients with dysvascular TTA may become habitual over long-term prosthetic use, creating long-term physical limitations within this population (Davies & Datta, 2003; Nehler et al., 2003; van Velzen et al., 2006). Accurate identification of potential consequential movement patterns during movement retraining following amputation in a variety of tasks required for community ambulation could help in the prevention of comorbidities that develop due to long-term overuse injuries.

CHAPTER 4: SEPARATION OF ROTATIONAL AND TRANSLATIONAL SEGMENTAL MOMENTUM TO ASSESS MOVEMENT COORDINATION DURING WALKING

4.1 Abstract

This investigation presents an analysis of segmental angular momentum to describe segmental coordination during walking. Generating and arresting momentum is an intuitive concept, and also forms the foundation of Newton-Euler dynamics. Total segmental angular momentum is separated into separate components, translational angular momentum (TAM) and rotational angular momentum (RAM), which provide different but complementary perspectives of the segmental dynamics needed to achieve forward progression during walking. TAM was referenced to the stance foot, which provides insight into the mechanisms behind how forward progression is achieved through coordinated segmental motion relative to the foot. Translational and rotational segmental moments were calculated directly from TAM and RAM, via Euler's 1st and 2nd laws in angular momentum form, respectively, and are composed of the effects of intersegmental forces and joint moments. Using data from 14 healthy participants, the effort required to generate and arrest momentum were assessed by linking the features of segmental angular momentum and the associated segmental moments to well-known spatiotemporal and kinetic features of the gait cycle. Segmental momentum provides an opportunity to explore and understand system-wide dynamics of coordination from an

alternative perspective that is rooted in fundamentals of dynamics, and can be estimated using only segmental kinematic measurements.

4.2 Introduction

Total segmental angular momentum is a foundational concept and quantity on which Newton-Euler mechanics are based. Generating and arresting momentum is an intuitive concept that is broadly and correctly used in nonscientific arenas (e.g., sports); however, in dynamic systems, momentum is primarily used as a stepping stone through which equations of motion are calculated (forward dynamics) or moments and forces are obtained (inverse dynamics). Joint kinetics, which are calculated using an iterative Newton-Euler method via inverse dynamics, are commonly used to describe both normal and pathologic human movement patterns and depend upon the total angular momentum of the surrounding segments (Robertson et al., 2004; Carollo & Matthews, 2009). Joint moments represent the net effect of forces (active muscle forces and passive tissues that cross a joint) that are used to generate and absorb power, and are used as a surrogate representation of joint demand during movement (Winter, 1984). Joint demand is often used to quantify the demands placed on the musculoskeletal system due to external biomechanical loads or muscle forces required for stabilization/segmental motion.

Theoretically, the Newton-Euler formulation on which joint kinetics are calculated provides a direct formulation of how forces and moments regulate segmental momentum. Euler's First Law relates the forces on a segment to motion through the time rate of change of momentum:

$$\mathbf{F}_{seg} = \frac{{}^I d}{dt} ({}^I \mathbf{p}_{seg}) = \frac{{}^I d}{dt} (m_{seg} {}^I \mathbf{v}_{seg}) \quad (4.1)$$

where ${}^I \mathbf{p}_{seg}$ is the linear momentum of a segment (segment mass times linear velocity) observed in an inertial reference frame. Although the aggregate effect of walking is

translational (moving from point A to point B), legged locomotion is accomplished through coordinated segmental rotations relative to other segments about shared axes at the joints, which is driven by joint moments (Kadaba et al., 1990). When a segment with mass rotates and translates, it has angular momentum that is related to external joint forces and moments through Euler's Laws. Similar to Newton's Second Law, Euler's Second Law of rotational motion relates the applied forces and moments to a segment to motion through a statement of momentum:

$$\mathbf{M}_o = \frac{^J d}{dt} (^J \mathbf{h}_o) \quad (4.2)$$

where \mathbf{M}_o is the sum of moments with respect to the inertially fixed point O applied to the segment and $^J \mathbf{h}_o$ is the total angular momentum of the segment with respect to O .

Total angular momentum of a segment is composed of two independent components that result from the rotation of the segment relative to a reference point as well as the rotation about its center of mass (COM) (Kasdin & Paley, 2011), which we label as translational angular momentum (TAM) and rotational angular momentum (RAM), respectively.

Considering the changes of TAM and RAM, which are separate components of segmental angular momentum, provides insight into segmental kinetics.

The change in TAM over time of a segment is roughly proportional to the net external force applied to the segment at the joints (intersegmental forces), at the muscle attachment points on the segment, and by gravity (referred to as Newton's Law in angular momentum form). The change in RAM over time is roughly proportional to the net moment provided by the muscles and connective tissue at each end of the segments.

Forward progression during walking is achieved through both translational and angular motion of individual segments, and therefore segment-based analysis of generating and arresting segmental angular momentum may provide additional insight into how the body coordinates segmental control. Because segmental angular momentum is embedded in inverse dynamic calculations that are commonly used to describe joint demand, we propose that kinetics derived from segmental momentum can provide insight into the effort required during movement (through segmental moments). Two investigations have employed a segmental angular momentum in walking by using principal component analysis (PCA) to examine contributions of total angular momentum of segment relative to the body COM to the sum of total angular momentum from all body segments, known as whole-body angular momentum (WBAM) using. Herr and Popovic (2008) concluded that despite large total segmental angular momentum with respect to the body COM, segment-to-segment cancellations occur to minimize WBAM. Bennett et al., (2010) accounted for synergistic control of segmental angular momenta using three principal components in each plane, and the synergies did not change with the gait speed. Although PCA demonstrates segmental synergies in orthogonal parameter spaces created by directions of variance (principal components), to our knowledge, the actual shapes and patterns of individual segmental angular momenta over time are less commonly reported. Two recent investigations have assessed the relative contributions of grouped segmental momenta (upper and lower body) to WBAM in patients with cerebral palsy (Russell et al., 2011) and patients with amputation (Pickle et al., 2016). Although this approach is useful for identifying strategies for maintaining balance and overall

control of the system, we propose that more detailed analyses of individual segmental angular momentum can provide additional insight into coordinated segmental motion. The identification of individual segmental movement patterns is what is done in a clinical movement retraining setting, but has not been accomplished using individual segmental momenta.

Measurement of segmental angular momentum is relevant to both observational and instrumented analyses because it depends on segment kinematics that can be used to gain inference on joint kinetics via Euler's Laws. Assessment of segmental kinematics is common in both observational and instrumented gait analyses, which are both used to identify movement dysfunction and assess outcomes of interventions that target movement quality (Saleh & Murdoch, 1985; Shull et al., 2014). Although important for guiding clinical reasoning, observational gait analysis lacks diagnostic standardization (particularly outside of level walking) and sensitivity (Toro et al., 2003), which can result in misidentification of compensatory movement patterns (Shores, 1980; Robinson & Smidt, 1981; Holden et al., 1984; Frigo et al., 1998) due to poor observer training, observer bias, parallax error, and poor intrarater reliability (Krebs et al., 1985; Coutts, 1999). Instrumented gait analysis, in contrast to observational gait analysis, is currently the gold standard for accurately quantifying human movement (through the measurement of segment velocities, accelerations, forces, moments, and muscle activity); however, it is not commonly used in a clinical setting due to high monetary, computational, and time expenses. By contrast, angular momentum can be easily measured using wearable sensors. At this time, however, the theoretical foundations of using segmental angular

momentum to provide insight regarding kinetics and the subsequent interpretation of the waveforms to identify movement patterns remains unknown.

The objective of this investigation was to explore the use of segmental angular momentum to describe coordination and effort during over-ground walking. We chose to apply this analysis to walking because gait is the most commonly assessed and taught task in biomechanics, and provides an opportunity to link it to other well-known aspects of gait biomechanics. Total angular momentum of each segment is described using the independent components TAM and RAM, and the segmental moments calculated by the time rate of change of TAM and RAM.

4.3 Dynamic Theory of Separation of Angular Momentum

4.3.1 Selection of the Reference Point

Selecting a reference point is critical for interpreting the translational component of segmental angular momentum. We propose that separate analysis of TAM referenced to the foot in contact with the ground and RAM will provide a unique insight into the mechanisms behind how forward progression is achieved with respect to the stance foot. Reference TAM to the foot during the stance period allows for interpretable insight using segmental kinetics about that point through the application of Euler's Laws of rotational motion.

4.3.2 Mathematical Foundations

The principle of angular momentum separation demonstrates that the total angular momentum (with respect to a chosen point) of a segment is the sum of two independent components: 1) the angular momentum of the segment center of mass (with respect to the

same point), often referred to as the “orbital” component, and 2) the angular momentum about the segment center of mass, often referred to as the “spin” component (Kasdin & Paley, 2011).

Total angular momentum of a segment with respect to the stance foot is the sum of segmental TAM (${}^I \mathbf{h}_{seg/foot}$) and segmental RAM (${}^I \mathbf{h}_{seg}$):

$${}^I \mathbf{h}_{foot} = {}^I \mathbf{h}_{seg/foot} + {}^I \mathbf{h}_{seg} \quad (4.3)$$

TAM of a segment is defined as the angular momentum of the center of mass (*seg*) of the segment relative to the reference point at the stance foot COM (*foot*):

$${}^I \mathbf{h}_{seg/foot} = \mathbf{r}_{seg/foot} \wedge m_{seg} {}^I \mathbf{v}_{seg/foot} \quad (4.4)$$

where $\mathbf{r}_{seg/foot}$ is the position vector of the segment relative to the stance foot, ${}^I \mathbf{v}_{seg/foot}$ is the velocity of the COM of the segment relative to the stance foot as observed in an inertial reference frame I, and m_{seg} is the mass of the segment (Figure 4.1a). RAM of the segment is the angular momentum about its own COM:

$${}^I \mathbf{h}_{seg} = \mathbf{I}_{seg} \cdot \boldsymbol{\omega}_{seg} \quad (4.5)$$

where \mathbf{I}_{seg} is the inertial tensor of the segment and $\boldsymbol{\omega}_{seg}$ is the angular velocity of the segment as observed in the inertial reference frame I (Figure 4.1a).

We can demonstrate the relationship of TAM and RAM to translational and rotational segmental kinetics by taking the time derivative of both sides of Eq. (4.3).

$$\begin{aligned}
{}^J \mathbf{h}_{\text{foot}} &= {}^J \mathbf{h}_{\text{seg/foot}} + {}^J \mathbf{h}_{\text{seg}} \\
\frac{{}^J d}{{}^J dt} ({}^J \mathbf{h}_{\text{foot}}) &= \frac{{}^J d}{{}^J dt} ({}^J \mathbf{h}_{\text{seg/foot}} + {}^J \mathbf{h}_{\text{seg}}) \\
\frac{{}^J d}{{}^J dt} ({}^J \mathbf{h}_{\text{foot}}) &= \frac{{}^J d}{{}^J dt} ({}^J \mathbf{h}_{\text{seg/foot}}) + \frac{{}^J d}{{}^J dt} ({}^J \mathbf{h}_{\text{seg}})
\end{aligned} \tag{4.6}$$

The two independent time derivatives on the right-hand side of Eq. (4.6) provide alternative expressions for Euler's Laws applied to the segment. The time derivative of translational angular momentum is an expression of Euler's 1st Law in angular momentum form:

$$\frac{{}^J d}{{}^J dt} ({}^J \mathbf{h}_{\text{seg/foot}}) = \mathbf{M}_{\text{seg/foot}} + (\mathbf{r}_{\text{seg/foot}} \times m_{\text{seg}} {}^J \mathbf{a}_{\text{foot}}) \tag{4.7}$$

where $\mathbf{r}_{\text{seg/foot}} \times m_{\text{seg}} {}^J \mathbf{a}_{\text{foot}}$ is the corrective inertial moment of the segment relative to the stance foot and is required to satisfy Euler's law when the stance foot accelerates at the end of the stance period.

The right hand side is the translational segmental moment about the stance foot, expressed as:

$$\mathbf{M}_{\text{seg/foot}} = \mathbf{r}_{\text{seg/foot}} \times \mathbf{F}_{\text{seg}}^{\text{Ext}} \tag{4.8}$$

where $\mathbf{F}_{\text{seg}}^{\text{Ext}}$ is the resultant force of all external forces applied to the segment (Figure 4.1b).

The time derivative of rotational angular momentum is the more familiar expression of Euler's 2nd Law:

$$\frac{{}^J d}{{}^J dt} ({}^J \mathbf{h}_{\text{seg}}) = \mathbf{M}_{\text{seg}}^{\text{Ext}} \tag{4.9}$$

The right hand side is the rotational segmental moment expressed as:

$$\mathbf{M}_{seg} = \sum_{i=1}^N \mathbf{M}_i^{Ext} + \sum_{i=1}^N (\mathbf{r}_{i/seg} \times \mathbf{F}_i^{Ext}) \quad (4.10)$$

where i is the distal and proximal locations of forces and moments. In a link segment model of the human body, \mathbf{M}_{seg} is the net moment created by adding sum of the applied (external) proximal and distal joint moments to the moments about the segment COM due to proximal and distal forces (Figure 4.1b):

$$\mathbf{M}_{seg} = \mathbf{M}_{distal} + \mathbf{M}_{proximal} + (\mathbf{r}_{distal/COM} \times \mathbf{F}_{distal}) + (\mathbf{r}_{proximal/COM} \times \mathbf{F}_{proximal}) \quad (4.11)$$

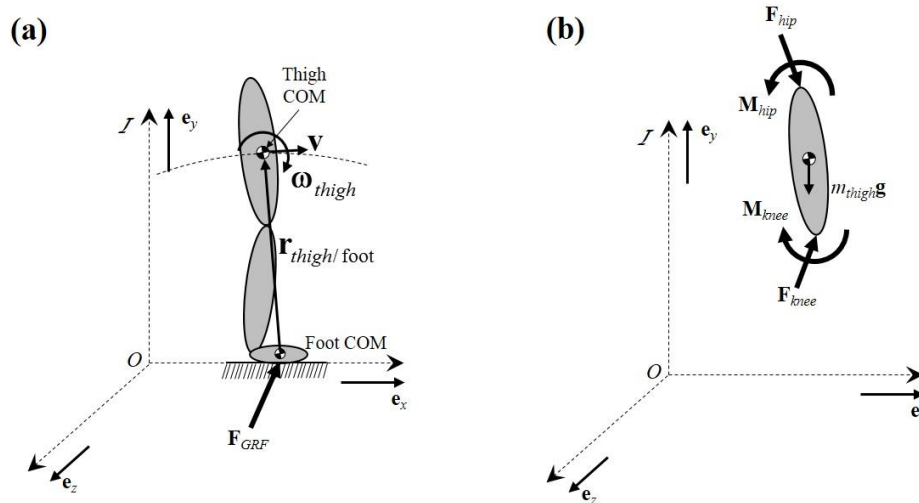


Figure 4.1. (a) An illustration of the stance limb just before midstance and the vectors used to calculate translational angular momentum (TAM) about the Foot. TAM is a cross product of the position vector with the linear momentum of the segment, which can be thought of as the “moment of momentum”. The length of the position vector is relatively invariant during the stance period for the stance limb segments and rotates similar to an inverted pendulum. (b) The free-body diagram of the thigh just before midstance, which shows all the forces and moments applied to the segment. The rotational segmental moment is the net moment about the COM applied (external) created by the hip and knee joint moments and the moments about the segment COM due to hip and knee joint intersegmental forces. For clarity, the net segmental moment is the summation of all moments due to external forces applied to the segment and joint moments. Therefore, it does not provide the detailed breakdown of moments and forces calculated from inverse dynamic analyses.

4.4 Experimental Methods

4.4.1 Participants

Fourteen healthy participants (3F, 11M, age: 61.5 ± 8.4 years, BMI: 25.2 ± 2.8 kg/m²) provided informed consent to the Colorado Multiple Institutional Review Board approved protocol. Further details regarding individual participant anthropometrics and levels of functional performance can be found in Appendices A and B, respectively.

Each participant performed three gait trials at 1 m/s ($\pm 5\%$ measured through gait timers) and was instrumented with 63 reflective markers, sampled at 100 Hz (Vicon, Centennial, CO). Kinematic data were low-pass filtered with a 4th-order Butterworth filter (6 Hz cutoff frequency). A 15-segment participant-specific model was created for each participant (see tables 4.1-4.2 for complete segment list) and used to obtain segment kinematics (Visual 3D, C-Motion, Inc., Germantown, MD) (Figure 4.2). Segment masses were based on percentage of total body weight (Dempster, 1955) and segment inertias were based on segment geometry (Hanavan, 1964).

4.4.2 Calculations

To facilitate anatomically planar analyses that considers the progression of the body through space, all angular momenta were expressed in a basis with respect to the path of the body COM: $\mathbf{e}_{\text{frontal}}$ (tangent to the horizontal path of the body COM), $\mathbf{e}_{\text{transverse}}$ (opposite direction of the gravity), and $\mathbf{e}_{\text{sagittal}}$ ($\mathbf{e}_{\text{frontal}} \times \mathbf{e}_{\text{transverse}}$) (Figure 4.2).

We calculated segmental TAM (Eq. (4.4)) and the translational segmental moment ($\mathbf{M}_{\text{seg}/\text{foot}}$) about the stance foot (Eq. (4.8)) only during the stance period (heel strike to toe off) because the foot accelerates, and becomes non-inertial after toe off. Calculation

of the segment RAM (Eq. (4.5)) and the rotational segmental moment about the segment COM (Eq. (4.10)) was performed during the entire gait cycle because the calculations do not depend on the point of reference being stationary.

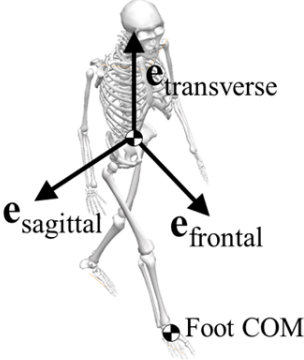
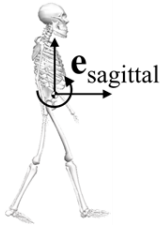
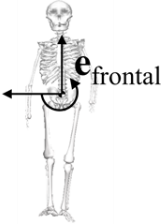
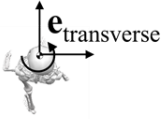
3-Dimensional View	Planar View	Positive Momentum	Negative Momentum
	<p><i>Sagittal Plane</i></p> 	<p>Momentum generated <i>posteriorly</i> with respect to the stance (right) foot</p>	<p>Momentum generated <i>anteriorly</i> with respect to the stance (right) foot</p>
	<p><i>Frontal Plane</i></p> 	<p>Mediolateral momentum generated <i>toward</i> the stance (right) foot</p>	<p>Mediolateral momentum generated <i>away</i> from the stance (right) foot</p>
	<p><i>Transverse Plane</i></p> 	<p>Axial momentum generated <i>away</i> from the stance (right) foot</p>	<p>Axial momentum generated <i>toward</i> the stance (right) foot</p>

Figure 4.2. All momentum and moment vectors were expressed in a basis with respect to the path of the body COM (defined by e_{sagittal} , e_{frontal} , and $e_{\text{transverse}}$ axes) to facilitate planar analyses.

4.4.3 Analysis of Segmental Contributions

Peak values of TAM and the translational segmental moment about the stance foot were identified during stance. Peak values of RAM and the rotational segmental moment about the segment COM were identified over the entire gait cycle. We chose to identify peak values of TAM and RAM, and their time derivatives, because they quantify the period of generating the most momentum (zero to peak value) and period of arresting momentum (peak value to zero), and the associated torques required (time derivatives).

Using the patterns of segmental generation and arresting of TAM and RAM, and the associated segmental torques, we qualitatively describe movement patterns used to achieve forward progression during walking.

4.5 Results and Interpretation

4.5.1 Translational Angular Momentum

Segmental translational angular momentum (TAM) and the translational segmental moments ($\mathbf{M}_{seg/foot}$) demonstrated similar shapes with respect to the stance foot across segments within planes and generally larger magnitudes with increasing distance from the foot in segments in the stance limb and axial skeleton (Table 4.2). The progressively larger magnitudes of superior segments with larger mass corresponds well with the inverted pendulum model of forward progression during gait, which represents the aggregate effect of translation and rotation of all body segments (identified through the body COM trajectory) about the fixed fulcrum point at the stance foot.

In the sagittal plane (Figure 4.3), TAM of all segments is negative at heel strike, which corresponds with generation of anterior translational momentum with respect to the stance foot, and then is amplified during weight acceptance (0-12% of the gait cycle). The generation of anterior TAM during weight acceptance is accomplished by net forces applied to each segment resulting in anterior segment translation with respect to the stance foot. This is likely a result of power generation from the contralateral limb during push-off as the weight is transferred between limbs to achieve forward progression. This increase in external forces applied to each segment from push-off creates a negative translational moment with respect to the stance foot, which coincides with an anterior

(clockwise) moment direction in the path reference frame (Figure 4.2). During the first half of single limb support (12-30% of the gait cycle), TAM of all segments is slightly arrested (less negative), which is accomplished by positive translational segmental moments (creating posterior translation) about the stance foot. The direction of segmental translational moment is a result of a posterior position of the segments with respect to the stance foot and the external vertical force due to gravity, which creates a posterior (counterclockwise) moment direction in the path reference frame (Figure 4.2). It is likely the musculoskeletal system takes advantage of the force due to gravity during early single limb support to prevent continuance or extenuated of “falling forward”. During the second half of single limb support (30-50% of the gait cycle), additional TAM of all segments is slightly generated (decreases), which corresponds with negative translational segmental moments (creating anterior translation) with respect to the stance foot. The translational moment becomes negative as the position vector of the body segments with respect to the foot shifts anteriorly, creating an anterior (clockwise) moment direction in the path reference frame (Figure 4.2). The largest translational moments occur during double support, which is likely due to high intersegmental forces, and remains small throughout single limb support.

In the frontal plane (Figure 4.3), TAM of all segments is positive at heel strike, which corresponds to angular momentum in the direction of the stance limb, and then rapidly decreases (arrested) during weight acceptance (0-12% of the gait cycle), which is accomplished by negative translational moments (creating segment translation away from foot) with respect to the stance foot. The rapid arresting translational angular momentum,

and the associated segmental torques, of the body away from the planted limb is required to arrest lateral translation and maintain the position of the body COM with respect to the medial border of the support foot (Shimba, 1984). Throughout the duration of the stance period, the translational segmental moments are negative (creating frontal plane segment translation away from the stance foot), which is likely due to the moment due to the external gravitational force and the relatively constant moment arms of the segments with respect to the stance foot. This negative (clockwise) translational segment moment in the path reference frame (Figure 4.2) corresponds with the mediolateral trajectory of the of the body COM during the swing limb advancement of the contralateral limb in preparation for weight transfer between limbs (MacKinnon & Winter, 1993; Perry & Burnfield, 2010).

In the transverse plane (Figure 4.3), TAM of all segments is positive at heel strike, corresponding to axial angular momentum away from the stance foot, and rapidly arrested during weight acceptance, which is accomplished by negative translational segmental moments (creating axial segmental rotation toward the stance foot) during weight acceptance (0-12% of the gait cycle). Similar to the sagittal plane, the increased external forces from the power generation of the contralateral limb push off likely create the negative (clockwise) translational segment moment in the path reference frame (Figure 4.2). During single limb support (12-50% of the gait cycle), transverse angular momentum remains constant, no applied segmental translational moments, which is likely a strategy used as a progression mechanism and straight line motion because axial angular momentum is not generated or arrested.

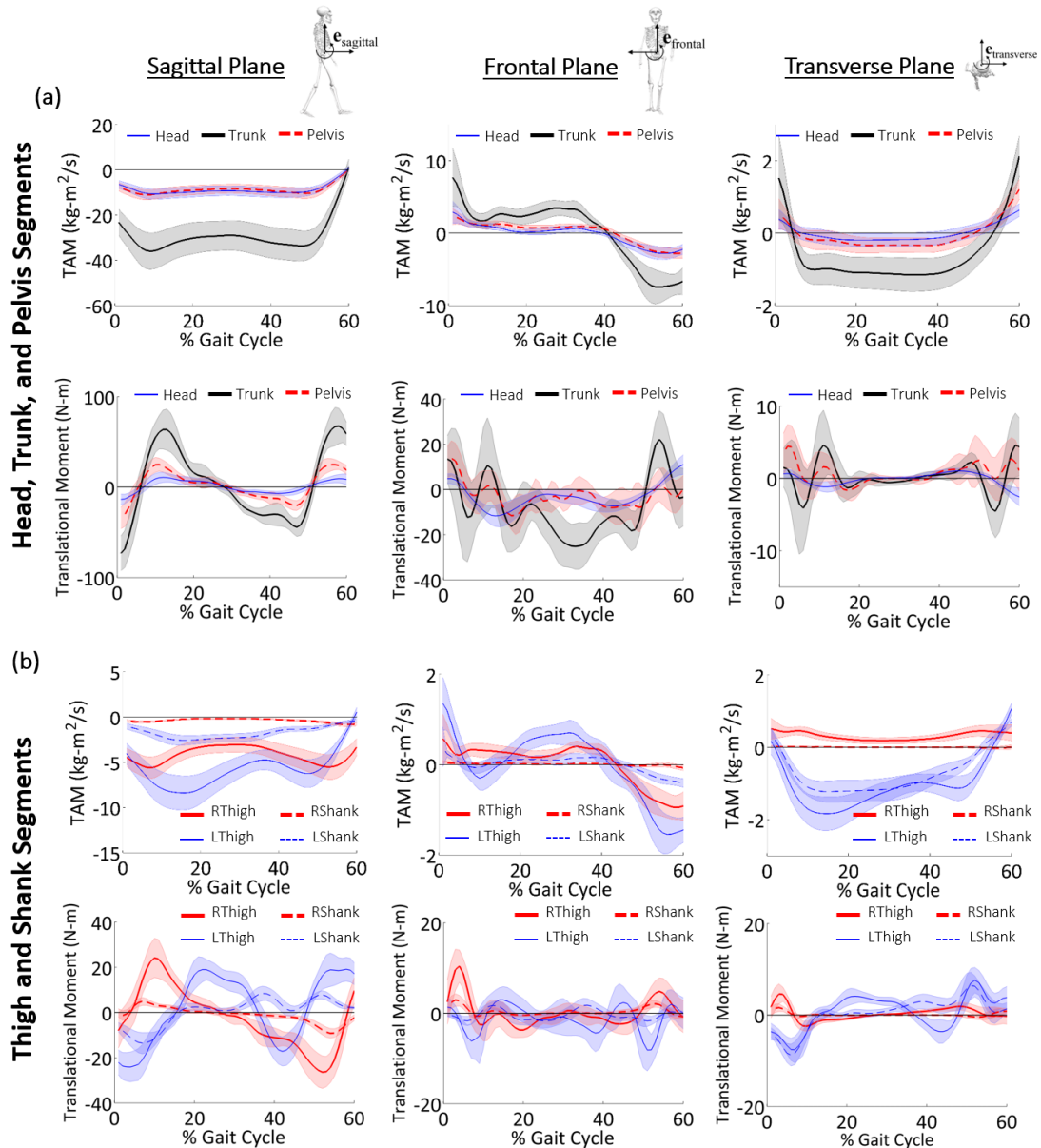


Figure 4.3. Mean (1 SD) translational angular momentum (TAM) and translational segmental moment about the stance foot of (a) the axial segments (head, trunk, pelvis) and (b) the lower extremities (bilateral thighs and shanks) in the sagittal, frontal, and transverse planes. TAM and the translational segmental moment about the stance foot were only calculated during the stance period (0-60% of the gait cycle) because that is the phase where the support limb is stationary (MacKinnon & Winter, 1993). Note the different scales between planes.

Table 4.1. Mean \pm SD peak (minimum and maximum) segmental translational angular momentum (TAM) during the stance period (right heel strike to right toe off). Units of TAM are ($\text{kg}\cdot\text{m}^2/\text{s}$) (Eq. (4.4)).

Segment	Sagittal Plane		Frontal Plane		Transverse Plane	
	<i>Minimum</i>	<i>Maximum</i>	<i>Minimum</i>	<i>Maximum</i>	<i>Minimum</i>	<i>Maximum</i>
Trunk	-36.94 \pm 8.08	0.40 \pm 3.18	-8.30 \pm 2.10	7.65 \pm 3.92	-1.14 \pm 0.46	2.48 \pm 1.01
Pelvis	-11.35 \pm 2.34	-0.36 \pm 0.96	-2.97 \pm 0.82	2.29 \pm 1.04	-0.42 \pm 0.24	1.22 \pm 0.40
Head	-10.99 \pm 2.39	0.44 \pm 1.02	-3.04 \pm 0.98	2.92 \pm 1.41	-0.17 \pm 0.17	1.22 \pm 0.40
Right Thigh	-5.92 \pm 1.30	-2.89 \pm 0.83	-1.07 \pm 0.26	0.68 \pm 0.37	0.17 \pm 0.11	0.63 \pm 0.21
Left Thigh	-8.33 \pm 1.88	0.42 \pm 0.54	-1.62 \pm 0.40	1.32 \pm 0.53	-1.89 \pm 0.46	1.10 \pm 0.22
Right Shank	-0.84 \pm 0.23	-0.14 \pm 0.06	-0.08 \pm 0.04	0.08 \pm 0.03	-0.03 \pm 0.04	0.06 \pm 0.05
Left Shank	-2.66 \pm 0.59	-0.34 \pm 0.17	-0.40 \pm 0.09	0.28 \pm 0.10	-1.24 \pm 0.33	0.71 \pm 0.14
Right Foot	0.00	0.00	0.00	0.00	0.00	0.00
Left Foot	-0.53 \pm 0.15	0.10 \pm 0.07	-0.13 \pm 0.05	0.08 \pm 0.03	-0.62 \pm 0.17	0.26 \pm 0.05
Right Upper Arm	-3.42 \pm 0.83	0.48 \pm 0.28	-0.62 \pm 0.18	0.53 \pm 0.31	0.08 \pm 0.09	0.60 \pm 0.16
Left Upper Arm	-2.76 \pm 0.59	-0.39 \pm 0.31	-0.97 \pm 0.25	0.86 \pm 0.33	-0.57 \pm 0.12	0.10 \pm 0.06
Right Forearm	-1.84 \pm 0.48	0.39 \pm 0.18	-0.18 \pm 0.06	0.19 \pm 0.12	-0.03 \pm 0.07	0.46 \pm 0.10
Left Forearm	-1.22 \pm 0.27	-0.32 \pm 0.11	-0.50 \pm 0.14	0.41 \pm 0.16	-0.38 \pm 0.08	-0.01 \pm 0.05
Right Hand	-0.63 \pm 0.18	0.13 \pm 0.07	-0.06 \pm 0.02	0.06 \pm 0.04	-0.05 \pm 0.04	0.20 \pm 0.05
Left Hand	-0.36 \pm 0.08	-0.06 \pm 0.08	-0.17 \pm 0.05	0.12 \pm 0.04	-0.14 \pm 0.04	0.00 \pm 0.02

Table 4. 2. Mean \pm SD peak (minimum and maximum) segmental translational moment during the stance period (right heel strike to right toe off). Units are (N·m) (Eq. (4.4)).

Segment	Sagittal Plane		Frontal Plane		Transverse Plane	
	Minimum	Maximum	Minimum	Maximum	Minimum	Maximum
Trunk	-74.61 \pm 18.48	79.55 \pm 20.63	-36.62 \pm 13.21	30.43 \pm 16.12	-7.31 \pm 5.11	7.81 \pm 4.66
Pelvis	-35.27 \pm 11.22	29.49 \pm 7.38	-17.09 \pm 6.92	16.49 \pm 7.28	03.74 \pm 1.79	6.55 \pm 2.61
Head	-13.28 \pm 5.80	12.86 \pm 5.30	-12.47 \pm 4.43	10.98 \pm 4.32	-2.67 \pm 1.21	1.20 \pm 0.41
Right Thigh	-27.92 \pm 7.54	25.54 \pm 7.48	-6.50 \pm 1.72	11.39 \pm 3.46	-3.22 \pm 1.05	5.11 \pm 1.80
Left Thigh	-26.15 \pm 6.00	24.43 \pm 7.09	-11.92 \pm 4.45	6.70 \pm 2.60	-7.86 \pm 2.76	9.07 \pm 3.50
Right Shank	-9.36 \pm 2.11	4.96 \pm 1.66	-1.21 \pm 0.47	3.23 \pm 1.19	-0.84 \pm 0.46	1.76 \pm 0.73
Left Shank	-14.11 \pm 4.38	12.04 \pm 9.03	-8.65 \pm 20.86	2.80 \pm 1.25	-9.07 \pm 2.50	7.39 \pm 2.27
Right Foot	0.00	0.00	0.00	0.00	0.00	0.00
Left Foot	-6.99 \pm 2.08	4.37 \pm 1.51	-1.70 \pm 0.63	1.88 \pm 1.44	-2.81 \pm 0.73	3.96 \pm 1.78
Right Upper Arm	-6.85 \pm 3.03	5.94 \pm 1.75	-3.17 \pm 0.99	2.80 \pm 0.97	-1.55 \pm 0.47	1.13 \pm 0.65
Left Upper Arm	-5.18 \pm 1.18	7.26 \pm 1.91	-4.04 \pm 1.44	3.91 \pm 1.78	-1.32 \pm 0.47	1.51 \pm 0.68
Right Forearm	-4.19 \pm 1.75	4.12 \pm 1.41	-1.61 \pm 0.49	1.05 \pm 0.37	-0.94 \pm 0.31	0.95 \pm 0.42
Left Forearm	-2.98 \pm 1.04	2.65 \pm 0.93	-1.65 \pm 0.61	1.66 \pm 0.77	-1.13 \pm 0.36	1.11 \pm 0.34
Right Hand	-1.57 \pm 0.66	2.00 \pm 0.60	-0.66 \pm 0.20	0.49 \pm 0.36	-0.82 \pm 0.31	0.48 \pm 0.18
Left Hand	-1.34 \pm 0.39	1.73 \pm 0.68	-0.80 \pm 0.35	0.72 \pm 0.33	-0.74 \pm 0.23	0.72 \pm 0.33

4.5.2 Rotational Angular Momentum

Unlike segmental TAM or translational moments, rotational angular momentum (RAM) and the rotational moments (\mathbf{M}_{seg}) of each segment had a unique shape. In addition, the magnitude of segmental RAM is one to two orders of magnitude smaller than segmental TAM. The TAM magnitude is largely influenced by the choice of reference point, and the moment arms can be large relative to the stance foot. However, because segmental RAM and TAM are both dynamically and geometrically independent from one another, and total segmental angular momentum (TAM + RAM) is not included in the present investigation, the difference in magnitude does not affect the current interpretation. The rotational moments represent the net moment about the segment COM due to the external moments (i.e., joint moments) and external forces at the proximal and distal ends of each segment (i.e., joint intersegmental forces). Because the rotational moment is driven by biomechanical loads, they are related to joint moments that are referred to as demand or effort moments (Carollo & Matthews, 2009). The relationship expressed in Eq. (4.9) enables a straightforward interpretation of RAM changes by kinetic principles of external demands.

In the sagittal plane, thigh and shank RAM were larger than any other segment (Figure 4.4). The symmetry across segments and limbs and timing of generation and arresting of segmental RAM curves correspond to the coordinated motion during swing limb advancement used to achieve forward progression. When not supporting the weight

of superior segments, timing of the peak thigh and shank rotational moments late in the swing period correspond with the external knee flexor moment required to arrest lower extremity momentum in preparation for foot placement.

In the frontal plane, the RAM of the trunk and thighs are larger than other segments, and occur primarily during loading and unloading periods of the stance limb (Figure 4.4, Table 4.3). During weight acceptance (2-12% of the gait cycle), trunk RAM is generated toward the stance foot. During loading, Trunk RAM is rapidly arrested by a distinct rotational moment away from the stance limb (negative), and then RAM is generated away from the stance foot throughout the duration of single limb support. Frontal plane trunk rotational moments are consistent with the low back lateral bend moments presented by Hendershot et al., (2014), likely to position the body COM away from the stance foot in preparation for contralateral heel strike. This pattern is inversely repeated when loading the contralateral limb. Peak frontal plane thigh RAM and the associated rotational moment away from the stance foot corresponds with the external hip abduction moment that occurs early in the swing period required for foot clearance.

In the transverse plane, the RAM of the trunk and pelvis are the larger than the other segments (Figure 4.4, Table 4.3). During loading response (0-2% of the gait cycle), RAM of the trunk is absent, and the rotational trunk moment is small, which indicates that no torques are required when loading the limb (Figure 4.4) and reflects the primarily sagittal plane motion. Transverse RAM of the trunk is generated for the duration of weight acceptance (2-12% of the gait cycle), and indicates axial rotation away from the stance foot (Figure 4.2). Transverse trunk RAM is then arrested for the duration of single limb support (12-50% of the gait cycle), which creates rotation toward the stance foot, until

and then absent during weight acceptance of the contralateral limb. Trunk rotational moments away from the stance limb during loading and toward the stance foot during single limb stance are correspond with the low back axial twist moments by Hendershot et al. (2014). Transverse RAM of the pelvis follows similar patterns in comparison to the trunk, but with smaller peak magnitudes, which is explained by the smaller inertia of the pelvis (Figure 4.4).

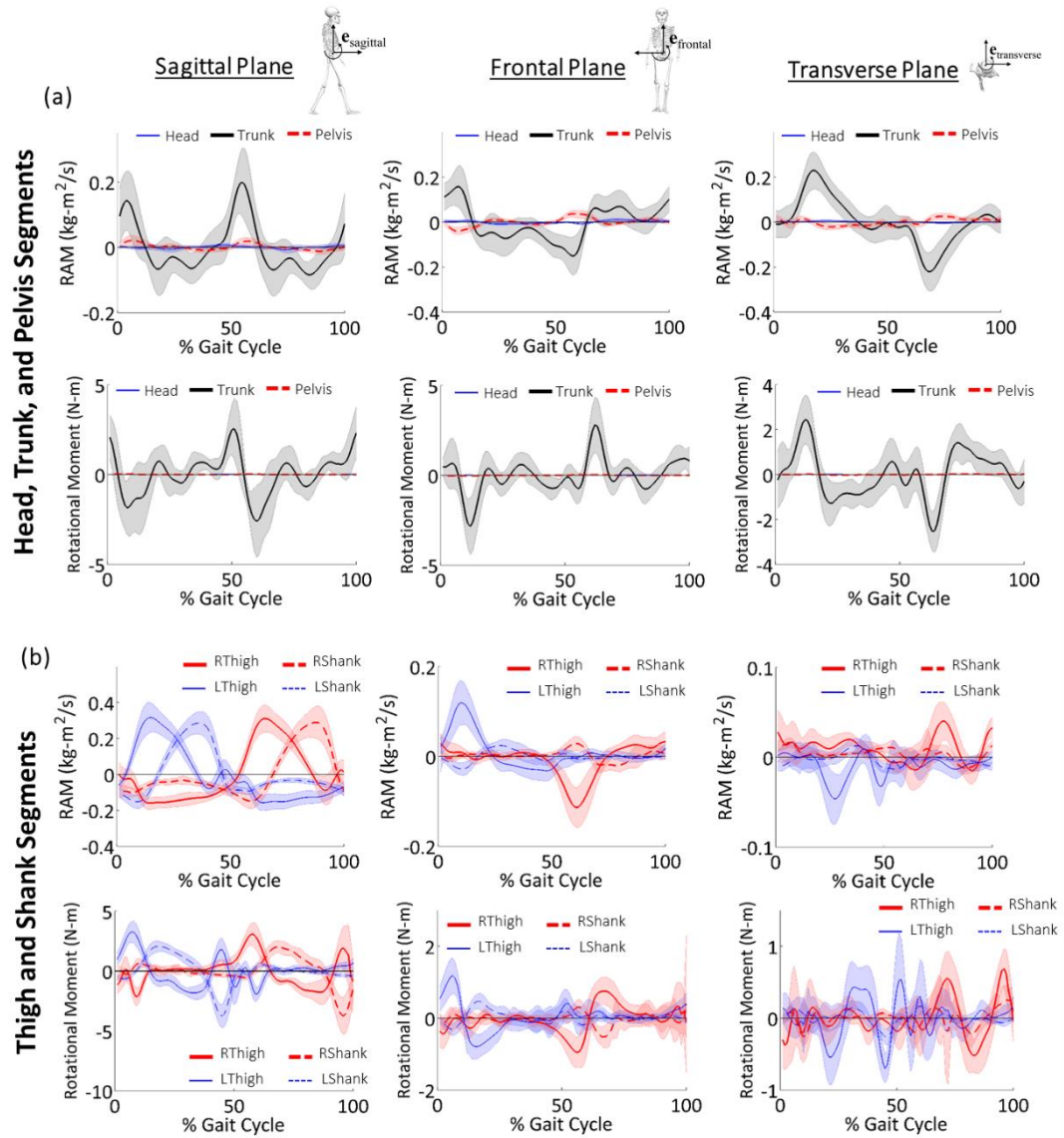


Figure 4.4. Mean (1 SD) rotational angular momentum (RAM) and translational segmental moment about the COM of the segment of (a) the axial segments (head, trunk, pelvis) and (b) the lower extremities (bilateral thighs and shanks) in the sagittal, frontal, and transverse planes. Note the different scales between planes.

Table 4.3. Mean \pm SD peak (minimum and maximum) segmental rotational angular momentum (RAM) during the gait cycle (right heel strike to right heel strike). Units of RAM are ($\text{kg}\cdot\text{m}^2/\text{s}$) (Eq. (4.5)).

Segment	Sagittal Plane		Frontal Plane		Transverse Plane	
	<i>Minimum</i>	<i>Maximum</i>	<i>Minimum</i>	<i>Maximum</i>	<i>Minimum</i>	<i>Maximum</i>
Trunk	-0.14 \pm 0.04	0.23 \pm 0.08	-0.19 \pm 0.08	0.20 \pm 0.10	-0.23 \pm 0.09	0.25 \pm 0.09
Pelvis	-0.02 \pm 0.01	0.03 \pm 0.01	-0.04 \pm 0.01	0.04 \pm 0.01	-0.04 \pm 0.01	0.04 \pm 0.01
Head	-0.01 \pm 0.01	0.01 \pm 0.01	-0.01 \pm 0.01	0.01 \pm 0.01	-0.005 \pm 0.00	0.01 \pm 0.00
Right Thigh	-0.19 \pm 0.04	0.34 \pm 0.07	-0.12 \pm 0.05	0.05 \pm 0.02	-0.03 \pm 0.01	0.06 \pm 0.02
Left Thigh	-0.18 \pm 0.05	0.33 \pm 0.08	-0.05 \pm 0.02	0.12 \pm 0.05	-0.06 \pm 0.02	0.03 \pm 0.01
Right Shank	-0.15 \pm 0.04	0.30 \pm 0.07	-0.03 \pm 0.02	0.12 \pm 0.05	-0.02 \pm 0.01	0.03 \pm 0.01
Left Shank	-0.16 \pm 0.03	0.29 \pm 0.05	-0.03 \pm 0.01	0.04 \pm 0.03	-0.04 \pm 0.07	0.02 \pm 0.01
Right Foot	-0.02 \pm 0.01	0.02 \pm 0.00	-0.01 \pm 0.00	0.01 \pm 0.00	0.00 \pm 0.00	0.00 \pm 0.00
Left Foot	-0.02 \pm 0.01	0.01 \pm 0.00	-0.01 \pm 0.00	0.01 \pm 0.00	0.00 \pm 0.00	0.00 \pm 0.00
Right Upper Arm	-0.02 \pm 0.01	0.02 \pm 0.01	-0.01 \pm 0.00	0.01 \pm 0.00	-0.01 \pm 0.00	0.01 \pm 0.00
Left Upper Arm	-0.02 \pm 0.01	0.02 \pm 0.01	-0.01 \pm 0.00	0.01 \pm 0.00	-0.01 \pm 0.0	0.01 \pm 0.01
Right Forearm	-0.01 \pm 0.01	0.01 \pm 0.00	-0.004 \pm 0.00	0.00 \pm 0.00	0.00 \pm 0.00	0.00 \pm 0.00
Left Forearm	-0.01 \pm 0.01	0.01 \pm 0.00	-0.003 \pm 0.00	0.00 \pm 0.00	0.00 \pm 0.00	0.00 \pm 0.00
Right Hand	0.00 \pm 0.00	0.00 \pm 0.00	0.00 \pm 0.00	0.00 \pm 0.00	0.00 \pm 0.00	0.00 \pm 0.00
Left Hand	0.00 \pm 0.00	0.00 \pm 0.00	0.00 \pm 0.00	0.00 \pm 0.00	0.00 \pm 0.00	0.00 \pm 0.00

Table 4.4. Mean \pm SD peak (minimum and maximum) segmental rotational moment about the stance foot during the gait cycle (right heel strike to right heel strike). Units are (N·m) (Eq. (4.8)).

Segment	Sagittal Plane		Frontal Plane		Transverse Plane	
	<i>Minimum</i>	<i>Maximum</i>	<i>Minimum</i>	<i>Maximum</i>	<i>Minimum</i>	<i>Maximum</i>
Trunk	-3.79 \pm 1.47	3.43 \pm 1.26	-3.62 \pm 2.38	3.46 \pm 1.54	-2.94 \pm 0.90	3.18 \pm 1.29
Pelvis	-0.41 \pm 0.19	0.48 \pm 0.20	-0.74 \pm 0.26	0.66 \pm 0.22	-0.83 \pm 0.35	0.80 \pm 0.29
Head	-0.35 \pm 0.23	0.28 \pm 0.11	-0.22 \pm 0.08	0.22 \pm 0.08	-0.06 \pm 0.03	0.07 \pm 0.06
Right Thigh	-3.25 \pm 1.39	3.89 \pm 1.20	-1.27 \pm 0.49	0.16 \pm 0.40	-0.88 \pm 0.29	1.02 \pm 0.30
Left Thigh	-2.70 \pm 0.93	3.35 \pm 0.80	-0.91 \pm 0.28	1.26 \pm 0.46	-0.94 \pm 0.26	0.98 \pm 0.35
Right Shank	-4.24 \pm 1.40	2.24 \pm 0.51	-0.71 \pm 0.42	0.93 \pm 1.66	-0.55 \pm 0.60	0.46 \pm 0.33
Left Shank	-3.91 \pm 0.89	2.65 \pm 1.59	-0.76 \pm 0.98	0.57 \pm 0.21	-0.33 \pm 0.11	1.19 \pm 3.24
Right Foot	-0.28 \pm 0.22	0.41 \pm 0.31	-0.29 \pm 0.65	0.16 \pm 0.37	-0.18 \pm 0.42	0.16 \pm 0.31
Left Foot	-0.22 \pm 0.06	0.35 \pm 0.09	-0.06 \pm 0.02	0.12 \pm 0.04	-0.08 \pm 0.04	0.07 \pm 0.04
Right Upper Arm	-0.15 \pm 0.06	0.14 \pm 0.07	-0.09 \pm 0.05	0.07 \pm 0.03	-0.18 \pm 0.13	0.14 \pm 0.10
Left Upper Arm	-0.13 \pm 0.05	0.14 \pm 0.06	-0.10 \pm 0.08	0.11 \pm 0.07	-0.22 \pm 0.25	0.25 \pm 0.23
Right Forearm	-0.12 \pm 0.05	0.10 \pm 0.04	-0.04 \pm 0.02	0.04 \pm 0.02	-0.03 \pm 0.01	0.03 \pm 0.01
Left Forearm	-0.11 \pm 0.04	0.09 \pm 0.03	-0.03 \pm 0.01	0.03 \pm 0.01	-0.03 \pm 0.01	0.04 \pm 0.01
Right Hand	-0.01 \pm 0.00	0.01 \pm 0.00	0.00 \pm 0.00	0.00 \pm 0.00	-0.01 \pm 0.00	0.01 \pm 0.00
Left Hand	-0.01 \pm 0.00	0.01 \pm 0.00	0.00 \pm 0.00	0.00 \pm 0.00	-0.01 \pm 0.00	0.01 \pm 0.01

4.6 Discussion

This investigation analyzed the individual contributions of total segmental angular momentum during walking, and the associated kinetics used to generate and arrest segmental angular momentum. The translational angular momentum (TAM) taken about the stance foot provides a coherent interpretation of forward progression during stance. The rotational angular momentum (RAM) about the segment COM can be used to identify specific movement patterns that are used to achieve forward progression through the variations in segmental angular velocity during the gait cycle. In addition, RAM is related to the net external moment through the rotational moment, and therefore represents an external biomechanical load that is representative of effort (Carollo & Matthews, 2009).

A key feature of segmental momentum and segmental moments is that they can be calculated using only kinematics, and may be more suitable for clinical implementation than inverse dynamics calculations. TAM requires segment localization, and is currently the more difficult of the two components to calculate outside of a motion capture laboratory. However, RAM may be achieved through small gyroscopes made possible by microelectromechanical systems (MEMS). MEMS sensors have been used to measure kinematics and spatiotemporal parameters during walking (Sinclair et al., 2013; Patterson et al., 2014; López-Nava et al., 2015) and used for movement retraining (Wall et al., 2009). As wearable sensors become more widely used in a clinical setting (typically for activity monitoring) (Butte et al., 2012; Redfield et al., 2013; Fulk et al., 2014), we anticipate more explorations will focus on how the implementation of wearable sensors

can be used to measure or infer biomechanically useful information outside of a traditional motion capture laboratory.

Although joint and muscle-based analyses are the most commonly used in human movement biomechanics, segmental-based analyses provide unique opportunities to examine how dynamic variables are transferred through the system to achieve the desired outcome (in this case, forward progression). Analysis of segmental angular momentum and segmental moments in combination with musculoskeletal simulation equations of motion will enhance our understanding of how momentum is generated, transferred, and arrested between segments. We also anticipate that the two components of total segmental angular momentum can be incorporated into system-based analyses related to flow of power across segments. Changes in segmental momentum intuitively correspond with altered segmental power, and linked to segmental power flow through systematic analyses such as bond graphs (Karnopp et al., 2012). Segment power flow has been presented in several investigations of human movement (Gordon et al., 1980; Neptune et al., 2001; McGibbon et al., 2002; Zajac et al., 2002), but is not ubiquitous within the biomechanics community, particularly in clinical applications where these analyses may be used to inform diagnoses and treatment.

Developing a clearer understanding of how segmental angular momentum and segmental moments are coordinated could assist in reinforcing or correcting movement patterns through muscle strengthening and retraining in the clinic. For example, lateral trunk lean toward the stance limb during gait is a common compensation due to weakness of the hip abductor muscles (Krautwurst et al., 2013). Quantification of peak trunk

segment momentum during stance can be used to document this movement compensation and potentially support the effectiveness of interventions, such as hip abductor strengthening, which are intended to change the altered movement pattern. The use of segmental moments in a clinical setting could be beneficial in identifying a potential consequential movement pattern, which is often diagnosed through increased joint demand that lead to overuse injuries (e.g., osteoarthritis). The current results demonstrate a large difference in magnitude of segmental moments between double and single limb support phases, which is consistent with differences of joint moments (identified via inverse dynamics) across these phases. This difference is due to increased power generation by the hip and ankle joints that is required to translate the body COM forward (Cappozzo et al., 1976; Wells, 1981; Winter, 1984; Winter et al., 1990; Hof, 2000) is consistent with previous angular momentum results showing larger changes in segmental angular momentum during double support (Robert et al., 2009). Future experimental work using these variables with clinical populations is needed to determine what deviations in segmental angular momentum and moments exist, and how sensitive and specific these variables for identifying movement deficits, and which are associated with consequential effects on the musculoskeletal system.

TAM provides a helpful framework to interpret intersegmental dynamics needed to maintain forward momentum of the body. TAM captures momentum generation during weight acceptance, little to no momentum generation through midstance (likely due to no hip or ankle power), and momentum arresting during pre-swing. The translational moments reflect the effects of muscle forces and that may or may not be attached to the

segment. For instance, the soleus and gastrocnemius are primary drivers of trunk acceleration and deceleration, respectively (Neptune et al., 2001; Zajac et al., 2003; Zmitrewicz et al., 2007), and are represented in the trunk translational moment. As a result, deviations in trunk translational moment may be indicator of problematic or ineffective plantar flexor function. With additional development and exploration through experiments and simulation, TAM and translational moments, which are calculated using only kinematics and inertial properties, may enhance in clinical inference and treatment.

Segmental RAM, which are segment angular velocities scaled by segment inertia, can enable analyses of kinematic strategy and effort (or demand). Because RAM is directly related to angular velocity, timing of segmental motion is easily observed and, for walking can be used to assess coordination of spatiotemporal events (Gaffney et al., 2016; Sigward et al., 2016). Scaling the angular velocity by segment inertia (calculating RAM) allows interpretation of the magnitudes in the context of effort needed to cause rotation. Intuitively, it is more difficult to generate and arrest momentum of segments with large inertia versus segments with small inertia. The rotational moment, which captures the kinetic effort needed to generate or arrest RAM includes both the joint moments and the moments due to proximal and distal forces (Eq. (4.11)), which are driven by segment motion and forces that propagate through the kinetic chain. However, it is important to note that the rotational moment does not provide the detailed breakdown of moments and forces calculated from inverse dynamic analyses. Therefore, this should not be used as a surrogate to inverse dynamic calculations, but rather used to provide insight regarding joint kinetics when instrumentation is not available (e.g. within a

clinical setting). With further exploration through experiments and simulation, the interactions and relationships between segmental RAM and rotational moments, and traditional biomechanical variables such joint moments and joint powers can be understood and applied. In support of this idea, a recent investigation by (Sigward et al., 2016) demonstrated a strong association of shank angular velocity with knee extensor moment during weight acceptance in patients following ACL reconstruction. This result suggests that analysis of the underlying dynamics through RAM and rotational moment of the shank during heel rocker may facilitate additional insight into the interactions between shank angular velocity, knee extensor moment, and the simultaneously occurring ankle dorsiflexor moment.

There are several limitations to this investigation that should be considered. First, this analysis should be limited to over-ground walking. During movements containing ballistic motion (e.g. flight phases of sports activities), angular momentum with respect to the body COM is a more appropriate analysis; and using the foot as a reference point is no longer valid. Second, all participants walked at the same speed. Segment patterns identified using this analysis will vary with gait speed. Future work should investigate how segmental movement patterns are altered to accommodate for a change in gait speed. Third, segmental RAM and TAM were calculated using kinematics measured from reflective markers placed on the skin, which are subject to error primarily through skin motion artifact and marker placement error. Finally, qualitative associations between rotational moments and joint moments were based off of well-known waveforms within the literature of joint moments calculated through inverse dynamics; and therefore,

specific quantitative associations between rotational moments and joint kinetics remain unknown. Future work should establish these associations to determine how the net external moment on a body segment, as determined through the time rate of change of RAM, is associated with joint demand.

4.7 Conclusion

This investigation assessed the individual contributions of each component of total segmental angular momentum (TAM and RAM) and the kinetics used to generate and arrest segmental angular momentum during walking. The timing and waveforms of the generation and arresting of TAM and RAM describe the coordinated segmental movement patterns used to achieve forward progression during walking. Through Euler's rotational laws, the time derivative of TAM and RAM can be used to describe the underlying external forces and moments applied to each segment that cause motion. Because these forces and moments are representative of an external biomechanical load, the generation and arresting of segmental angular momentum is likely an indicator of the demand placed on the musculoskeletal system.

**CHAPTER 5: IDENTIFICATION OF TRUNK AND PELVIS MOVEMENT
COMPENSATIONS IN PATIENTS WITH TRANSTIBIAL AMPUTATION
USING ANGULAR MOMENTUM SEPARATION**

5.1 Abstract

Patients with unilateral dysvascular transtibial amputation (TTA) have a higher risk of developing low back pain than their healthy counterparts, which may be related to movement compensations used in the absence of ankle function. Assessing components of segmental angular momentum provides a unique framework to identify and interpret these movement compensations alongside traditional observational analyses. Angular momentum separation indicates two components of total angular momentum: 1) transfer momentum and 2) rotational momentum. The objective of this investigation was to assess movement compensations in patients with dysvascular TTA, patients with diabetes mellitus (DM), and healthy controls (HC) by examining patterns of generating and arresting trunk and pelvis segmental angular momenta during gait. We hypothesized that all groups would demonstrate similar patterns of generating/arresting total momentum and transfer momentum in the trunk and pelvis in reference to the groups (patients with DM and HC). We also hypothesized that patients with amputation would demonstrate different (larger) patterns of generating/arresting rotational angular momentum in the trunk. Patients with amputation demonstrated differences in trunk and pelvis transfer angular momentum in the sagittal and transverse planes in comparison to the reference

groups, which indicates postural compensations adopted during walking. However, patients with amputation demonstrated larger patterns of generating and arresting of trunk and pelvis rotational angular momentum in comparison to the reference groups. These segmental rotational angular momentum patterns correspond with high eccentric muscle demands needed to arrest the angular momentum, and may lead to consequential long-term effects such as low back pain.

5.2 Introduction

Over one million Americans currently have a lower-limb amputation, and this number is projected to double by 2050 (Ziegler-Graham et al., 2008) due to dysvascular pathologies (e.g. diabetes mellitus (DM)) (Dillingham et al., 2002). Patients with dysvascular amputation commonly have multiple comorbidities and 40-50% have limited physical function (Davies & Datta, 2003), which require different treatments apart from patients with traumatic amputation. Although patients with dysvascular amputation differ in age, BMI, prosthetic use time, and comorbidities from patients with traumatic amputation (Davies & Datta, 2003; van Velzen et al., 2006), it is common to combine them into a single group when investigating how amputation affects functional movement characteristics (Fey et al., 2010; Silverman & Neptune, 2011). Because patients with DM prior to amputation move differently than healthy controls (Mueller et al., 1994), differences in movement compensations between patients with dysvascular amputation to patients with DM alone could be used as physical rehabilitation targets for movement retraining following amputation.

Patients with unilateral transtibial amputation (TTA) are at increased risk of developing low back pain (LBP) (Ehde et al., 2001), which may relate to necessary movement compensations to achieve forward progression and balance during walking. For example, to accomplish forward progression in the absence of an ankle plantar flexor, patients with unilateral TTA increase hip extensor power during the stance period of the residual limb (Winter & Sienko, 1988). Patients with unilateral TTA demonstrate exaggerated lateral trunk lean toward the amputated limb (compensated Trendelenburg)

(Molina-Rueda et al., 2014) and altered foot placement of the intact limb, which leads to uneven step length, swing time, and stance time (Winter & Sienko, 1988). While these compensations may be necessary to accomplish mobility, asymmetric movements are linked to the development of LBP (Kumar, 2001). This coordination of excessive trunk and pelvic motion during walking likely contributes to step-to-step asymmetric loading at the low back previously measured in patients with unilateral TTA (Hendershot & Wolf, 2014), and may increase the risk of developing LBP, which was previously demonstrated in patients with transfemoral amputation (Hendershot & Wolf, 2015; Russell Esposito & Wilken, 2014).

Clinicians rely on observational gait analysis to identify movement compensations which is highly subjective and unreliable for identifying consequential movement compensations in amputees (Saleh & Murdoch, 1985). Although laboratory-based gait analysis is valid and reliable for quantitatively measuring movement, it is accompanied by high computational and economic expenses, and currently impractical in the vast majority of clinical settings. Because clinicians use observational gait analysis to guide interventions and gait retraining in patients with unilateral TTA, the ability to obtain accurate measures of trunk and pelvis movement patterns could help tailor treatment to patients and ultimately prevent injuries, such as LBP.

Identification of segmental strategies used to generate and arrest segmental angular momentum can provide insight into muscle demands following unilateral dysvascular TTA. During walking, muscles are used concentrically and eccentrically as the primary mechanisms to generate and arrest segment angular momentum (Neptune & McGowan,

2011). Measuring and understanding segmental angular momentum is a promising approach to bridge the gap between observational and quantitative gait analysis. We previously demonstrate a framework to describe clinical movement compensations during gait using separation of translational angular momentum referenced to the stance foot (Gaffney et al., 2017). Total segmental angular momentum can be separated into two components, each with a unique interpretation: 1) Translational Angular Momentum (TAM): angular momentum created by linear velocity of the segment with mass with respect to a point and 2) Rotational Angular Momentum (RAM): angular momentum created by the rotational velocity of an object with inertia (Kasdin & Paley, 2011).

The objective of this investigation was to assess movement compensations in patients with unilateral dysvascular TTA and patients with DM by examining translational angular momentum and rotational angular momentum of the trunk and pelvis during walking for patterns of generating/arresting momentum. We hypothesized that patients with unilateral dysvascular TTA, patients with DM, and healthy control participants would demonstrate similar patterns of generating/arresting TAM of the trunk and pelvis when walking at similar speeds. We also hypothesized that patients with unilateral dysvascular TTA would demonstrate higher RAM of the trunk and pelvis than the other groups, which illustrates potentially consequential movement compensations that can be retrained through clinical intervention.

5.3 Methods

5.3.1 Participants

Ten patients with DM and unilateral TTA 1-3 years post amputation (AMP) (Table 5.1) (10 M; age: 56.8 ± 4.3 years; mass: 97.6 ± 15.2 kg; height: 1.8 ± 0.1 m), 11 patients with DM (2F, 9 M; age: 61.4 ± 8.0 years; mass: 94.3 ± 22.0 kg; height: 1.7 ± 0.1 m), and 13 healthy control patients (HC) (3 F, 10 M; age: 63.1 ± 7.7 years; mass: 77.7 ± 13.2 kg; height: 1.7 ± 0.1 m) were enrolled. Further details regarding individual participant anthropometrics and levels of functional performance can be found in Appendices A and B, respectively.

Eligibility criteria included: age: 50-85 years; BMI ≤ 40 kg/m²; independent community ambulation (ability to walk for four minutes without rest or assistive device); 1-3 years post amputation (AMP group); controlled Type-II diabetes mellitus (AMP and DM groups); no traumatic or cancer-related amputation (AMP group); no major amputation on contralateral limb (AMP group); no cardiovascular, orthopaedic, neurologic, wounds, or ulcers that limit physical function; no history of LBP (HC group); no diagnosed rheumatoid arthritis (HC group); no diagnosed osteoarthritis (HC group); and no total hip/knee joint arthroplasty (HC group). Each participant provided a written, informed consent in accordance with the Colorado Multiple Institutional Review Board prior to the start of the experimental session and completed one data collection in which whole body kinematics were collected.

Table 5.1 Participant characteristics for patients with dysvascular unilateral transtibial amputation (AMP) group.

Time since Amputation (Months)	Residual Limb Length (cm)	Socket Type	Prosthetic Foot
17.4 ± 5.1	14.8 ± 2.5	Total contact carbon fiber	Dynamic elastic response

5.3.2 Motion Analysis

Each participant was instrumented with 63 reflective markers used to obtain whole-body kinematics during gait. Motion was recorded from eight infrared cameras (Vicon) sampled at 100 Hz. Each participant performed three gait trials at 1.0 m/s (± 0.05 m/s) on a 10-m walkway. Motions were averaged across the three trials and used for group comparisons.

5.3.3 Data Analysis

Kinematic data were low-pass filtered with a 4th-order Butterworth filter (6 Hz cutoff frequency). A 15-segment subject-specific model (head, upper arms, forearms, hands, trunk, pelvis, thighs, shanks, and feet) was created in Visual 3D (C-Motion, Inc.). Segment masses were based on a percentage of total body weight and segment inertias were based on segment geometry (Dempster & Aitkens, 1995). For the AMP group, mass the center of mass position, and inertial properties of the prosthetic shank (residual limb + prosthetic socket) and prosthetic foot were determined using a reaction board technique and oscillation method (Smith et al. 2014).

TAM (angular momentum of a segment with respect to the stance foot) is described as:

$${}^I \mathbf{h}_{i/\text{foot}} = (\mathbf{r}_i - \mathbf{r}_{\text{Foot}}) \times m_i ({}^I \mathbf{v}_i - {}^I \mathbf{v}_{\text{Foot}}) \quad (5.1)$$

where \mathbf{r}_i and \mathbf{r}_{Foot} are the position vectors of the i^{th} segment and foot, respectively, m_i is the mass of the i^{th} segment, and ${}^{\mathcal{I}}\mathbf{v}_i$ and ${}^{\mathcal{I}}\mathbf{v}_{\text{Foot}}$ are the inertial velocities of the i^{th} segment and foot respectively. RAM (angular momentum of a segment with respect to its center of mass) is described as:

$${}^{\mathcal{I}}\mathbf{h}_i = \mathbf{I}_i \cdot \boldsymbol{\omega}_i \quad (5.2)$$

where \mathbf{I}_i is the moment of inertia tensor and $\boldsymbol{\omega}_i$ is the angular velocity of the segment. To facilitate planar analyses, all angular momenta vectors were expressed in a path reference frame, that is defined by the velocity vector of the body COM: $\mathbf{e}_{\text{frontal}}$ (tangent to the horizontal path of the body COM), $\mathbf{e}_{\text{transverse}}$ (opposite direction of the gravity vector), and $\mathbf{e}_{\text{sagittal}}$ ($\mathbf{e}_{\text{frontal}} \times \mathbf{e}_{\text{transverse}}$). Within the path reference frame, positive momenta values in each plane are defined as: sagittal – posterior rotation away from stance foot, frontal – medial-lateral rotation toward stance foot, transverse – rotation away from stance foot.

5.3.4 Statistical Analysis

Patient anthropometrics (mass and height) were compared across groups using a one-way ANOVA followed by Tukey HSD for post hoc comparison ($\alpha = 0.05$).

All momenta were calculated during one gait cycle (AMP: amputated limb heel strike to amputated limb heel strike; DM and HC: right heel strike to right heel strike). TAM (${}^{\mathcal{I}}\mathbf{h}_{i/\text{Foot}}$) was calculated with respect to the stance foot was analyzed during the stance period. RAM (${}^{\mathcal{I}}\mathbf{h}_i$) was analyzed during the entire gait cycle. To quantify generation and arresting of trunk and pelvis angular momentum, global minimums and maximums were determined.

Magnitudes of the global minima and maxima in each segmental angular momentum variable (TAM and RAM) were compared across the three groups using an ANCOVA (covariates: mass and height) followed by pairwise comparisons using Tukey HSD ($\alpha = 0.05$). Qualitative analysis was performed to assess when peak momenta values occurred throughout the functional phases of gait: weight acceptance (0-12%), single limb support (12-50%), swing limb advancement (50-100%) (Perry & Burnfield, 2010).

5.4 Results

5.4.1 Patient Anthropometrics

Body mass was larger in the AMP group than the HC group ($P = 0.03$). No differences in height existed across groups.

5.4.2 Translational Angular Momentum

In the sagittal plane, peak posterior trunk and pelvis TAM was lower in the AMP group than the DM group (trunk: $P = 0.01$, pelvis: $P = 0.01$) at the end of single limb support (Figure 5.1, Table 5.1). In the sagittal plane, peak anterior trunk TAM was lower in the DM group than the HC group ($P = 0.03$) at the beginning of single limb support (Figure 5.1a, Table 5.2).

In the frontal plane, peak lateral trunk TAM toward the stance foot was lower in the AMP group than the DM group ($P < 0.001$) during weight acceptance (Figure 5.1a, Table 5.2).

In the transverse plane, peak trunk and pelvis TAM toward the stance foot was higher in the AMP group than the DM group (trunk: $P = 0.03$, pelvis: $P = 0.01$) at the beginning of single limb support (Figure 5.1, Table 5.2). Peak pelvis TAM away from the stance

foot was lower in the AMP group than both the DM group ($P < 0.001$) and the HC group ($P < 0.001$) at the end of single limb support (Figure 5.1b, Table 5.2). All other comparisons were not statistically significant.

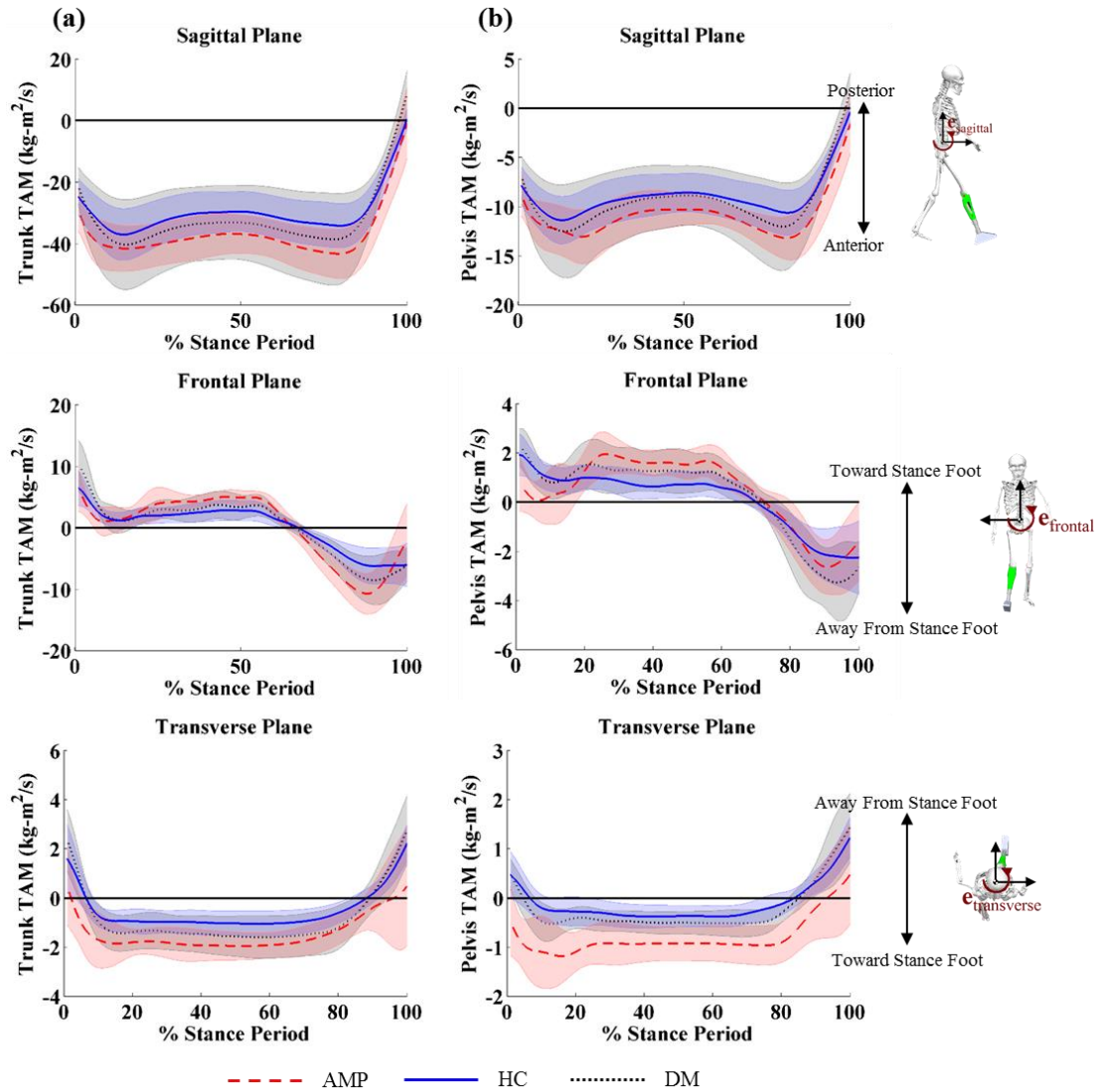


Figure 5.1. Translational angular momentum (TAM) of the (a) trunk and (b) pelvis with respect to the stance foot in the sagittal, frontal, and transverse plane healthy controls (blue solid line), patients with diabetes mellitus (DM) (black dotted line), and patients with DM and transtibial amputation (AMP) (red dashed line).

5.4.3 Rotational Angular Momentum

In the sagittal plane, peak anterior trunk RAM was higher in the AMP group than both the DM group ($P = 0.02$) and HC group ($P = 0.01$) at the beginning of single limb support (Figure 5.2a, Table 5.3). Peak posterior trunk RAM was lower in the AMP group than both the DM group ($P = 0.04$) and HC group ($P = 0.05$) at the beginning of swing limb advancement (Figure 5.2a, Table 5.2). Peak anterior pelvis RAM was higher in the AMP group than both the DM group ($P = 0.04$) and the HC group ($P = 0.04$) at the beginning of single limb support (Figure 5.2b, Table 5.3).

In the frontal plane, peak lateral trunk RAM toward the stance foot was higher in the AMP group than the DM group ($P = 0.04$) during swing limb advancement (Figure 5.2a, Table 5.3).

In the transverse plane, peak pelvis RAM toward the stance foot was higher for the AMP group than both the DM group ($P = 0.02$) and the HC group ($P = 0.03$) at the beginning of single limb support (Figure 5.2b, Table 5.3). All other comparisons were not statistically significant.

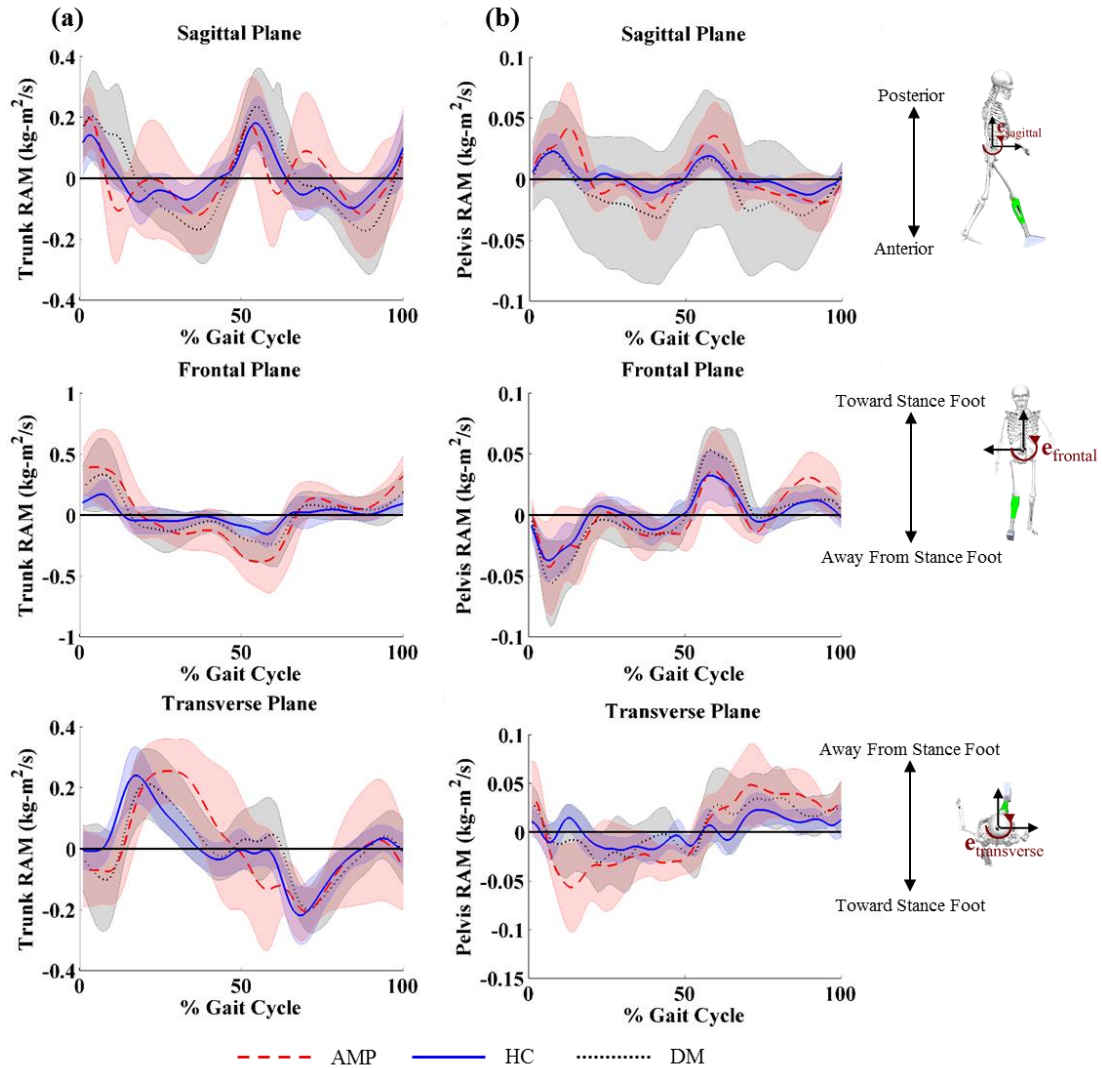


Figure 5.2. Rotational angular momentum (RAM) of the (a) trunk and (b) pelvis with respect to the stance foot in the sagittal, frontal, and transverse plane healthy controls (blue solid line), patients with diabetes mellitus (DM) (black dotted line), and patients with DM and transtibial amputation (AMP) (red dashed line).

5.5 Discussion

The objective of this investigation was to identify and compare movement patterns in patients with dysvascular transtibial amputation (AMP), patients with diabetes mellitus (DM), and healthy controls (HC) using patterns of generating and arresting trunk and pelvis angular momentum during walking. We observed differences in translational

angular momentum in all three planes between the AMP, DM, and HC groups, which indicates unique movement patterns adopted by each group during walking. Loss of ankle function in the AMP group is linked to different movement compensations, and results in higher generation of trunk and pelvis RAM in all three planes compared to the DM and HC groups. Large trunk angular momentum with small pelvis momentum is a compensation in the AMP group that may result in high paraspinal muscle demand, which leads leading to LBP. The identification of movement compensations through analysis of segmental RAM has potential important clinical applications in a gait retraining setting through wearable sensors.

Patterns of trunk and pelvis TAM indicate the use of a postural compensation by the AMP group to maintain balance and achieve forward progression without ankle function. TAM is a function of position and linear momentum of each segment relative to the stance foot (Eq. (5.1)). In the sagittal plane, trunk and pelvis anterior TAM is generated about the stance foot during weight acceptance, is slightly arrested throughout single limb support, and then arrested completely at the transition to swing limb advancement (Figure 5.1). Without active plantar flexion at the end of single limb support, the AMP group generated smaller posterior angular momentum when compared to the DM group, which is adopted to maintain forward progression when unloading the amputated limb. In the frontal plane, trunk and pelvis TAM toward the stance limb is rapidly arrested during loading response and then is gradually arrested throughout the remainder of single limb support until angular momentum is generated away from the stance limb during the preparation of swing limb advancement as weight is transferred between limbs. In the

transverse plane, trunk and pelvis TAM were arrested during loading response and then remained constant throughout the duration of single limb stance. Remarkably, trunk and pelvis TAM at initial foot contact in the AMP group were directed toward the stance (amputated) limb, which is opposite of both the HC and DM groups. This difference is likely a result of excessive propulsion by the intact limb, which creates a transverse rotation toward the amputated limb. Because each group walked at the same speed, the large transverse TAM toward the stance foot throughout the duration of single limb support in the AMP group occurs by a more medial position of the segment with respect to the stance foot. In the frontal plane, this corresponds to a wider step width, which is a commonly observed finding in amputee gait (Winter & Sienko, 1988).

Segment rotational angular momentum provides a unique framework for identifying differences in movement patterns by highlighting the motion of the segment, which can assist in characterizing and interpreting movement compensations observed in the clinic. In the sagittal plane, large anterior rotational angular momentum in the AMP group leads to a forward trunk lean that is frequently observed during single limb support, and represents an adaptive strategy to maintain forward progression in light of ankle plantar flexor loss (Miff et al., 2005). The hip and trunk extensor demands needed to arrest the large anterior trunk rotational angular momentum, which occurs at approximately 10% of the gait cycle, may contribute to overuse injuries in the lower extremity and the low back (Gailey, 2008; Silverman & Neptune, 2014).

In the frontal plane, the AMP group generated larger trunk RAM toward the amputated limb during weight acceptance, and arrested trunk angular momentum later in

the gait cycle, compared to the DM and HC groups, corresponding to large trunk displacement toward the stance limb. To prevent a fall at this point in the gait cycle, the AMP group must quickly arrest a large amount of angular momentum that has been generated in the trunk, which creates high paraspinal muscle demand (Friel et al., 2005). The lateral trunk posture over the stance limb (compensated Trendelenburg posture) corresponds with increased loading in the low back, and is linked to the development of LBP (Hendershot et al., 2013).

In the transverse plane, the AMP group generated substantially larger pelvis RAM toward the amputated limb than the HC and DM groups during weight acceptance. This large RAM, due to excessive angular speed, may be linked to the large ankle power in the intact limb needed to achieve forward progression (Nolan & Lees, 2000). Therefore, pelvis RAM must be arrested following peak generation to maintain balance during swing and continue progression during gait.

Our results indicate that the AMP group generates and arrests trunk and pelvis momenta differently than either the DM or HC groups, and the associated muscle demands with the observed movement patterns after amputation may be linked to LBP (Ehde et al., 2001; Hendershot & Wolf, 2014). Because a patient with unilateral TTA cannot create propulsive ankle joint moments, the generating demands shifts higher in the kinetic chain. Our results show that movement compensations occur in the pelvis and trunk, and indicates that demand on local muscles (e.g. multifidus, erector spinae, obliques, etc.) is likely higher for patients with amputation than other populations, which potentially could be consequential in the development for LBP.

Identifying movement compensations using segmental RAM may have important clinical applications. RAM combines inertia with angular speed of the segment, which is a parameter that can be interpreted in light of the effort needed to generate or arrest the measured momentum. Because observational analysis is based on presence of events and postures (e.g. compensated Trendelenburg sign), a clinician can gain insight into the effort needed to accomplish the observed event by supplementing with angular momentum. Measuring rotational angular momentum is easily facilitated by wearable sensors such as gyroscopes, and would not require additional instrumentation (e.g. force platforms). Use of low-cost wearable sensors have emerged in biomechanics that facilitate spatiotemporal gait characteristics (Rueterbories et al., 2010; Sabatini et al., 2005) as well as segment and joint kinematics (Watanabe et al., 2011). With additional research, angular momentum may provide clinicians and patients with immediate and accurate information on their ability understand when movement compensations occur and increase the efficacy of targeted movement retraining following amputation.

Several limitations should be considered. First, the analysis did not consider consecutive gait cycles; therefore, repeatability of movement compensations was not characterized in these measures. In future investigations we will extend this analysis using repeated over ground trials or a treadmill. Second, we do not know how segmental angular momentum variables correspond with traditional biomechanical variables. In future investigations we will associate how these movement compensations correlate with traditional quantitative biomechanical analyses (e.g. joint moments, joint loading, etc.). Third, neither the AMP and DM groups were screened for LBP at the time of testing;

therefore, we cannot determine if any compensatory movement patterns adopted by each group were a habitual movement pattern or a result of LBP. Finally, because only patients with dysvascular amputation were included, we are unable to generalize these findings to patients with unilateral TTA from other causes other than dysvascular disease.

5.6 Conclusion

This investigation demonstrated the use of segmental angular momenta to identify movement compensations in the trunk and pelvis in patients with unilateral dysvascular transtibial amputation. Coordinated compensations between the trunk and pelvis promote forward progression during locomotion, but may have long-term adverse effects from the demand placed on the musculoskeletal system to generate and arrest segmental momentum.

CHAPTER 6: TRUNK KINETIC EFFORT DURING STEP ASCENT AND DESCENT IN PATIENTS WITH TRANSTIBIAL AMPUTATION USING ANGULAR MOMENTUM SEPARATION

6.1 Abstract

Patients with transtibial amputation adopt trunk movement compensations that alter effort and increase the risk of developing low back pain. However, the effort required to achieve high-demand tasks, such as step ascent and descent, remains unknown.

Kinematics were collected during bilateral step ascent and descent tasks from two groups:

1) seven patients with unilateral transtibial amputation and 2) seven healthy control

subjects. Trunk kinetic effort was quantified using translational and rotational segmental moments (time rate of change of segmental angular momentum). Peak moments during

the loading period were compared across limbs and across groups. During step ascent,

patients with transtibial amputation generated larger sagittal trunk translational moments

when leading with the amputated limb compared to the intact limb ($P = 0.01$). The

amputation group also generated larger trunk rotational moments in the frontal and

transverse planes when leading with either limb compared to the healthy group ($P = 0.01$,

$P \ll 0.017$, respectively). During step descent, the amputee group generated larger trunk

translational and rotational moments in all three planes when leading with the intact limb

compared to the healthy group ($P \ll 0.017$). This investigation identifies how differing

trunk movement compensations, identified using the separation of angular momentum,

require high kinetic effort during stepping tasks in patients with transtibial amputation compared to healthy individuals. Compensations that produce identified increased and symmetric trunk segmental moments, may increase the risk of the development of low back pain in patients with amputation.

6.2 Introduction

Currently in the United States, over 80% of all lower-limb amputations result from neurovascular pathologies (e.g. diabetes mellitus), and this number is increasing due to an aging population and growing prevalence of patients with diabetes mellitus (Dillingham et al., 2002; Davies & Datta, 2003). Patients with dysvascular amputation have additional comorbidities that are associated with poor physical function such as aging, osteoarthritis, obesity and low back pain (LBP) (Ehde et al., 2001; Norvell et al., 2005; Ziegler-Graham et al., 2008; Morgenroth et al., 2011; Rosenberg et al., 2012). Because of these comorbidities, 40-50% of patients with dysvascular transtibial amputation (TTA) fail to achieve community ambulation one year following amputation (Davies & Datta, 2003), and the majority report difficulty achieving high-demand tasks such as step ambulation (De Laat et al., 2013).

To compensate for the loss of active ankle plantarflexion, patients with TTA adopt exaggerated trunk movements, which assist forward progression in the sagittal plane and help maintain balance/stability in the frontal plane during ambulation (Sagawa et al., 2011). Because the trunk accounts for almost two-thirds of the body mass (Winter, 1990), even seemingly small trunk movement compensations can lead to high loads in the low back and increased risk developing LBP (Kumar, 2001; McGill, 2007). In the sagittal plane, patients with amputation use an exaggerated forward trunk lean (Anzel et al., 1994; Goujon-Pillet et al., 2008), and generate large anterior momentum during the pre-swing phase of walking (Gaffney et al., 2016). The anterior position of the body COM relative to the knee requires less ankle propulsion during pre-swing to translate the body

COM forward with respect to the stance foot, and reduces quadriceps demand on the amputated limb (Torburn et al., 1990; Anzel et al., 1994; Powers et al., 1998). In the frontal plane, patients with TTA exaggerate the lateral trunk lean toward the amputated limb during weight acceptance (compensated Trendelenburg gait pattern) (Molina-Rueda et al., 2014; Gaffney et al., 2016). Because these patients have similar hip abductor strength in the amputated and intact limbs (Nadollek et al., 2002), the compensated Trendelenburg pattern is likely a strategy used to maintain balance on the amputated limb. Although the development of LBP remains idiopathic, these patterns are likely associated with detrimental low back loading (Devan et al., 2014; Hendershot & Wolf, 2014; Esposito & Wilken, 2014).

A clear understanding of trunk compensations adopted by patients with TTA during high-demand tasks may help identify potential risk factors that can be used to improve the efficacy of movement retraining following amputation. Step ascent and descent are common high-demand tasks required for community ambulation that are more difficult for patients with TTA than their healthy counter parts (De Laat et al., 2013; Jayakaran et al., 2013). Because step ambulation places higher demand on the musculoskeletal system (higher joint loading and muscle activation) compared to over-ground walking, compensations adopted during these tasks likely increase the risk of sustaining additional disabling comorbidities (McFadyen & Winter, 1988; Nadeau et al., 2003; Protopapadaki et al., 2007; Reeves et al., 2008). However, most biomechanics investigations of movement compensations in patients with TTA focus on over-ground walking.

The objective of this investigation was to evaluate trunk compensations and the associated kinetic effort needed for patients with unilateral TTA to perform step ascent and descent tasks, and compare to healthy control subjects. We accomplished this by identifying translational and rotational trunk moments, which were calculated from segmental momenta. We previously identified differences in the generation and arresting of trunk translational (TAM) and rotational angular momentum (RAM) in patients with unilateral dysvascular TTA during over-ground walking (Gaffney et al., 2016). In the current study, we hypothesized that patients with TTA would require larger trunk kinetic effort when stepping onto the amputated limb compared to the intact limb or healthy controls for both step ascent and descent. By establishing the link between trunk movement compensations and kinetic effort, we can gain additional insight into consequential effects of compensatory movement patterns, and can provide more targeted movement retraining following TTA.

6.3 Methods

6.3.1 Participants

Seven male patients with unilateral TTA and seven male healthy control (HC) participants were enrolled (Table 6.1). Further details regarding individual participant anthropometrics and levels of functional performance can be found in Appendices A and B, respectively. For the TTA group, all prostheses were passive with total contact carbon fiber sockets and dynamic elastic response feet. Each participant provided written, informed consent in accordance with the Colorado Multiple Institutional Review Board prior to the start of the experimental session. Each participant visited the laboratory for

one data collection session, in which whole body kinematics were collected during step ascent and descent tasks.

Table 6.1. Mean (1 SD) participant characteristics for patients with dysvascular unilateral transtibial amputation (TTA) and healthy control (HC) groups. Functional performance was quantified using the stair climb test (SCT) (Powers et al., 1997; Bean et al., 2007; Schmalz et al., 2007; Bennell et al., 2011).

Group	Age (Years)	BMI (kg/m²)	Time since Amputation (Months)	Residual Limb Length (cm)	SCT – Ascent Time (s)	SCT – Total Time (s)
TTA	56.3 (4.5)	28.3 (2.7)	16.7 (5.2)	14.4 (2.9)	11.3 (3.3)	23.0 (6.7)
HC	64.6 (5.5)	27.4 (3.3)	-	-	4.8 (0.8)	9.0 (1.6)

6.3.2 Motion Analysis

Each participant was instrumented with 63 reflective markers to obtain whole-body kinematics during the step ascent and descent tasks. Three-dimensional positions of the markers with respect to the inertial origin were recorded from eight near-infrared cameras (100 Hz sampling frequency) (Vicon, Centennial, CO). Each participant performed three bilateral step ascent and descent trials onto a 20-cm platform. No instructions were provided to the participants regarding the speed at which to complete each task. Data were averaged across the three trails and used for between-limb and between-group comparisons.

6.3.3 Data Analysis

Marker position data were low-pass filtered with a 4th order Butterworth filter (6 Hz cutoff frequency). A 15-segment subject model was created in Visual 3D (C-Motion, Germantown, MD) (Gaffney et al., 2016). Intact segment inertial parameters were based on regression equations of segment geometry (Dempster, 1955) and inertial parameters of

the prosthetic shank (residual limb + prosthetic socket) and prosthetic foot were measured using a reaction board and oscillation method (Smith et al., 2014).

Trunk translational angular momentum (TAM) with respect to the leading stance foot described as:

$${}^I \mathbf{h}_{\text{Trunk/Foot}} = \mathbf{r}_{\text{Trunk/Foot}} \times m_{\text{Trunk}} {}^I \mathbf{v}_{\text{Trunk/Foot}} \quad (6.1)$$

where $\mathbf{r}_{\text{Trunk/Foot}}$ is the position vector of the trunk relative to the stance foot, ${}^I \mathbf{v}_{\text{Trunk/Foot}}$ is the velocity of the trunk COM relative to the stance foot as observed in an inertial reference frame I , and m_{Trunk} is the mass of the trunk. The time derivative of trunk TAM is an expression of Euler's 1st Law in angular momentum form:

$$\frac{{}^I d}{dt} ({}^I \mathbf{h}_{\text{Trunk/Foot}}) = \mathbf{M}_{\text{Trunk/Foot}} + (\mathbf{r}_{\text{Trunk/Foot}} \times m_{\text{Trunk}} {}^I \mathbf{a}_{\text{Foot}}) \quad (6.2)$$

where $\mathbf{r}_{\text{Trunk/Foot}} \times m_{\text{Trunk}} {}^I \mathbf{a}_{\text{Foot}}$ is the corrective inertial moment of the trunk relative to the stance foot and is required to satisfy Euler's laws when the foot accelerates during the task. The translational trunk segmental moment about the foot, expressed as:

$$\mathbf{M}_{\text{Trunk/Foot}} = \mathbf{r}_{\text{Trunk/Foot}} \times \mathbf{F}_{\text{Trunk}}^{\text{Ext}} \quad (6.3)$$

where $\mathbf{F}_{\text{Trunk}}^{\text{Ext}}$ is the net of all external forces applied to the trunk.

Trunk rotational angular momentum (RAM) is described as:

$${}^I \mathbf{h}_{\text{Trunk}} = \mathbf{I}_{\text{Trunk}} \cdot \boldsymbol{\omega}_{\text{Trunk}} \quad (6.4)$$

where $\mathbf{I}_{\text{Trunk}}$ and $\boldsymbol{\omega}_{\text{Trunk}}$ are the inertial tensor and angular velocity of the trunk, respectively. The time derivative of trunk RAM is an expression of Euler's 2nd Law:

$$\frac{J}{dt} d \left({}^J \mathbf{h}_{\text{Trunk}} \right) = \mathbf{M}_{\text{Trunk}}^{\text{Ext}} \quad (6.5)$$

The right hand side of Eq. (6.5) is the rotational trunk moment expressed as:

$$\mathbf{M}_{\text{Trunk}}^{\text{Ext}} = \sum_{i=1}^N \mathbf{M}_i^{\text{Ext}} + \sum_{i=1}^N \left(\mathbf{r}_{i/\text{Trunk}} \times \mathbf{F}_i^{\text{Ext}} \right) \quad (6.6)$$

where i is the distal and proximal locations of forces and moments applied to the trunk.

Segmental kinetic effort was defined as the net moment created by gravity, joint reaction forces, and muscle forces to generate and arrest segmental angular momentum. This relationship between angular momentum and segmental moment is indicated in Euler's Second Law (Rao, 2006; Kasdin & Paley, 2011). The net segmental rotational moment is the sum of the applied (external) proximal and distal joint moments to the moments about the segment COM due to proximal and distal forces (Gaffney et al., 2017).

To facilitate anatomically planar analyses, all momenta and moment vectors were expressed in a basis with respect to the path of the body COM: $\mathbf{e}_{\text{frontal}}$ (tangent to the horizontal path of the body COM), $\mathbf{e}_{\text{transverse}}$ (opposite direction of the gravity), and $\mathbf{e}_{\text{sagittal}}$ ($\mathbf{e}_{\text{frontal}} \times \mathbf{e}_{\text{transverse}}$) (Figure 6.1).

	<u>Sagittal Plane</u>	<u>Frontal Plane</u>	<u>Transverse Plane</u>
Positive Momentum	Momentum generated posteriorly with respect to the stance foot	Mediolateral momentum generated toward the stance foot	Axial momentum generated away from the stance foot
Negative Momentum	Momentum generated anteriorly with respect to the stance foot	Mediolateral momentum generated away from the stance foot	Axial momentum generated toward from the stance foot

Figure 6.1. All momenta and moment vectors were expressed in a path reference frame that is defined by the path of the body COM (e_{sagittal} , e_{frontal} , $e_{\text{transverse}}$).

All trunk translational and rotational moments were normalized by time during the loading period (leading limb heel strike (0%) to trailing limb heel strike (100%)). The functional sub-phases of the step ascent and descent used for qualitative description of timing were based on the sub-phases defined by Zachazewski et al., (1993) and normalized to the loading period (Figure 6.2).

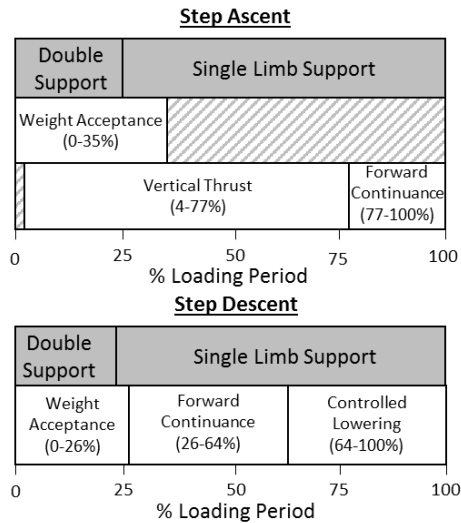


Figure 6.2. Tasks (double and single limb support) and functional phases of the step ascent and descent expressed as a percentage of the loading period (leading limb heel strike to trailing limb heel strike), as defined by Zachazewski et al., (1993).

6.3.4 Statistical Analysis

To quantify the effort needed to generate or arrest trunk angular momentum we identified global minima and maxima of the trunk translational and rotational moments in all three planes during the step ascent and descent trials (dependent variables). Three one-way mixed-factor models were used for each dependent variable: *between* subjects (amputated vs. HC and intact vs. HC) and *within* subjects (amputated vs. intact) while controlling for differences in height and mass (covariates). When statistically significant differences were found, pairwise comparisons with a Bonferroni adjustment for multiple comparisons were used ($\alpha_B = 0.05/3 = 0.017$). For the HC group, only trials performed on the right limb were used for comparison. To quantify the amount of change between dependent variables, Hedges' *g* effect size and bootstrapped 95% confidence intervals were calculated (Hedges, 1981; Hentschke & Stuttgen, 2011; Halsey et al., 2015) and

categorized as small effect ($g \leq 0.2$), medium effect ($0.2 < g < 0.8$), or large effect ($g \geq 0.8$). Only confidence intervals that did not cross zero were considered to be statistically significant (Curran-Everett, 2009).

6.4 Results

6.4.1 Step Ascent

In the sagittal plane, the peak posterior translational trunk moment (positive) was larger in the TTA group with the amputated limb than the intact limb during vertical thrust of ascent ($P = 0.01$, $g = 1.52$ [1.16 2.64]) (Figure 6.3).

In the frontal plane, the peak mediolateral translational trunk moment toward the leading stance foot (positive) was larger in the TTA group when leading with the amputated limb compared to the HC group during weight acceptance ($P = 0.01$, $g = 1.8$ [1.17 3.53]) (Figure 6.3). Peak mediolateral rotational trunk moment toward the leading stance foot (positive) was larger in the TTA group when leading with the intact limb compared to the HC group at the beginning of weight acceptance ($P < 0.01$, $g = 1.56$ [0.89 3.52]) (Figure 6.4).

In the transverse plane, peak axial translational moment toward the leading stance foot (negative) was larger in the TTA group when leading with either the amputated or intact limb compared to the HC group during weight acceptance ($P < 0.01$, $g = 2.36$ [1.62 5.01]; $P = 0.01$, $g = 1.47$ [1.01 2.94], respectively) (Figure 6.3). The peak axial rotational moment away from the leading stance foot (positive) was larger in the TTA group when leading with either the amputated or intact limb compared to the HC group during

vertical thrust ($P = 0.015$, $g = 1.50$ [1.18 3.46]; $P < 0.01$, $g = 2.52$ [1.85 5.59] respectively) (Figure 6.4).

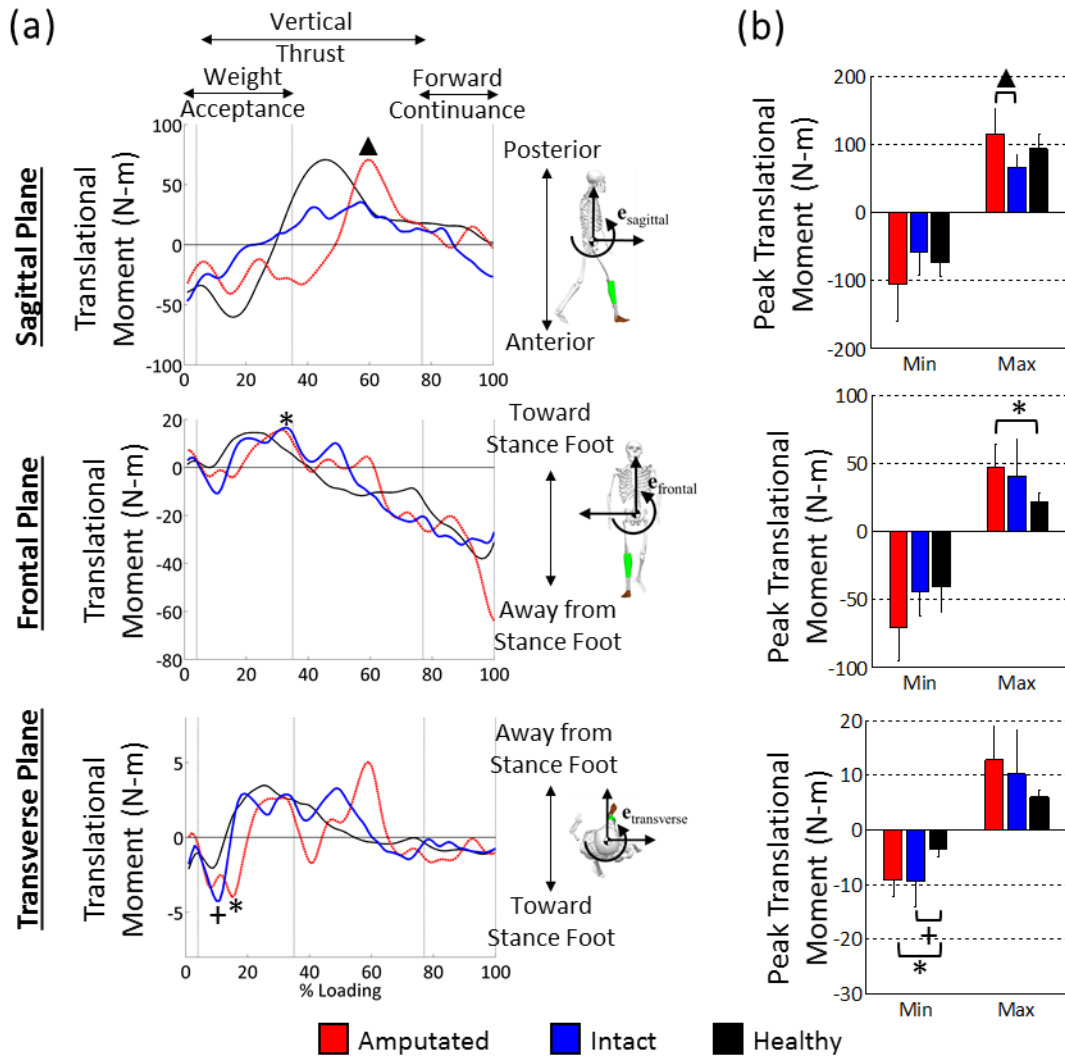


Figure 6.3. Trunk translational moment about the leading stance foot (a) mean ensemble averages and (b) mean (1 SD) peak (minimum and maximum) during the step ascent on the amputated limb (red), intact limb (blue), and right limb of healthy controls (black). Significant differences ($P < 0.017$) within and across groups are as follows: amputated vs. intact limb (▲), amputated limb vs. healthy controls (*), and intact limb vs. healthy controls (+).

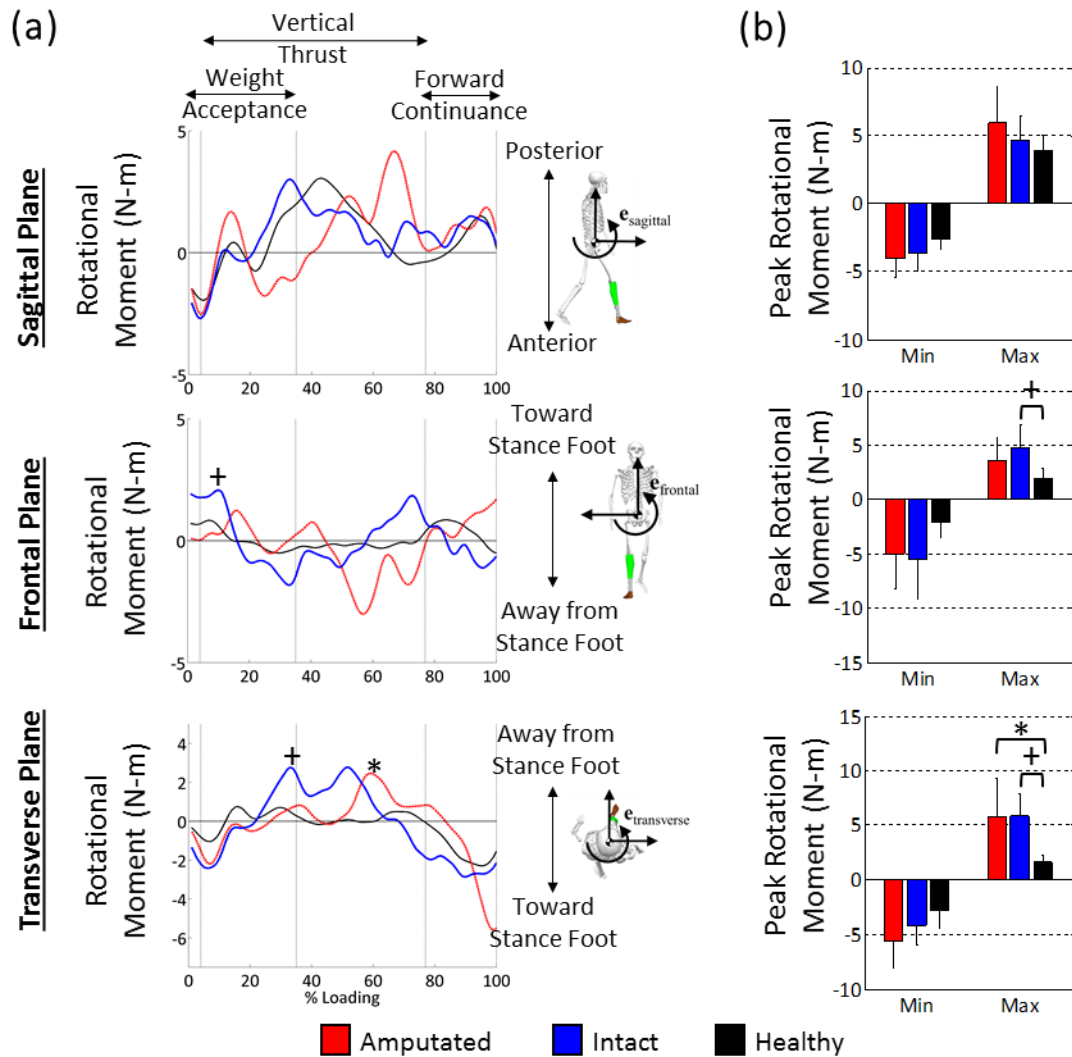


Figure 6.4. Trunk rotational moment about the leading stance foot (a) mean ensemble averages and (b) mean (1 SD) peak (minimum and maximum) during the step ascent on the amputated limb (red), intact limb (blue), and right limb of healthy controls (black). Significant differences ($P < 0.017$) within and across groups are as follows: amputated vs. intact limbs (\blacktriangle), amputated limb vs. healthy controls ($*$), and intact limb vs. healthy controls ($+$).

6.4.2 Step Descent

In the sagittal plane, peak anterior (negative) and posterior (positive) translational trunk moments were larger in the TTA group when leading with the intact limb compared

to the HC group ($P < 0.01$, $g = 1.83$ [1.48 3.70]; $P = 0.01$, $g = 1.16$ [0.40 3.16], respectively) (Figure 6.5). Peak anterior rotational trunk moment (negative) was larger in the TTA group when leading with the intact limb than the HC group during weight acceptance ($P = 0.017$, $g = 0.99$ [0.10 3.42]) (Figure 6.6).

In the frontal plane, peak translational moment away from the leading stance foot (negative) was larger in the TTA group when leading with either the amputated or intact limb compared to the HC group during weight acceptance ($P < 0.01$, $g = 1.70$ [1.14 3.55]; $P < 0.01$, $g = 1.51$ [0.53 4.93], respectively). The peak translational moment toward the leading stance foot (positive) was larger in the TTA group when leading with the intact limb than the HC group at the beginning of forward continuance ($P = 0.01$, $g = 1.16$ [0.40 3.16]). The peak mediolateral rotational trunk moment away from the leading stance foot (negative) was larger in the TTA group when leading with the intact limb compared to the HC group ($P < 0.01$, $g = 3.52$ [1.99 11.86]). The peak mediolateral rotational trunk moment toward the leading stance foot (positive) was larger in the TTA group when stepping onto either the amputated or intact limb compared to the HC group ($P < 0.01$, $g = 2.11$ [1.19 5.07]; $P < 0.01$, $g = 2.06$ [1.77 8.87], respectively) (Figure 6.6).

In the transverse plane, peak axial rotational trunk moment toward the leading stance foot (negative) was larger in the TTA group when stepping onto either the amputated or intact limb compared to the HC group ($P = 0.01$, $g = 1.13$ [0.01 3.78]; $P = 0.017$, $g = 1.45$ [0.30 4.28], respectively) (Figure 6.6).

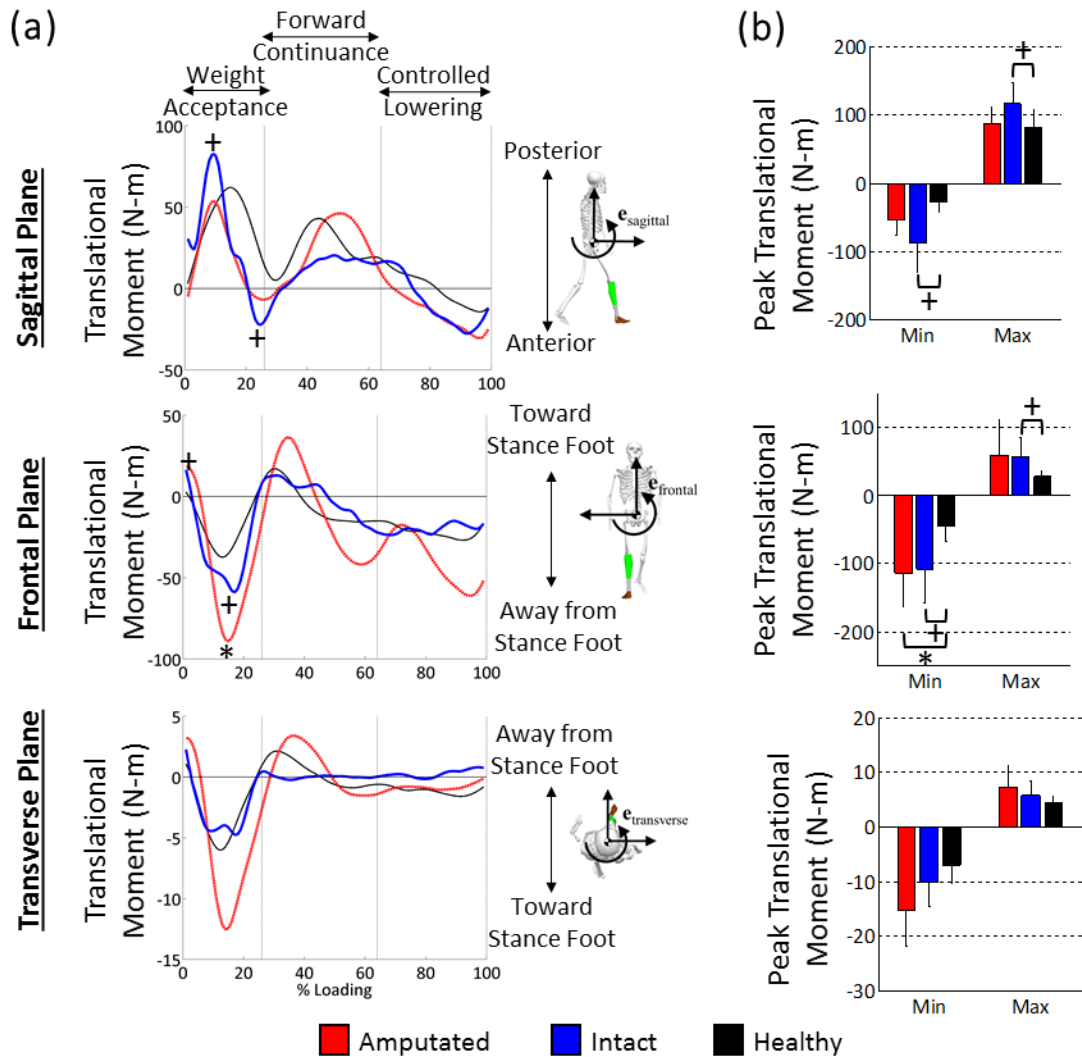


Figure 6.5. Trunk translational moment about the leading stance foot (a) mean ensemble averages and (b) mean (1 SD) peak (minimum and maximum) during the step descent on the amputated limb (red), intact limb (blue), and right limb of healthy controls (black). Significant differences ($P < 0.017$) within and across groups are as follows: amputated vs. intact limbs (\blacktriangle), amputated limb vs. healthy controls (*), and intact limb vs. healthy controls (+).

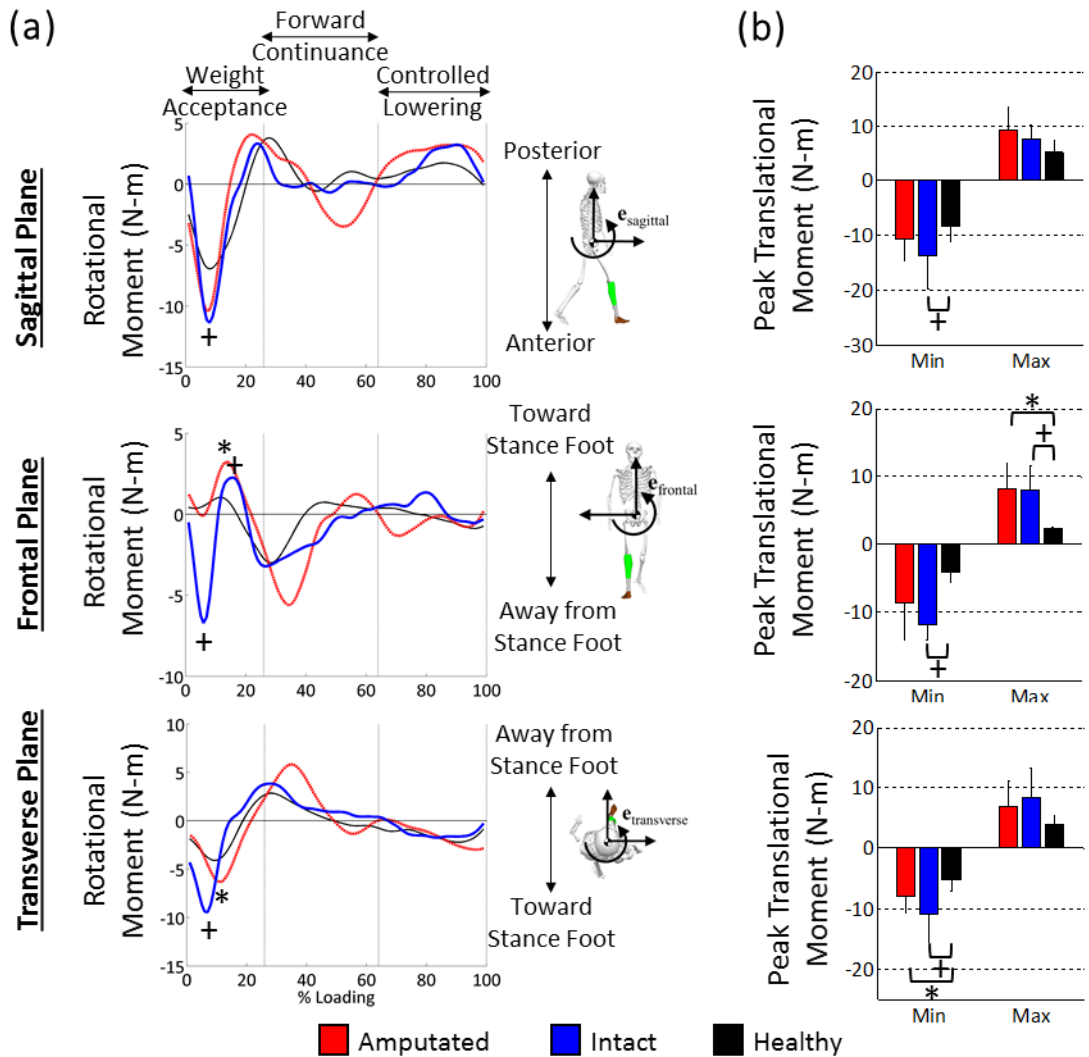


Figure 6.6. Trunk rotational moment about the leading stance foot (a) mean ensemble averages and (b) mean (1 SD) peak (minimum and maximum) during the step descent on the amputated limb (red), intact limb (blue), and right limb of healthy controls (black). Significant differences ($P < 0.017$) within and across groups are as follows: amputated vs. intact limbs (\blacktriangle), amputated limb vs. healthy controls (*), and intact limb vs. healthy controls (+).

6.5 Discussion

The purpose of this study was to determine how movement compensations altered the required trunk kinetic effort during step ascent and descent in patients with unilateral

transtibial amputation and healthy controls by analysis of translational and rotational segmental moments. Changes in segmental moments are created by simultaneous changes in gravitational moments, joint reaction forces, and joint moments (moments created by muscle forces). We measured differences between groups in trunk translational and rotational segmental moments in three anatomical planes during step ascent and descent tasks across limbs and across groups, which represent differences in the effort required to successfully complete the task. During initial training following amputation, patients are often instructed to ascend steps leading with the intact limb and descend steps leading with the amputated limb (Jones et al., 2005, 2006; Schmalz et al., 2007; Alimusaj et al., 2009; Barnett et al., 2014). These trained patterns may influence the long-term gait patterns during step ambulation. Our results indicate that patients with amputation adopt unique strategies dependent upon which limb is being loaded, which alters the kinetic effort of the trunk required for successful completion of the task.

6.5.1 Step Ascent

In the sagittal plane, patients with amputation demonstrated a larger posterior trunk translational moment when stepping up with the amputated limb than the intact limb, which may increase the demand on the back extensor muscles (e.g. multifidus, erector spinae). The increase in posterior translational moment is a result of increased external joint reaction forces applied to the trunk at the low back that originate from the push off forces from the intact limb. When stepping onto the amputated limb, patients with amputation adopt a ‘hip-extensor dominant’ strategy in the intact limb to elevate the body COM (Powers et al., 1997; Yack et al., 1999; Schmalz et al., 2007). This strategy creates

anteriorly directed net external joint forces at the hip and low back and posterior motion of the trunk. A large posterior trunk translational moment when stepping up with the amputated limb makes the peak posterior moment late in the vertical thrust phase (Figure 6.3a) necessary to arrest the trunk momentum and maintain balance.

Asymmetric low back loading is associated with an increased risk of LBP in the frontal plane (Davis & Marras, 2000); therefore, compensations identified during step ambulation may have potential long-term adverse effects on the low back through repetitive increased and asymmetric loading. Patients with amputation demonstrated compensations in the trunk translational and rotational moments that are likely used for stability (Jones et al., 2005). In addition, patients with amputation use the momentum of the trunk, generated by increasing concentric muscle activation of the intact limb hip abductors (Nadeau et al., 2003), to elevate the pelvis and avoid contact between the amputated swing limb and the step. When stepping up with the amputated limb, the translational trunk moment toward the stance limb was larger in comparison healthy controls, which is consistent with previous findings of high laterally-directed low back joint reaction shear forces measured in patients with amputation (Hendershot & Wolf, 2014). When stepping onto the intact limb, patients with amputation demonstrated larger rotational trunk moments toward the stance limb early during early weight acceptance. The rotational segment moments are used in iterative Newton-Euler inverse dynamic analyses to calculate joint moment; therefore, the current findings are likely consistent with our previous findings (Murray et al., *In Review*), which found increased lateral bend moments coupled with increased frontal plane trunk displacement directed toward the

stance limb during step ambulation in the current amputation group. Increased trunk displacement toward the stance limb is consistent with a compensated Trendelenburg pattern and is hypothesized to improve lateral balance in patients with lower-limb amputation (Michaud et al., 2000; Tura et al., 2010; Molina-Rueda et al., 2014). However, the compensated Trendelenburg pattern increases demand on the spine and surrounding musculature which may increase the risk of developing LBP (Hendershot et al., 2013).

In the transverse plane, patients with amputation ascend steps with bilateral movement strategies that require increased trunk kinetic effort compared to healthy subjects. Patients with amputation increased axial rotational moments away from the stance limb, which may be a strategy required to arrest momentum. However, the timing of peak axial rotational moment away from the stance limb differs between limbs, indicating limb-dependent strategies. When stepping onto the intact limb, peak axial rotational trunk moment away from the stance foot occurs earlier in the vertical thrust phase compared to the amputated limb, which is likely an aggregate effect of increased rotation due to the hip strategy required to elevate the body COM onto the step in the absence of active ankle plantar flexion from the trailing (amputated) limb. However, when stepping onto the amputated limb, the peak axial rotational trunk moment away from the stance limb occurs later in the vertical thrust phase, which is likely result in the delayed and increased ground reaction forces created from the trailing (intact) limb (Schmalz et al., 2007).

6.5.2 *Step Descent*

In the sagittal plane, patients with amputation demonstrated a higher posterior trunk translational moment when stepping onto the intact limb than the amputated limb which provides insight into the kinetic strategies used to achieve forward progression. This higher in translational moment is caused by a higher in vertical ground reaction force (GRF) that propagate up the kinetic chain, and is associated with ‘falling’ onto the intact limb with limited control. This strategy is commonly adopted by patients with amputation when stepping onto the intact limb because of reduced ability to actively control the lowering of the body COM with the amputated limb (Schmalz et al., 2007). During weight acceptance, the amputation group had greater trunk rotational moment when stepping onto either the amputated or intact limbs in comparison to the healthy group, which is consistent with an anterior trunk lean strategy over the leading stance limb. This strategy reduces the demand on the trailing limb quadriceps muscles during loading, but may have long-term detrimental effects due to increased demand placed on the trunk and hip extensor musculature (Hendershot & Wolf, 2014).

In the frontal plane, similar to step ascent, patients with amputation increased trunk rotational moments in the direction toward the stance limb (compensated Trendelenburg) compared to healthy controls. However, in contrast to step ascent, this strategy occurred when leading with either the amputated or intact limb. Patients with amputation may employ this strategy bilaterally during step descent to compensate for hip abductor weakness (Molina-Rueda et al., 2014) and maximize stability during this highly destabilizing task (Michaud et al., 2000; Tura et al., 2010). Although the compensated

Trendelenburg pattern assists with maintaining balance, the excessive and abrupt loading may have negative long-term consequences on the spine and low back musculature.

In the transverse plane, patients with amputation demonstrated 74% (amputated limb) and 42% (intact limb) larger trunk rotational moments toward the stance limb during weight acceptance onto either limb compared to healthy controls. This difference demonstrates that the loading strategies adopted by patients with amputation increase the kinetic effort required for successful completion, regardless of the limb being loaded. When stepping onto the amputated limb, the increased trunk rotational moment is likely an effect of the inability to arrest axial rotation with the prosthetic limb. When stepping onto the intact limb, the increased trunk rotational moment is likely an effect of the increased momentum from ‘falling’ off of the step and landing abruptly. Lack of axial control of trunk rotation is frequently linked to the development of LBP (Van Dieën et al., 2003; van den Hoorn et al., 2012), and therefore indicates that these adaptations may have potential long term adverse effects.

6.5.3 Limitations

Several limitations to this investigation should be considered. First, this investigation included a small sample of individuals with dysvascular TTA, which may limit generalizability to individuals with other types of TTA (e.g. traumatic, oncologic, congenital). Second, the TTA group was not screened for LBP at the time of testing; therefore, we cannot conclude if compensatory movement patterns observed were habitual or a result of LBP. However, we do not expect this to have a confounding effect on the present results because no participants reported LBP at the time of testing. Finally,

the step ascent and descent trials did not consist of alternating stairs; therefore, the compensatory movement patterns during each task may not indicate strategies that are required for ascending and descending stairs.

6.6 Conclusion

This investigation identified the trunk movement compensations that alter effort during step ascent and descent in patients with unilateral transtibial amputation by identifying the translational and rotational trunk moments. Trunk compensations are required to successfully ascend and descend steps without active plantarflexion, but may have long-term adverse effects through the increased and asymmetric demand placed on the low back musculature. It remains unclear what level of trunk movement compensation that can be used to compensate for the loss of active ankle plantarflexion without having potential adverse effects through increased demand on the low back.

**CHAPTER 7: TRUNK MOVEMENT COMPENSATIONS AND ALTERATIONS
IN CORE MUSCLE DEMAND DURING STEP AMBULATION IN PATIENTS
WITH UNILATERAL TRANSTIBIAL AMPUTATION**

7.1 Abstract

The objective of this investigation was to identify demands from core muscles that corresponded with trunk movement compensations during bilateral step ambulation in people with unilateral transtibial amputation (TTA). Trunk rotational angular momentum (RAM) was measured using motion capture and bilateral surface EMG was measured from four bilateral core muscles during step ascent and descent tasks in people with TTA and healthy controls. During step ascent, the TTA group generated larger mediolateral ($P = 0.01$) and axial ($P = 0.01$) trunk RAM toward the leading limb when stepping onto the intact limb than the control group, which corresponded with high demand from the bilateral erector spinae and oblique muscles. During step descent, the TTA group generated larger trunk RAM in the sagittal ($P < 0.01$), frontal ($P < 0.01$), and transverse planes ($P = 0.01$) than the control group, which was an effect of falling onto the intact limb. To maintain balance and arrest trunk RAM, core muscle demand was larger throughout the loading period of step descent in the TTA group. However, asymmetric trunk movement compensations did not correspond to asymmetric core muscle demand during either task, indicating a difference in motor control compensations dependent on the leading limb.

7.2 Introduction

Rehabilitation following transtibial amputation (TTA) aims to maximize functional independence by retraining movement patterns during gait to lower the risk of secondary overuse injuries, such as low back pain (LBP) (Cutson and Bongiorno, 1996). Although the underlying mechanisms behind the development of LBP following amputation remains idiopathic, asymmetric trunk movement patterns are known to increase the risk of LBP (Kumar, 2001; McGill, 2007). Trunk movement asymmetry during gait is often associated with asymmetric core muscle demand, which can indicate poor control of the trunk muscles (Tsao et al., 2008). Thus, clinicians target interventions to minimize asymmetric trunk compensations adopted in the absence of ankle plantarflexion after TTA. However, rehabilitation in people with dysvascular TTA is exceedingly complex due to common neurovascular comorbidities and poor physical health (Cutson and Bongiorno, 1996), which compounds the risk of LBP (Shiri et al., 2010).

To achieve independence of mobility, people with TTA must daily perform high-demand functional tasks such as step ambulation, which can be highly challenging (Nadeau et al., 2003; De Laat et al., 2013). To maximize early functional recovery, step ambulation retraining consists of instructing patients to ascend steps leading with the intact limb and descend steps leading with the amputated limb (Barnett et al., 2014; Schmalz et al., 2007). During both step ascent and descent, people with TTA typically adopt a “quadriceps avoidance” strategy, which improves trunk and pelvic stability, and has been linked with a forward trunk lean (although not directly measured) (Schmalz et al., 2007). We recently associated changes in trunk kinematics with large low back

internal extension moments in people with TTA compared to healthy participants during step ambulation (Murray et al., *In Review*). However, the muscle demand resulting from trunk movement compensations during stepping tasks after TTA remains unknown. In addition, it is unknown if motor control compensations differ between trained and novel tasks, which likely affect a patient's ability to navigate unexpected obstacles to avoid a fall through postural control (Marigold et al., 2005).

We previously quantified asymmetric trunk movement compensations in people with TTA using segmental angular momentum during walking (Gaffney et al., 2016). People with TTA generate larger amounts of trunk rotational angular momentum (RAM) compared to healthy people of similar age. Segmental RAM is foundational in joint moments calculated via inverse dynamics (Gaffney et al., 2017), which represent the net effect of all agonist and antagonist muscle activity spanning a joint. Therefore, we interpreted our previous findings of larger asymmetric generation of segmental RAM to potentially correspond with movement strategies that result in high asymmetric eccentric muscle demand needed to arrest segmental momentum. However, muscle demand was not included in these level-ground walking analyses, and it is not clear that asymmetric generation of momentum corresponds with asymmetric muscle demand.

The primary objective of this investigation was to identify demands from core trunk/pelvis muscles that correspond with trunk movement compensations, quantified with segmental RAM, during step ascent and descent in people with unilateral TTA. The secondary objective was to establish if asymmetric movement compensations corresponded with asymmetric core muscle demand. We hypothesized that trunk

movement compensations adopted by people with TTA would correspond with higher muscle demand than seen in healthy individuals with symmetrical trunk movement. We also hypothesized that differences across limbs in the TTA group would identify motor control compensations that were habituated following amputation.

7.3 Methods

7.3.1 Participants

Ten male participants with unilateral dysvascular TTA and ten male healthy control (HC) participants were enrolled (Table 7.1). Eligibility for both groups included a BMI less than 40 and an age between 50-85 years; and for the TTA group included amputation due to a neurovascular pathology in the previous one to three years and the ability to walk for four minutes without rest. Each participant visited the laboratory for one data collection in which whole body kinematics and core muscle activity were collected during bilateral stepping tasks. Three of the ten TTA participants were unable to perform the tasks bilaterally, and were excluded from the analysis. Each participant provided written, informed consent in accordance with the Colorado Multiple Institutional Review Board. Further details regarding individual participant anthropometrics and levels of functional performance can be found in Appendices A and B, respectively.

Table 7.1. Participant characteristics for patients with dysvascular unilateral transtibial amputation (TTA) and healthy control (HC) groups.

Group	Age (Years)	BMI (kg/m ²)	Time since Amputation (Months)	Residual Limb Length (cm)	Socket Type	Prosthetic Foot
TTA	56.3 ± 4.5	28.3 ± 2.7	16.7 ± 5.2	14.4 ± 2.9	Total contact carbon fiber	Dynamic elastic response
HC	64.6 ± 5.5	27.4 ± 3.3	N/A	N/A	N/A	N/A

7.3.2 Instrumentation and Experimental Protocol

Each participant was instrumented with 63 reflective markers used to obtain whole-body kinematics during step ascent and descent (Gaffney et al., 2016). Marker trajectories were recorded from eight infrared cameras surrounding the motion capture volume (100 Hz sampling frequency) (Vicon, Centennial, CO).

Core muscle activity was recorded from round bipolar Ag/AgCl surface electrodes (inter-electrode distance: 1cm; CMRR > 100 db) (Vermed, Buffalo, NY) placed on the bilateral erector spinae (ESPN), external oblique (OBL), gluteus maximus (GMAX), and gluteus medius (GMED) according to SENIAM guidelines (2000 Hz sampling frequency) (Merletti and Hermens, 2000) (Measurement Systems Inc., Livonia, MI). A ground electrode was placed at the C7 vertebrae. Prior to electrode placement, the skin was prepared by shaving and lightly abrading using sterilizing alcohol wipes. Maximum voluntary contractions (MVCs) for these muscles were measured across three trials for approximately three seconds each. For ESPN MVCs, each participant laid prone and were instructed to extend their trunk while resisting a downward force applied at the

shoulders. For OBL MVCs, each participant was seated and instructed to rotate their trunk while resisting a force opposite the direction of the axial rotation applied at the ipsilateral shoulder of the OBL of interest. For GMAX MVCs, each participant laid prone and were instructed to extend their hip while resisting a downward force applied above the knee. For GMED MVCs, each participant laid side-lying and were instructed to abduct their hip while resisting a downward force applied above the knee.

Each participant performed three step ascent and descent trials with each limb onto a 20-cm platform. No instructions were provided regarding the speed at which to complete each task.

7.3.3 Data Analysis

Trunk movement compensations were quantified using RAM and muscle demand was quantified using surface EMG measured from the four bilateral core muscles. Kinematic data were low-pass filtered with a 4th-order Butterworth filter (6 Hz cutoff frequency). A 15-segment subject-specific model was created in Visual 3D and used for analysis (C-Motion, Inc., Germantown, MD).

Trunk RAM is described as:

$${}^J \mathbf{h}_{\text{Trunk}} = \mathbf{I}_{\text{Trunk}} \cdot \boldsymbol{\omega}_{\text{Trunk}} \quad (7.1)$$

where $\mathbf{I}_{\text{Trunk}}$ and $\boldsymbol{\omega}_{\text{Trunk}}$ are the inertial tensor and angular velocity of the trunk, respectively. To facilitate anatomically planar analyses, all momenta vectors were expressed in a basis with respect to the path of the body COM (Figure 7.1).

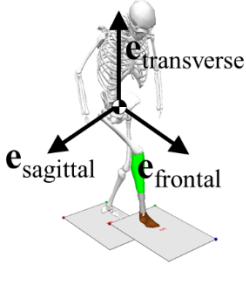
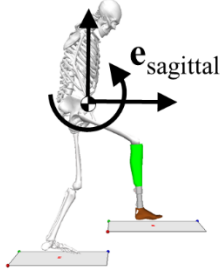
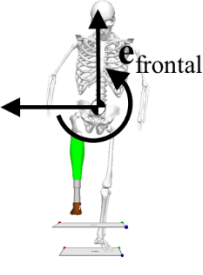
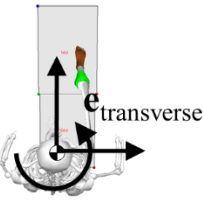
	Sagittal Plane	Frontal Plane	Transverse Plane
			
Positive Rotation	<i>Posterior</i> rotation with respect to the stance foot	Mediolateral rotation toward the stance foot	Axial rotation away from the stance foot
Negative Rotation	<i>Anterior</i> rotation with respect to the stance foot	Mediolateral rotation away from the stance foot	Axial rotation toward from the stance foot

Figure 7.1. All momenta vectors were expressed in a path reference frame that is defined by the path of the body COM (e_{sagittal} , e_{frontal} , $e_{\text{transverse}}$).

EMG data were band-pass filtered with a 4th-order Butterworth filter (10-350 Hz cutoff frequencies) and smoothed to create a linear envelope using full-wave rectification and a 4th-order zero phase-lag Butterworth filter (6 Hz cutoff frequency). To reduce variability across participants of muscle activation, EMG signals were normalized by the maximum value obtained across the three MVC trials of the respective muscle.

All dependent variables were calculated and normalized to the loading period (leading limb foot strike (0%) to trailing limb foot strike (100%)). Within the loading period, the functional phases of the step ascent and descent were based on the sub-phases previously defined by Zachazewski et al., (1993) (Figure 7.2).

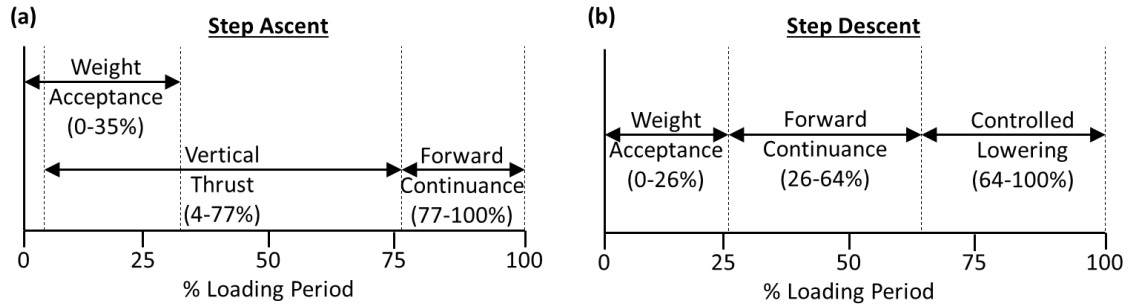


Figure 7.2. Functional phases of the (a) step ascent and (b) step descent tasks expressed as a percentage of the loading period (leading limb foot strike to trailing limb foot strike) (Zachazewski et al., 1993).

Muscle demand was quantified by integrating the normalized EMG data:

$$iEMG_{\text{muscle}} = \int_{t_0}^{t_f} nEMG_{\text{muscle}} dt \quad (7.2)$$

where $nEMG_{\text{muscle}}$ is the linear envelope EMG signal normalized to the maximum linear envelope of the MVC of the respective muscle, and t_0 and t_f are the integrating bounds of $iEMG$. To assess where the differences in muscle demand occurred in each task, $iEMG$ for each muscle was calculated in each functional phase (Figure 7.2).

7.3.4 Statistical Analysis

Trunk RAM and $iEMG$ data were averaged across the three trails for each participant and used for comparison.

Task completion times were compared across groups (leading with the amputated limb vs. HC and leading with the intact limb vs. HC) using two-tailed independent t -tests ($\alpha = 0.05$).

To quantify the demand from core muscles to support trunk movement, we identified the peak (minimum and maximum) trunk RAM in all three planes and $iEMG$ of each

muscle within each phase of the step ascent and descent trials (dependent variables). Dependent variables were compared using three one-way mixed-factor models: *between* subjects (amputated limb ascent vs. HC and intact limb ascent vs. HC) and *within* subjects (amputated vs. intact limb ascent) while accounting for differences in height and mass across groups in momenta comparisons (covariates). When a significant main or interaction effect occurred, pairwise comparisons with a Bonferroni adjustment for multiple comparisons were used to determine significance ($\alpha_B = 0.05/3 = 0.017$). For the HC group, no differences were found between limbs; and therefore, only trials performed on the right limb were used for statistical comparison. To quantify the change between dependent variables, Hedges' g effect size and bootstrapped 95% confidence intervals were calculated (Hentschke & Stuttgen, 2011) and categorized as small ($g \leq 0.2$), medium ($0.2 < g < 0.8$), and large effect ($g \geq 0.8$).

7.4 Results

7.4.1 Task Completion Time

No differences in task completion time existed between the TTA and HC groups or between limbs in the TTA group during step ascent or step descent (Figure 7.3).

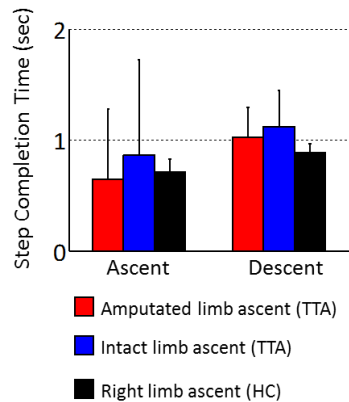


Figure 7.3. Mean (1 SD) step ascent and descent completion times (leading limb foot strike to trailing limb foot strike) for the TTA and HC groups.

7.4.2 Step Ascent

7.4.2.1 Trunk Rotational Angular Momentum (RAM)

In the frontal plane, during vertical thrust phase of step ascent, the peak trunk RAM away from the leading stance foot (negative) was larger in the TTA group when leading with the intact limb than HC ($P = 0.01$, $g = 1.84$ [1.34 4.24]) (Figure 7.4). During forward continuance, the peak trunk RAM away from the stance foot was greater in the TTA group when stepping onto the amputated limb than HC ($P = 0.01$, $g = 2.51$ [1.41 4.87]) (Figure 7.4).

In the transverse plane, during weight acceptance, the peak trunk RAM toward the leading stance foot (negative) was greater in the TTA group when leading with both the amputated and intact limbs than HC ($P = 0.01$, $g = 1.85$ [1.05 4.18]; $P = 0.01$, $g = 1.96$ [1.49 3.88], respectively) (Figure 7.4). During vertical thrust, the peak trunk RAM toward the leading stance foot (negative) was greater in the TTA group when leading with both the amputated and intact limb ($P < 0.01$, $g = 3.02$ [2.10 5.94]; $P < 0.01$, $g = 2.59$ [2.22 4.83], respectively) and the peak trunk RAM away from the leading stance foot (positive)

was greater in the TTA group when leading with the intact limb than HC ($P = 0.01$, $g = 1.78$ [1.08 5.16]) (Figure 7.4).

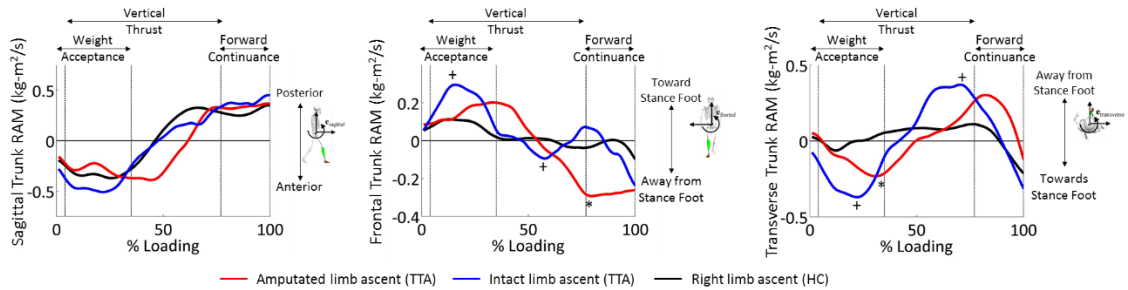


Figure 7.4. Ensemble averages of trunk rotational angular momentum (RAM) during the step ascent. Statistically significant differences ($P < 0.017$) within and across groups are: amputated vs. intact limbs (▲), amputated limb vs. healthy controls (*), and intact limb vs. healthy controls (+).

7.4.2.2 Integrated EMG (iEMG)

During the weight acceptance phase, the iEMG from the ESPN and OBL on the leading and trailing limb sides were larger in the TTA group when leading with the intact limb than HC. During the vertical thrust phase, the iEMG from the ESPN and OBL on the leading and trailing limb sides were larger in the TTA group when leading with both the amputated or intact limb than HC. During the forward continuance phase, the iEMG from the ESPN and OBL on the leading and trailing limb sides, as well as from the GMAX on the leading limb side, was larger in the TTA group when leading with both the amputated or intact limb than HC (Figure 7.5, Table 7.2).

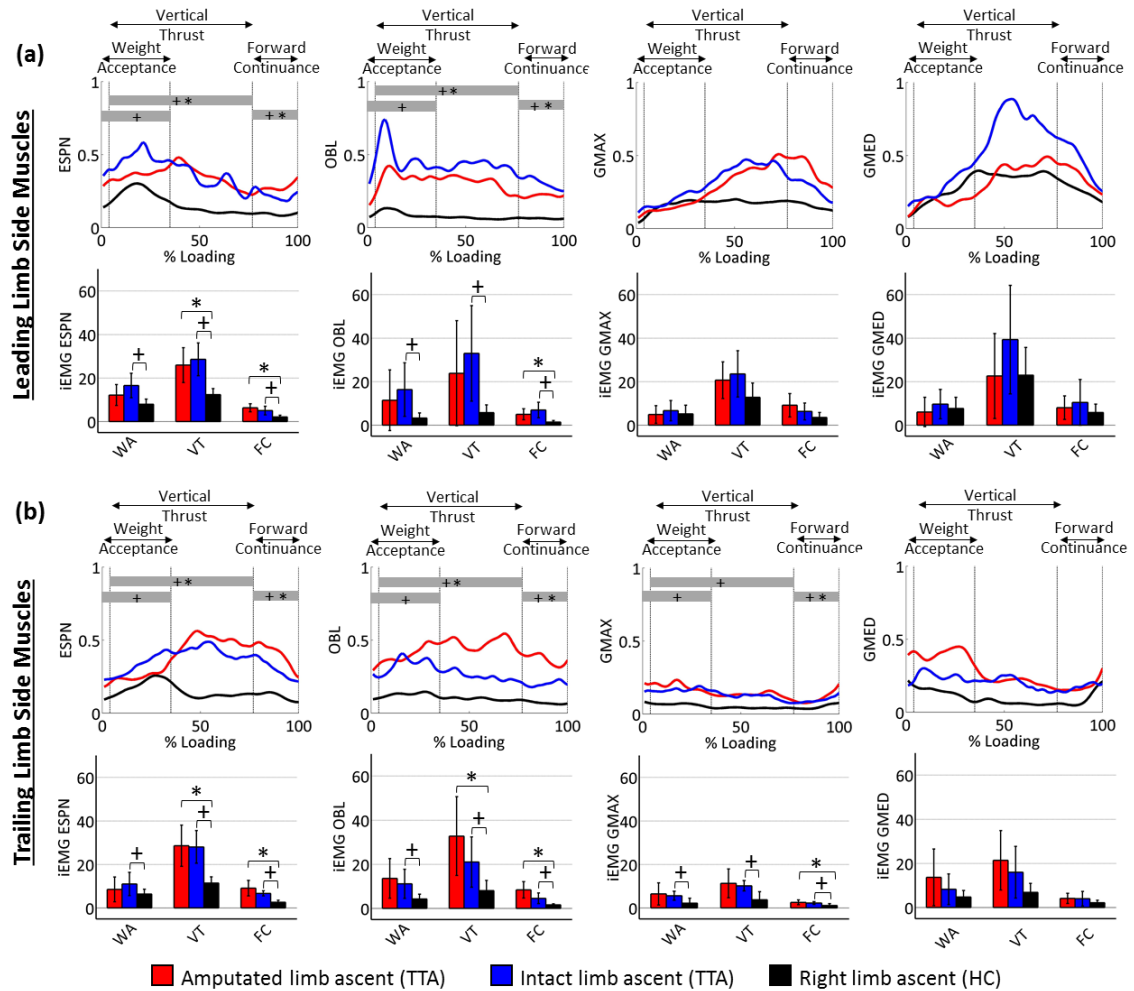


Figure 7.5. Ensemble averages of normalized linear envelope and integrated EMG (iEMG) of (a) leading limb side and (b) trailing limb side muscles during the phases of the step ascent: weight acceptance (WA), vertical thrust (VT), and forward continuance (FC). See Section 7.2.2 for specific muscle acronym definitions. Statistically significant differences ($P < 0.017$) within and across groups are: amputated vs. intact limbs (▲), amputated limb vs. healthy controls (*), and intact limb vs. healthy controls (+).

Table 7.2. Statistical comparisons across groups (intact limb vs. healthy control and amputated limb vs healthy control) and across limbs (amputated vs. intact limbs) during the functional phases of the step ascent. See Section 7.2.2 for specific muscle acronym definitions. Statistically significant differences ($P < 0.017$) of pairwise comparisons are noted as: intact limb vs. healthy controls (+), amputated limb vs. healthy controls (*), and amputated vs. intact limbs (\blacktriangle).

	<i>Muscle</i>	<i>Weight Acceptance</i>		<i>Vertical Thrust</i>		<i>Forward Continuance</i>	
		<i>P-value</i>	Hedge's <i>g</i>	<i>P-value</i>	Hedge's <i>g</i>	<i>P-value</i>	Hedge's <i>g</i>
<i>Intact Limb vs. Healthy Control</i>	Ipsilateral ESPN	<0.01 ⁺	1.87 [1.12 4.43]	<<0.01 ⁺	2.68 [2.04 5.16]	<0.01 ⁺	1.80 [1.40 6.26]
	Ipsilateral OBL	<0.01 ⁺	1.88 [1.24 3.89]	0.01 ⁺	1.62 [1.05 3.30]	<0.01 ⁺	2.02 [1.37 4.49]
	Ipsilateral GMAX	0.30	0.55 [-0.19 1.58]	0.041	1.14 [0.60 2.63]	0.13	0.82 [0.54 1.62]
	Ipsilateral GMED	0.22	0.64 [-0.07 1.71]	0.15	0.77 [0.21 1.69]	0.30	0.82 [0.54 1.62]
	Contralateral ESPN	0.02 ⁺	1.39 [0.69 2.95]	<0.01 ⁺	3.60 [2.55 8.22]	<0.01 ⁺	4.82 [3.79 8.63]
	Contralateral OBL	0.01 ⁺	1.44 [0.81 2.93]	0.01 ⁺	1.49 [0.80 3.04]	<0.01 ⁺	1.79 [1.35 3.31]
	Contralateral GMAX	0.01 ⁺	1.46 [0.82 4.49]	<0.01 ⁺	1.93 [1.02 5.95]	0.02 ⁺	0.015 [1.41 0.57]
	Contralateral GMED	0.17	0.74 [-0.16 1.73]	0.07	1.00 [0.36 2.23]	0.30	0.55 [-0.40 1.56]
<i>Amputated Limb vs. Healthy Control</i>	Ipsilateral ESPN	0.06	1.05 [0.49 2.33]	<0.01 [*]	2.12 [1.50 4.24]	<0.01 [*]	2.81 [1.67 6.73]
	Ipsilateral OBL	0.15	0.76 [0.49 2.60]	0.07	0.98 [0.71 2.83]	<0.01 [*]	1.81 [1.17 4.06]
	Ipsilateral GMAX	0.90	0.07 [-1.02 1.49]	0.07	0.97 [0.26 2.01]	0.03	1.27 [0.87 2.29]
	Ipsilateral GMED	0.63	0.25 [-0.68 1.76]	0.97	0.02 [-0.67 0.84]	0.40	0.44 [-0.06 1.45]
	Contralateral ESPN	0.37	0.47 [-0.83 1.53]	<0.01 [*]	2.30 [1.78 4.33]	<0.01 [*]	2.32 [1.75 4.70]
	Contralateral OBL	0.02	1.34 [0.83 2.58]	<0.01 [*]	1.78 [1.21 3.47]	<0.01 [*]	2.41 [1.94 4.81]
	Contralateral GMAX	0.07	1.00 [0.59 2.21]	0.022	1.32 [0.74 2.63]	0.02 [*]	1.39 [0.57 4.01]
	Contralateral GMED	0.10	0.89 [0.60 1.61]	0.02	1.67 [1.02 3.51]	0.07	1.01 [2.82 2.33]
<i>Amputated vs. Intact Limbs</i>	Ipsilateral ESPN	0.14	0.79 [-0.07 2.16]	0.53	0.32 [-0.62 1.39]	0.23	0.62 [-0.26 3.64]
	Ipsilateral OBL	0.21	0.66 [-0.06 2.37]	0.47	0.37 [-0.16 1.63]	0.26	0.59 [-0.03 1.69]
	Ipsilateral GMAX	0.25	0.61 [0.06 1.92]	0.58	0.28 [-0.31 1.25]	0.28	0.56 [0.09 1.30]
	Ipsilateral GMED	0.14	0.79 [-0.20 2.55]	0.19	0.70 [0.35 1.53]	0.60	0.27 [-0.48 0.80]
	Contralateral ESPN	0.23	0.63 [0.12 2.00]	0.69	0.20 [-0.32 1.28]	0.19	0.74 [-0.12 1.7]
	Contralateral OBL	0.96	0.03 [-0.84 0.88]	0.37	0.47 [-0.38 1.79]	0.13	0.81 [0.04 1.95]
	Contralateral GMAX	0.72	0.19 [-1.28 1.15]	0.72	0.19 [-0.84 1.30]	0.52	0.33 [-0.70 1.78]
	Contralateral GMED	0.53	0.33 [-0.54 1.53]	0.57	0.29 [-0.39 1.04]	0.67	0.22 [-0.23 1.24]

7.4.3 Step Descent

7.4.3.1 Trunk Rotational Angular Momentum (RAM)

In the sagittal plane, during the weight acceptance phase of step descent, the peak posterior trunk RAM was larger in the TTA group when leading with the intact limb than HC ($P < 0.01$, $g = 2.60 [2.03\ 4.52]$) (Figure 7.6).

In the frontal plane, during the weight acceptance phase, the peak trunk RAM away from the leading stance foot (negative) was larger in the TTA group when leading with the intact limb than HC ($P < 0.01$, $g = 1.03 [0.11\ 2.62]$) (Figure 7.6).

In the transverse plane, during the weight acceptance phase, the peak trunk RAM toward the leading limb (negative) was larger in the TTA group when leading with the intact limb than HC ($P = 0.01$; $g = 1.59 [0.65\ 4.45]$) (Figure 7.6).

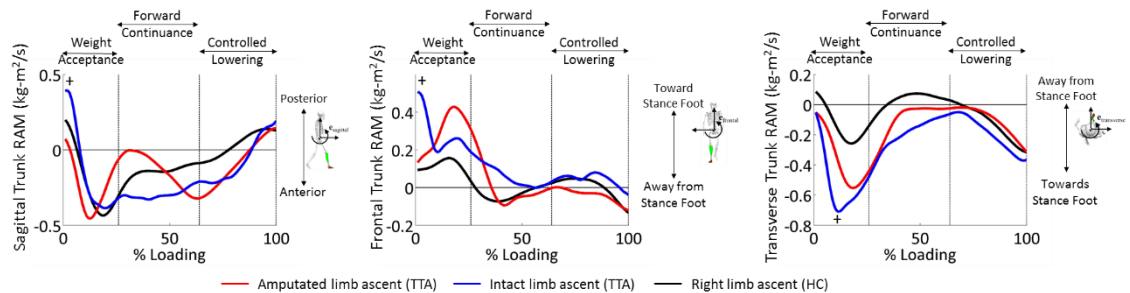


Figure 7.6. Ensemble averages of trunk rotational angular momentum (RAM) during the step descent. Statistically significant differences ($P < 0.017$) within and across groups are: amputated vs. intact limbs (▲), amputated limb vs. healthy controls (*), and intact limb vs. healthy controls (+).

7.4.3.2 Integrated EMG (iEMG)

When measured across the entire loading period, the iEMG from the OBL on the leading limb side was larger when leading with the intact limb than HC. During the forward continuance and controlled lowering phases, the iEMG from the ESPN on the trailing limb side was larger in the TTA group when leading with both the amputated or

intact limb than HC. During the controlled lowering phase, the iEMG from the OBL on the trailing limb side was larger in the TTA group when leading with the amputated limb than HC, and the iEMG from the GMAX on the trailing limb side was larger in the TTA group when leading with the intact limb than HC (Figure 7.7, Table 7.3).

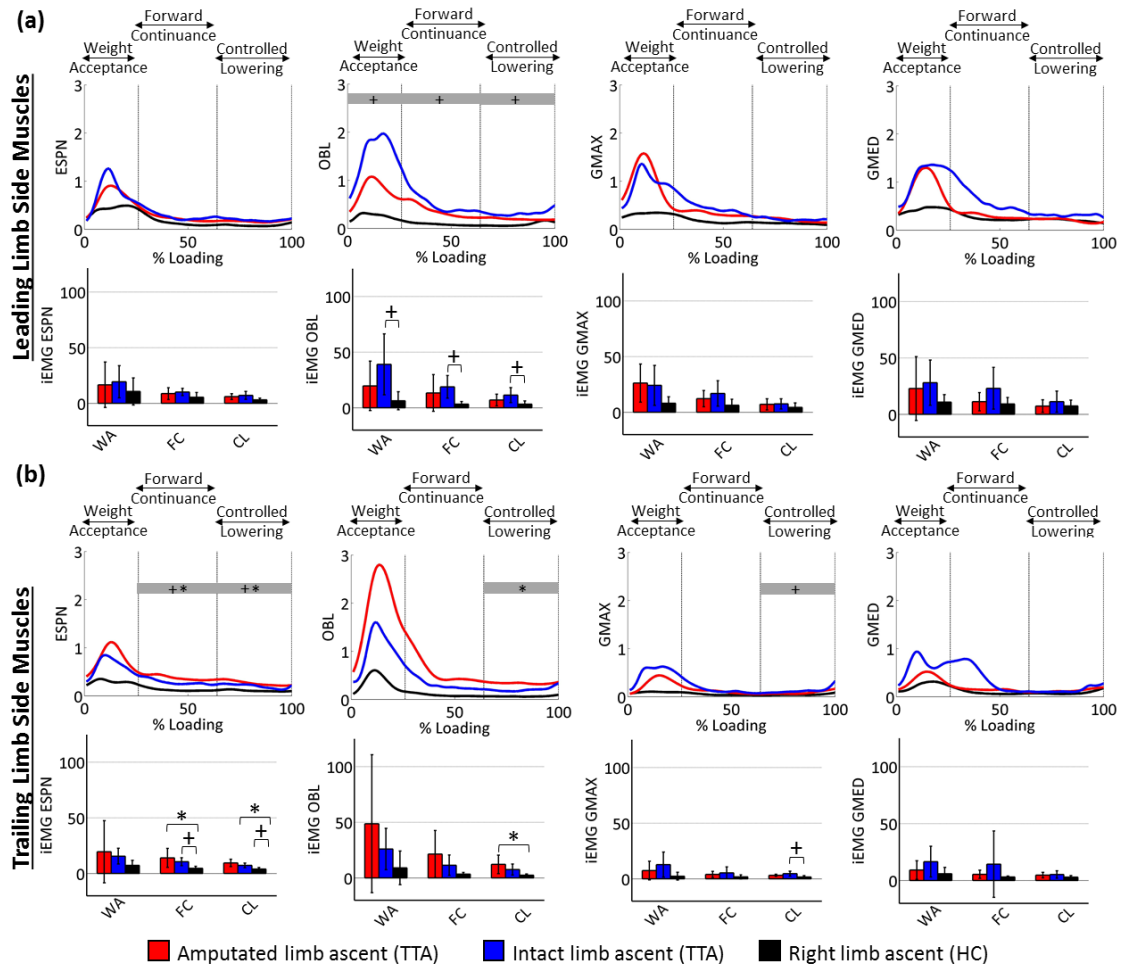


Figure 7.7. Ensemble averages of normalized linear envelope and integrated EMG (iEMG) of (a) leading limb side and (b) trailing limb side muscles during the phases of the step descent: weight acceptance (WA), forward continuance (FC), and controlled lowering (CL). See Section 7.2.2 for specific muscle acronym definitions. Statistically significant differences ($P < 0.017$) within and across groups are: amputated vs. intact limbs (\blacktriangle), amputated limb vs. healthy controls ($*$), and intact limb vs. healthy controls ($+$).

Table 7.3. Statistical comparisons across groups (intact limb vs. healthy control and amputated limb vs healthy control) and across limbs (amputated vs. intact limbs) during the functional phases of the step descent. See Section 7.2.2 for specific muscle acronym definitions. Statistically significant differences ($P < 0.017$) of pairwise comparisons are noted as: intact limb vs. healthy controls (+), amputated limb vs. healthy controls (*), and amputated vs. intact limbs (\blacktriangle).

	<i>Muscle</i>	<i>Weight Acceptance</i>		<i>Forward Continuance</i>		<i>Controlled Lowering</i>	
		<i>P-value</i>	<i>Hedge's g</i>	<i>P-value</i>	<i>Hedge's g</i>	<i>P-value</i>	<i>Hedge's g</i>
<i>Intact Limb vs. Healthy Control</i>	Ipsilateral ESPN	0.25	0.61 [0.04 3.05]	0.04	1.16 [0.36 3.53]	0.02	1.37 [0.93 2.58]
	Ipsilateral OBL	0.01 ⁺	1.52 [0.96 3.02]	<0.01 ⁺	1.98 [1.27 4.16]	0.01 ⁺	1.48 [0.73 3.64]
	Ipsilateral GMAX	0.04	1.13 [0.59 2.16]	0.05	1.11 [0.60 2.14]	0.20	0.68 [0.16 1.81]
	Ipsilateral GMED	0.05	1.08 [0.46 2.16]	0.08	0.95 [0.52 1.94]	0.37	0.47 [-0.35 1.26]
	Contralateral ESPN	0.02	1.29 [0.57 2.50]	<0.01 ⁺	1.95 [1.23 3.92]	<0.01 ⁺	1.82 [0.78 4.17]
	Contralateral OBL	0.09	0.94 [-0.15 2.82]	0.04	1.16 [1.23 3.92]	0.02	1.33 [0.74 3.16]
	Contralateral GMAX	0.04	1.16 [0.79 2.31]	0.13	0.83 [0.12 1.85]	0.01 ⁺	1.63 [0.96 3.16]
	Contralateral GMED	0.08	0.96 [0.36 2.20]	0.33	0.51 [-0.49 1.40]	0.12	0.84 [-0.27 2.87]
<i>Amputated Limb vs. Healthy Control</i>	Ipsilateral ESPN	0.52	0.34 [-0.70 1.12]	0.23	0.64 [0.03 1.83]	0.02	1.30 [0.38 3.41]
	Ipsilateral OBL	0.16	0.75 [0.55 2.77]	0.13	0.80 [0.64 3.13]	0.12	0.85 [-0.10 3.93]
	Ipsilateral GMAX	0.02	1.33 [0.76 3.43]	0.11	0.86 [0.30 2.38]	0.29	0.56 [0.01 1.70]
	Ipsilateral GMED	0.30	0.55 [-0.51 1.30]	0.61	0.26 [-0.31 1.07]	0.95	0.03 [-0.72 0.80]
	Contralateral ESPN	0.28	0.57 [0.23 1.90]	0.02*	1.42 [1.07 3.71]	<0.01*	2.05 [1.33 4.45]
	Contralateral OBL	0.13	0.82 [0.18 1.73]	0.04	1.13 [0.72 2.16]	0.01*	1.53 [0.99 3.12]
	Contralateral GMAX	0.17	0.72 [0.43 2.40]	0.10	0.88 [0.38 2.31]	0.02	1.26 [0.48 5.69]
	Contralateral GMED	0.40	0.44 [-0.39 1.40]	0.12	0.83 [0.04 1.92]	0.13	0.81 [-0.26 2.32]
<i>Amputated vs. Intact Limbs</i>	Ipsilateral ESPN	0.78	0.14 [-0.30 2.34]	0.54	0.32 [-0.15 1.78]	0.55	0.31 [-0.34 0.77]
	Ipsilateral OBL	0.17	0.73 [0.14 2.00]	0.47	0.38 [-0.43 2.51]	0.20	0.69 [-0.02 1.94]
	Ipsilateral GMAX	0.83	0.11 [-0.71 0.98]	0.38	0.46 [-0.30 1.33]	0.88	0.08 [-0.45 0.79]
	Ipsilateral GMED	0.70	0.20 [-0.47 1.92]	0.16	0.78 [0.46 1.51]	0.36	0.48 [-0.01 1.08]
	Contralateral ESPN	0.73	0.18 [-1.76 0.75]	0.35	0.50 [-0.23 1.17]	0.17	0.74 [-0.23 2.13]
	Contralateral OBL	0.99	0.47 [-0.65 1.18]	0.29	0.57 [-0.44 1.57]	0.23	0.63 [-0.22 1.85]
	Contralateral GMAX	0.33	0.51 [-0.13 1.53]	0.58	0.29 [-0.89 0.94]	0.13	0.85 [0.09 1.88]
	Contralateral GMED	0.25	0.62 [0.10 1.57]	0.45	0.40 [-1.23 1.01]	0.73	0.18 [-0.58 1.08]

7.5 Discussion

The purpose of this study was to identify muscle demand from core trunk/pelvis muscles associated with trunk movement compensations during step ambulation in people with unilateral transtibial amputation (TTA). Patients with TTA adopted asymmetric trunk movements during both step ascent and descent than healthy controls (HC) when quantified using rotational angular momentum (RAM), which corresponded with high bilateral core muscle demand. These results indicate that regardless of an asymmetric trunk movement compensation, the TTA group demonstrated high bilateral core muscle demand during step ambulation, which has potential implications related to development of LBP (Kulkarni et al., 2005; Morgenroth et al., 2010). Because people with TTA are often required to lead with either limb when navigating unexpected obstacles (Molina-Rueda et al., 2015), they would benefit from interventions that target trunk motor control, enable faster activation, and provide more postural support during high-demand tasks (Marigold et al., 2005).

7.5.1 *Step Ascent*

When leading with the intact limb, the TTA group adopted a ‘hip dominant strategy’, which may compensate for limited active push-off force from the trailing amputated limb. This strategy corresponded to larger GMAX muscle demand on the leading limb side than the HC group. This observation is similar, but more exaggerated, than the hip-extensor dominant strategy that typically occurs in level-ground amputee gait to assist with forward progression (Fey et al., 2010). Linked to the hip-extensor dominant strategy, the TTA group generated a larger amount of anterior trunk RAM than the HC group to

elevate the body COM during weight acceptance, which corresponded with the larger demand from the bilateral ESPN and OBL muscles.

The TTA group demonstrated greater muscle activity from the trunk lateral bend and rotator muscles than the HC group. This strategy adopted during the high-demand task was likely adopted to maintain balance throughout the loading period due to the larger generation of RAM generated by the TTA group in the frontal and transverse planes, which is consistent with a compensated Trendelenburg pattern (Gaffney et al., 2016). This strategy is a common characteristic of amputee gait that results in larger frontal plane trunk displacement, and aids in lateral balance by shifting the body COM closer to the base of support (Hof, 2008). In addition, this compensation may also be a result of ‘hip hiking’, where the person elevates the pelvis on the swing limb side to avoid contact between the swing limb and the step, which increase demand from the GMED on the leading limb side (Nadeau et al., 2003; Gottschall et al., 2012).

Because the TTA group adopted asymmetric trunk movement compensations during step ascent that did not correspond to asymmetric muscle demand, this may indicate limb-dependent motor control compensations that are dependent upon the leading limb. When ascending onto the intact limb, which is what is taught in movement retraining, the TTA group adopted different movement compensations than the HC group that corresponded with high muscle demand throughout the loading period. By contrast, when ascending onto the amputated limb, the TTA group did not adopt a different movement strategy early in the loading period than the HC group, yet had larger demand from the core muscles than the HC group. We attribute this to the novelty of the task presented

here because it is well-known that motor control strategies change with training (Thoroughman and Shadmehr, 1999; Lay et al., 2002).

7.5.2 *Step Descent*

When descending onto the intact limb, the TTA group had reduced ability to actively lower the body COM with the amputated limb (Schmalz et al., 2007). As a result, the TTA group ‘falls’ into the intact limb, which we identified with larger RAM generation during loading compared to the HC group. In preparation for the increased loading, the TTA group adopted a trunk stiffening compensation to prevent collapse and maintain balance by limiting perturbations to the body COM, which that resulted in large core muscle demand. This protective strategy may be used to prevent additional injury surrounding the spine (Solomonow et al., 2008).

In general, this cohort of people with TTA self-reported that step descent onto the intact limb was the hardest task to complete, which may have induced fear or anxiety while performing this task. This is similar to a previous investigation describing the anxiety of people descending steps with an ankle foot orthosis (Nahorniak et al., 1999). Fear of movement increases muscle demand (Tsao and Hodges, 2008; Massé-Alarie et al., 2016) and decreases functional performance (Miller et al., 2001). While lack of ankle control with a passive prosthesis limits controlled lowering, core strengthening may help improve self confidence in preparation for the increased loads that are associated with stepping onto the intact limb.

When stepping onto the amputated limb, the TTA group adopted a similar trunk stiffening by increasing the demand of core trunk/pelvis musculature, yet this did not

coincide with asymmetric movement patterns. To maintain balance when descending onto the amputated limb, the TTA group limited the perturbations of the body COM by limiting the amount of generation of trunk RAM, which increased bilateral core muscle demand. This strategy is consistent with other conservative strategies adopted by people with TTA to prevent falls when loading the amputated limb during high-demand tasks such as walking on uneven surfaces (Paysant et al., 2006), stairs (Ramstrand and Nilsson, 2009), or inclined surfaces (Vickers et al., 2008). Although people with TTA may have greater confidence performing the task by increasing muscle demand to account for balance deficits, increased and repetitive guarding strategies may become consequential over time through the development of LBP or other overuse injuries (Kumar, 2001; McGill, 2007).

7.5.3 Clinical Application

Based on our results, we recommend that movement retraining for people with TTA include step ambulation leading with either limb, which can likely be accomplished through targeted strengthening of core muscles (Kahle and Tevald, 2012). Without the function of the ankle and reduced lever arm of the knee extensors, movement compensations are necessary. However, with training and familiarity, the unnecessary core muscle demand could be reduced, and may provide more efficient postural control resulting in a reduced risk of falling when navigating an unexpected obstacle (Marigold et al., 2005; Molina-Rueda et al., 2015).

Establishing the association between muscle demand and movement compensations identified through RAM, which is based on angular velocity, has clinical implications as

wearable sensors (e.g. gyroscopes) become more feasible for clinical use (Watanabe et al., 2011). Future work should focus on identifying specific associations between movement compensations and muscle demand, and using them to improve rehabilitation for optimizing movement patterns following lower limb amputation.

7.5.4 Limitations

There are several limitations to consider with this investigation. First, there was a wide variability of step completion time among patients with TTA, which has been shown to be correlated with muscle demand (Larsen et al., 2008). Second, the TTA group was not screened for LBP at the time of testing; therefore, we cannot conclude if the movement compensations adopted were a habitual patterns or a result of LBP. However, we do not anticipate this to have a confounding effect on our results because no patient with TTA reported LBP at the time of testing. Finally, this investigation included a small homogenous sample size of patients with dysvascular TTA, which may limit the generalizability of these results to patients with unilateral TTA from other causes than neurovascular disease.

7.6 Conclusion

To our knowledge, these results are the first to demonstrate the high demand of core muscles in people with TTA needed to support trunk movement compensations during step ascent and descent. Regardless of the leading limb, the TTA group increased demand from the core muscles during both step ascent and descent. Asymmetric trunk movement compensations did not correspond to asymmetric core muscle demand, which has

potential clinical implications related to targeting research and clinical interventions to improve movement symmetry and prevent secondary pain conditions, such as LBP.

CHAPTER 8: CONCLUSIONS AND RECOMMENDATIONS

The Specific Aims presented in this dissertation used the foundation and applications of the separation of segmental angular momentum to describe trunk and pelvis movement compensations, and the underlying mechanisms behind the movements, in patients with unilateral dysvascular transtibial amputation (TTA) during over-ground walking and step ambulation. Movement compensations adopted by patients with TTA described in the context of segmental angular momentum were consistent with compensations identified in both instrumented and observational analyses, indicating that this approach may be applicable in both research and clinical settings.

8.1 Conclusions of Specific Aims

The foundations of the separation of angular momentum presented in Chapter 4 were used to describe segmental movement coordination and effort during over-ground walking in healthy adults. Total segmental angular momentum can be separated into translational (TAM) and rotational angular momentum (RAM). By referencing TAM to the stance foot, a coherent interpretation of forward progression throughout the stance period was described. Segmental RAM was used to identify specific segmental movement patterns that were adopted to achieve forward progression based on the variations in segmental angular velocity. Using Euler's Laws in rotational form, the translational and rotational segment moments were obtained using the time derivative of TAM and RAM, respectively. The translational moment is dependent upon all external

forces acting on a segment; and therefore, provides insight regarding changes in gravitational forces, intersegmental joint forces, and linear forces applied through muscle force actuators. The rotational moment is foundational in inverse dynamic analyses and is dependent upon all external moments (i.e., joint moments) applied to the segment, as well as moments due to external forces (i.e., intersegmental joint forces). Because these forces and moments are representative of external biomechanical loads, the generation and arresting of segmental angular momentum likely indicates the demand placed on the musculoskeletal system.

In Chapter 5, these foundations were used to describe trunk and pelvis movement compensations in patients with unilateral TTA during over-ground walking by assessing patterns of generating and arresting segmental TAM and RAM. In the sagittal and transverse planes, patients with TTA increased the generation of trunk and pelvis RAM throughout the stance period compared to the reference groups. Because the gait speed was fixed (1 m/s), this is interpreted as an alteration in the position vector between the segment and stance foot. This corresponds to a wider step width, which is a common characteristic of amputee gait that is adopted to improve stability by widening the base of support. In addition, patients with TTA demonstrated larger patterns of generating and arresting trunk and pelvis RAM in all three planes compared to the reference groups. Of particular interest, patients with TTA generated larger mediolateral trunk RAM during loading and unloading of the amputated limb, which is consistent with a compensated Trendelenburg posture. In order to arrest the increased segmental RAM to maintain

balance throughout the gait cycle, the eccentric muscle demand is likely increased which may lead to consequential long-term effects such as low back pain.

Chapter 6 used the trunk segmental moments, calculated through Euler Laws, to evaluate the trunk compensations and the associated kinetic effort needed for patients with unilateral TTA to perform step ascent and descent tasks. During the step ascent, patients with TTA generated larger sagittal trunk translational moments when leading with the amputated limb compared to the intact limb. This arises from increased anterior intersegmental joint forces at the low back due to a hip-extensor dominant strategy that is commonly adopted by patients with TTA to elevate the body COM without active push off forces from the amputated ankle. In the frontal and transverse planes, the TTA group generated larger rotational moments when leading with either limb compared to healthy controls, which are likely required to maintain balance. During the step descent, patients with TTA demonstrated higher translational moments when stepping onto the intact limb compared to the amputated limb. This increase arises from increased ground reaction forces that propagate up the kinetic chain, and are associated with ‘falling’ onto the intact limb without active control of lowering the body COM with the amputated limb. Through increased trunk translational and rotational segment moments, patients with TTA require high kinetic effort during stepping tasks compared to healthy controls. The compensations that resulted in the increased and asymmetric trunk segmental moments may increase the risk of the development of low back pain in patients with TTA through increased and asymmetric loading patterns.

Changes from core muscle demand that were required to support trunk movement compensations during step ascent and descent tasks adopted by patients with unilateral TTA were examined in Chapter 7. Demand from the core muscles was quantified using the normalized integrated EMG signal measured from the bilateral erector spinae (ESPN), external oblique (OBL), gluteus maximus (GMAX), and gluteus medius (GMED) throughout each functional phase of the step ascent and descent tasks; and movement compensations were quantified using trunk RAM. Patients with TTA adopted asymmetric trunk movements during both step ascent and descent than healthy controls (HC) when quantified using rotational angular momentum (RAM), which corresponded with high bilateral core muscle demand. During both step ascent and descent, patients with TTA had larger demand from core muscles than HC when stepping onto both limbs, yet only adopted different movement compensations during early loading when leading with the intact limb. Early after amputation during the acute care phase of movement retraining, patients with amputation are taught to compensate for lack of knee and ankle control on the amputated limb by ascending steps with the intact limb and descend steps with the amputated limb. Based on our results, patients with TTA increased core muscle demand during both habitual and novel tasks, yet adopted asymmetric movement compensations, which indicates different motor control strategies used to support the movements. Because patients with TTA are likely required to ambulate steps on both limbs for community ambulation, this population may benefit from rehabilitation that retrains motor control compensations and targeted core muscle strengthening to provide faster reflexes and more postural support, that can be used to better support trunk

movements during high-demand tasks. Additionally, these results provide preliminary support that asymmetric movement compensations correspond to asymmetric core muscle demand, which has potential valuable clinical implications towards the measurement of RAM using wearable sensors in a clinical setting.

8.2 Summary of Limitations

There are several limitations to this project that should be considered. First, both components of segmental angular momentum (TAM and RAM) are calculated using kinematics measured from reflective markers placed over anatomic landmarks. Motion of reflective markers relative to landmarks are prone to measurement error, primarily attributed to marker placement error and skin motion artifact (Della Croce et al., 1999; Chiari et al., 2005; Gao & Zheng, 2008), which will propagate through to TAM and RAM calculations. Without performing an analysis of uncertainty, the effect of measurement variability on TAM on RAM remains unknown. Second, the sample of patients with amputation used was small and only included TTA resulting from dysvascular pathologies, which may limit generalizability to individuals with other types of TTA (e.g. traumatic, oncologic, congenital). Third, neither the AMP and DM groups were screened for LBP at the time of testing; therefore, we cannot determine if compensatory movement patterns adopted by each group were habitual movement pattern or the result of LBP. Fourth, analyses that use the TAM component of segmental angular momentum are only applicable when the foot is in contact with the ground during over-ground dynamic movements. During ballistic motion, the body COM may be a more appropriate point of reference, but this has not been explored.

8.3 Recommendations for Future Work

In order to further establish the use of separation of angular momentum in a clinical setting, there are several areas of this work that should be expanded. First, the associations between segmental RAM and muscle demand need to be further established. Second, the movement compensations identified using segmental TAM and RAM need to be associated to levels of physical function to widen the clinical impact of this approach. Third, this approach should be applied to other populations with movement pathologies that differ from those adopted by patients with TTA (e.g. cerebral palsy, total joint replacement, etc.) to establish its robustness across different patient populations. Finally, segmental angular momentum should be measured using wearable sensors and validated against a traditional instrumented passive motion capture system to establish the feasibility of using this approach in a clinical setting.

The association between movement compensations identified using segmental RAM and muscle demand should be further explored. Due to the limitations of the muscle set used and small sample size in Specific Aim 4, the direct association between segmental RAM and muscle demand was not able to be made. The central nervous system is a highly redundant system because the number of muscles vastly exceeds the number of degrees of freedom; and therefore movements can be achieved by an infinite number of muscle recruitment strategies (Groote et al., 2014). As a result, there are likely numerous motor control strategies adopted by patients with TTA that resulted in the movement compensations identified in the generation and arresting of trunk RAM, which likely attributed to demand from muscles not included in these results. If established, the

associations between movement compensations and the corresponding muscle demand could help improve the efficacy of rehabilitation by providing quantitative evidence to a clinician regarding the underlying motor control compensations pertaining to a pathologic movement.

The relation between segmental movement compensations identified using TAM and RAM and levels of physical functional performance should be explored. Based on initial qualitative inspection of individual of clinical measures of functional performance of patients with TTA (Appendix B) and individual curves of segmental TAM and RAM (Appendix C), patients with lower physical function tended to demonstrate larger peaks of TAM and RAM during walking in the frontal and transverse planes. However, these patients were not included in the analyses used in Chapters 6 and 7 because they were unable to perform the stepping tasks bilaterally, therefore it is unknown if movement compensations during high-demand tasks are associated with function. By associating movement compensations to clinical measures of functional performance, clinicians will have foundational evidence to base practice guidelines regarding which specific movement patterns adopted by a patient may benefit or degrade their ability to maintain high levels of functional ability.

The use of segmental angular momentum to describe movement compensations in other populations with movement pathologies should be explored. For example, patients with total joint replacement, stroke, spinal cord injury, or cerebral palsy all adopt movement compensations specific to their pathology that are likely different than the compensations presented in this dissertation. Therefore, it is unknown if compensations

identified using segmental RAM are robust to identify common compensations identified via instrumented and observational analyses in this populations. If established, the use of segmental angular momentum to describe movement compensations across different movement pathologies will have significant clinical impact by improving a clinician's ability to diagnose correct subject specific compensations that may become consequential by the development of secondary pain conditions.

Finally, clinical implementation of measuring angular momentum outside of a traditional instrumented laboratory using wearable sensors should be explored. Of the two components of segmental angular momentum, segmental RAM is currently the easiest of the two components to calculate outside of a motion capture laboratory because it does not require segment localization. Because RAM depends on the angular velocity of the segment, the segmental movement patterns identified using RAM can be implemented in a clinic through small gyroscopes made possible by microelectromechanical systems (MEMS). MEMS sensors have been used to measure kinematics and spatiotemporal parameters during walking (Sinclair et al., 2013; Patterson et al., 2014; López-Nava et al., 2015) and used for movement retraining (Wall et al., 2009). As wearable sensors become more widely used in a clinical setting (typically for activity monitoring) (Butte et al., 2012; Redfield et al., 2013; Fulk et al., 2014), future work should focus on how the implementation of wearable sensors can be used to measure or infer biomechanically useful information outside of a traditional motion capture laboratory using segmental momentum.

In summary, the techniques presented in this dissertation provide foundational evidence to describe segmental movement patterns in a way that is relevant to both researchers and clinicians that can be used to improve the overall efficacy of movement retraining. Identification of segmental movement compensations using angular momentum has clinical importance by potentially improving three foundational aspects of clinical gait analysis: 1) providing a common language between clinicians and researchers focusing on gait analysis, 2) providing an interpretable way to quantify movement patterns, and 3) identifying targets for intervention.

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APPENDIX A: Subject Characteristics

Tables A.1, A.2, and A.3 provide individual subject characteristics at the time of testing for the three groups (patients with unilateral dysvascular transtibial amputation (TTA), patients with diabetes mellitus (DM), and healthy controls (HC)) included for all experimental studies. HC subject 43 was not included in Chapter 4 due to time of testing conflicting with time of manuscript preparation. DM subjects 18 and 33 were not included in chapter 5 due to no sex matched subjects in the TTA group. TTA subjects 2, 24, and 28 were excluded from Chapters 6 and 7 because they were unable to perform bilateral step ascent and descent tasks. Four subjects were recruited and included in the experimental collection, but were not included in any analyses for the following reasons: subject 4 (HC) had a right total knee replacement, subject 12 (TTA) injured intact ankle prior to experimental session and was not willing to participate at a later testing date, subject 15 (TTA) had peripheral artery disease and a powered prosthesis, and subject 34 (TTA) exceeded the BMI inclusion criteria.

Table A.1. Patient anthropometrics for the unilateral dysvascular transtibial amputation (TTA) group.

Subject ID	Age (years)	Sex	Height (m)	Mass (kg)	Time since Amputation (Months)	Residual Limb Length (cm)	Amputated Leg	Hemoglobin A1C
2	63	M	1.8	107.0	23	15.4	R	-
5	58	M	1.8	98.6	20	15.5	R	-
11	54	M	1.8	88.6	10	16.0	R	8.1
17	52	M	1.9	100.5	17	17.0	R	-
22	60	M	1.8	84.6	21	10.5	L	6.2
24	55	M	1.7	99.8	21	16.0	L	8.0
26	52	M	1.8	99.6	13	11.5	R	8.1
28	56	M	1.7	122.0	13	16.0	L	11.2
30	54	M	1.9	109.1	12	18.0	L	5.9
36	64	M	1.6	65.9	24	12.5	L	7.7
Mean	56.8	-	1.8	97.6	17.4	14.8	-	7.9
Standard Deviation	4.3	-	0.1	15.2	5.1	2.5	-	1.7

Table A.2. Patient anthropometrics for the diabetes mellitus (DM) group.

Subject ID	Age (years)	Sex	Height (m)	Mass (kg)	HbA1c
8	67	M	1.8	127.9	-
10	59	M	1.9	115.7	8.3
18	58	F	1.6	73.9	6.9
19	53	M	1.9	115.7	6.4
21	75	M	1.8	77.1	-
27	71	M	1.8	99.8	-
29	71	M	1.8	98.9	6.9
31	57	M	1.8	63.5	-
32	71	M	1.8	86.2	12.0
33	50	F	1.5	69.9	8.0
35	52	M	1.7	77.1	6.5
40	60	M	1.9	123.3	8.3
41	65	M	1.8	98.9	9.0
44	60	M	1.8	120.7	6.5
Mean	62.1	-	1.8	96.3	7.9
Standard Deviation	8.0	-	0.1	21.9	1.7

Table A.3. Patient anthropometrics for the healthy control (HC) group.

Subject ID	Age (years)	Sex	Height (m)	Mass (kg)
1	60	F	1.6	63.5
3	57	M	1.9	93.9
6	61	M	1.6	74.4
7	70	M	1.8	79.4
9	63	M	1.8	94.4
14	64	M	1.8	77.1
16	73	M	1.8	89.4
20	59	F	1.6	60.8
23	79	F	1.5	59.4
25	54	M	1.8	75.8
37	65	M	1.8	91.6
38	60	M	1.8	103.0
39	65	M	1.8	79.4
42	50	M	1.7	83.0
43	55	M	1.8	74.8
Mean	61.6	-	1.7	79.4
Standard Deviation	7.8	-	0.1	12.6

APPENDIX B: Clinical Measures of Functional Performance

To quantify physical function of all participants, common clinical tasks to measure functional performance of each subject were used to assess patient outcomes based on time to complete a task and distance traveled. The functional tests used included the Two-Minute Walk Test (2MW), the Timed Up and Go Test (TUG), the Stair Climb Test (SCT), and the self-selected walking speeds. The 2MW, which measures the distance walked over two minutes, is a reliable and valid measure to assess function in patients with lower-limb amputation (Simpson et al., 1982; Datta et al., 1996; Brooks et al., 2001; Resnik & Borgia, 2011). The TUG, which measures the time to stand from a chair, walk three meters, turn, walk back to the chair, and sit down without physical assistance, is a reliable test for assessing the mobility, strength, balance, and agility of a patient (Shumway-Cook & Brauer, 2000; Ng & Hui-Chan, 2005; Bennell et al., 2011). The SCT is a test that assesses the ability to ascend and descend a flight of stairs, and assess the lower-extremity strength, power, and balance of a patient (Powers et al., 1997; Schmalz et al., 2007; Bean et al., 2007; Bennell et al., 2011). Self-selected walking speed is a reliable measure that has shown to correlate with physical function (Steffen et al., 2002). Each participant performed each test three times. Times and distances from the second and third trials were averaged together. Tables B.1, B.2, and B.3 provide the individual functional measures from each task for the three groups.

Table B.1. Functional performance task results of each patient with unilateral dysvascular transtibial amputation.

Subject ID	TUG Time (s)	SCT Ascent Time (s)	SCT Total Time (s)	2MW Distance (m)	Self-Selected Walking Speed (m/s)
2	11.8	18.0	39.1	122.5	0.8
5	11.0	9.7	18.7	160.0	1.2
11	9.4	8.9	20.0	164.6	1.1
17	9.0	9.1	18.6	164.3	1.1
22	10.4	8.6	18.4	134.1	0.9
24	12.7	16.5	37.5	115.2	0.8
26	10.4	11.0	23.8	167.0	1.1
28	14.5	14.3	29.3	121.3	0.8
30	11.6	12.8	24.5	131.1	0.9
36	12.0	17.2	37.2	132.6	0.9
Mean	11.3	12.6	26.7	141.3	1.0
Standard Deviation	1.6	3.7	8.5	20.4	0.1

Table B.2. Functional performance task results of each patient with diabetes mellitus.

Subject ID	TUG Time (s)	SCT Ascent Time (s)	SCT Total Time (s)	2MW Distance (m)	Self-Selected Walking Speed (m/s)
8	12.7	9.7	22.5	135.9	1.0
10	11.7	8.8	20.4	153.0	1.0
18	5.2	3.6	7.4	243.8	1.6
19	8.2	5.0	9.1	225.6	1.1
21	16.8	13.5	33.5	97.5	0.7
27	8.9	5.1	12.0	198.1	1.2
29	7.5	6.3	12.2	223.1	1.1
31	11.0	7.6	16.7	142.6	0.8
32	12.4	9.0	20.1	128.0	0.9
33	6.3	4.0	8.3	224.0	1.2
35	5.5	4.0	7.5	237.7	0.9
40	8.3	7.6	18.0	172.2	1.1
41	10.0	6.6	13.2	176.8	1.2
44	10.1	7.7	13.4	161.2	1.1
Mean	9.6	7.0	15.3	180.0	1.1
Standard Deviation	3.2	2.7	7.2	46.1	0.2

Table B.3. Functional performance task results of healthy control subjects.

Subject ID	TUG Time (s)	SCT Ascent Time (s)	SCT Total Time (s)	2MW Distance (m)	Self-Selected Walking Speed (m/s)
1	7.4	6.4	11.7	204.2	1.4
3	6.7	4.9	8.7	191.4	1.3
4	7.0	4.2	7.8	271.9	1.1
6	5.4	5.1	9.2	219.5	1.4
7	6.6	4.6	9.3	237.7	1.4
9	6.8	4.3	7.6	274.3	1.7
14	9.1	5.7	11.0	213.4	1.0
16	7.5	4.8	9.3	198.1	1.1
20	18.9	5.3	10.4	178.9	1.1
23	6.3	3.9	7.4	250.9	1.6
25	4.4	3.6	6.3	280.4	1.4
37	6.2	3.9	8.3	192.0	1.2
38	8.1	6.5	13.6	189.0	1.5
39	5.7	2.9	4.9	219.5	1.4
42	5.4	3.4	5.6	242.3	1.4
43	8.5	4.7	9.1	234.4	1.8
Mean	7.5	4.6	8.8	224.9	1.4
Standard Deviation	3.3	1.0	2.3	32.6	0.2

APPENDIX C: Individual Curves of all Dependent Variables in Patients with Transtibial Amputation

Figures C.1 – C.10 provide the individual ensemble average curves of all dependent variables in Chapters 5-7 of each patient with unilateral dysvascular transtibial amputation (TTA). In addition, each figure contains the TTA group average (1 SD) indicated by the shaded grey region

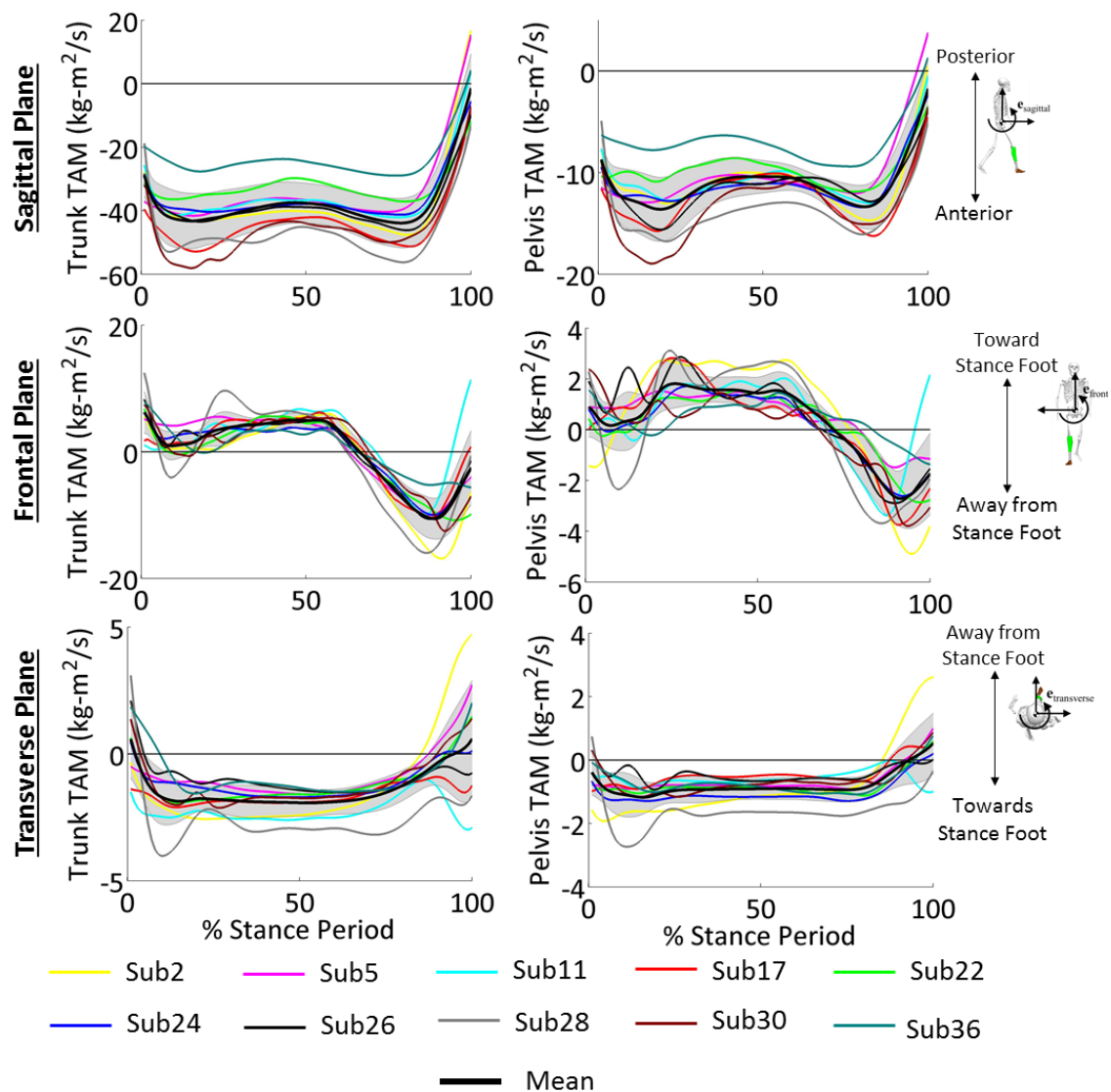


Figure C.1. Individual ensemble averages of the translational angular momentum (TAM) of the trunk and pelvis with respect to the stance foot in the sagittal, frontal, and transverse planes in the TTA group. Grey shaded region indicates the group average (1 SD).

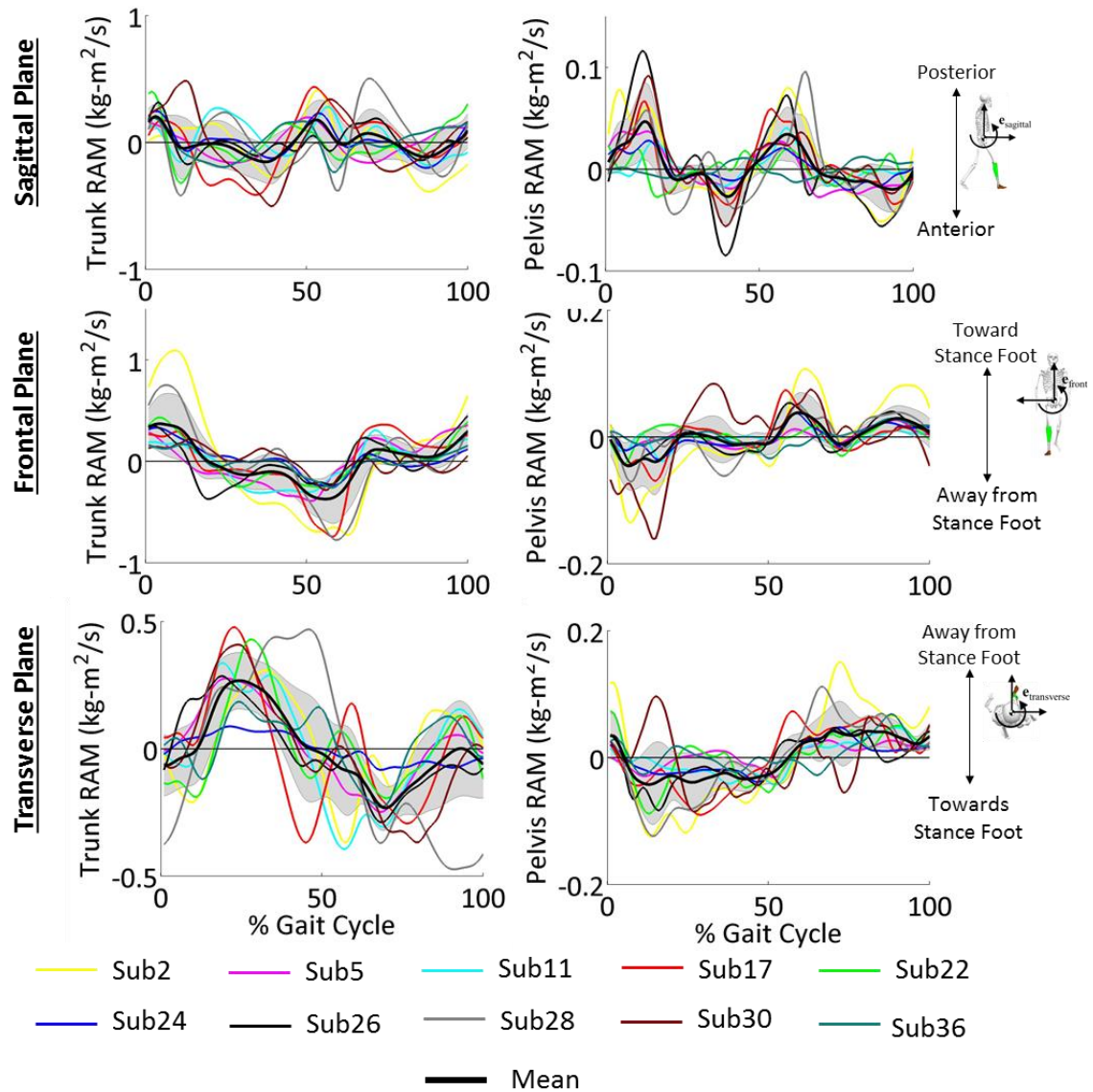


Figure C.2. Individual ensemble averages of the rotational angular momentum (RAM) of the trunk and pelvis with respect to the stance foot in the sagittal, frontal, and transverse planes in the TTA group. Grey shaded region indicates the group average (1 SD).

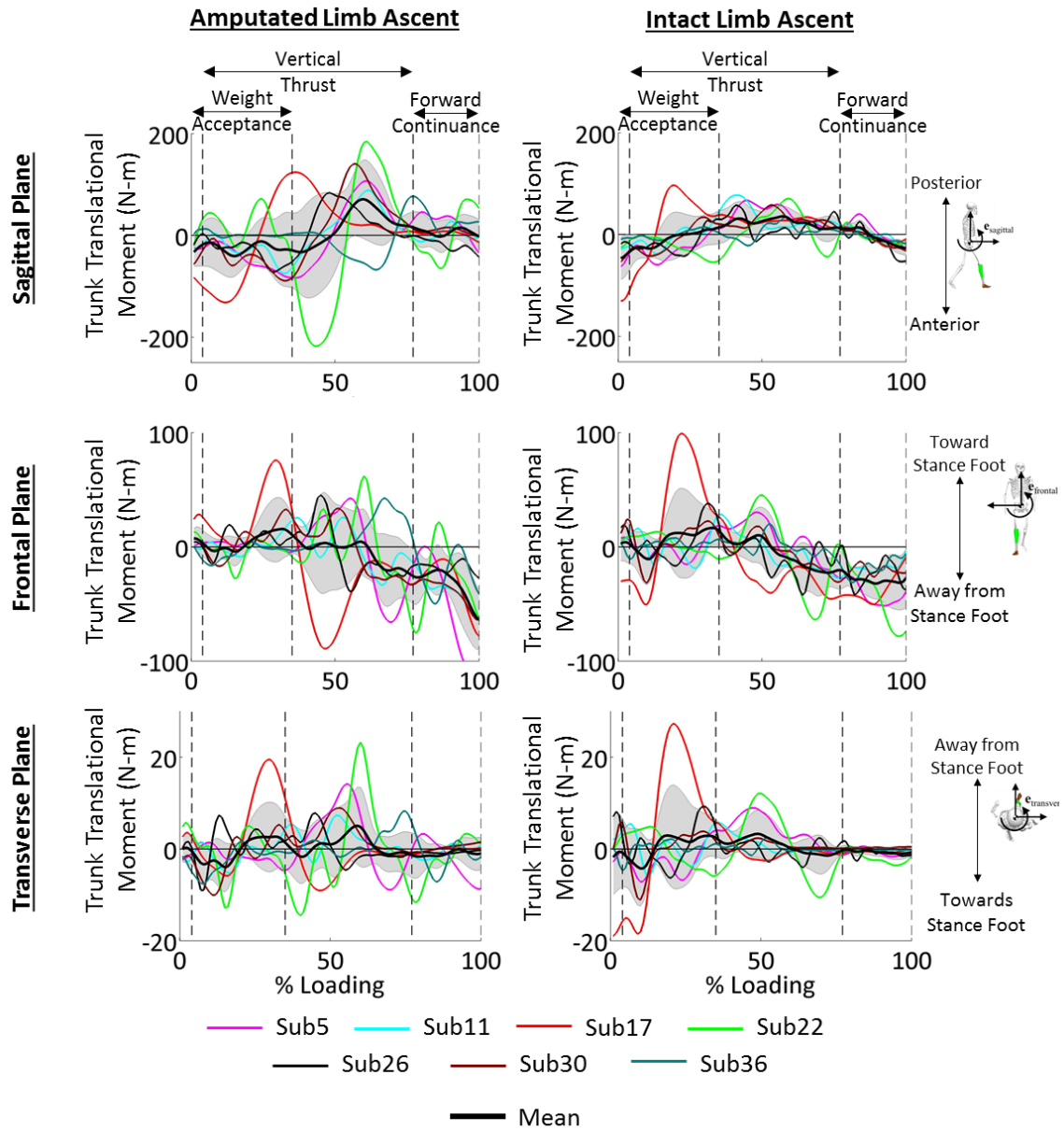


Figure C.3. Individual ensemble averages of the trunk translational moment during the step ascent with respect to the stance foot in the sagittal, frontal, and transverse planes in the TTA group. Grey shaded region indicates the group average (1 SD).

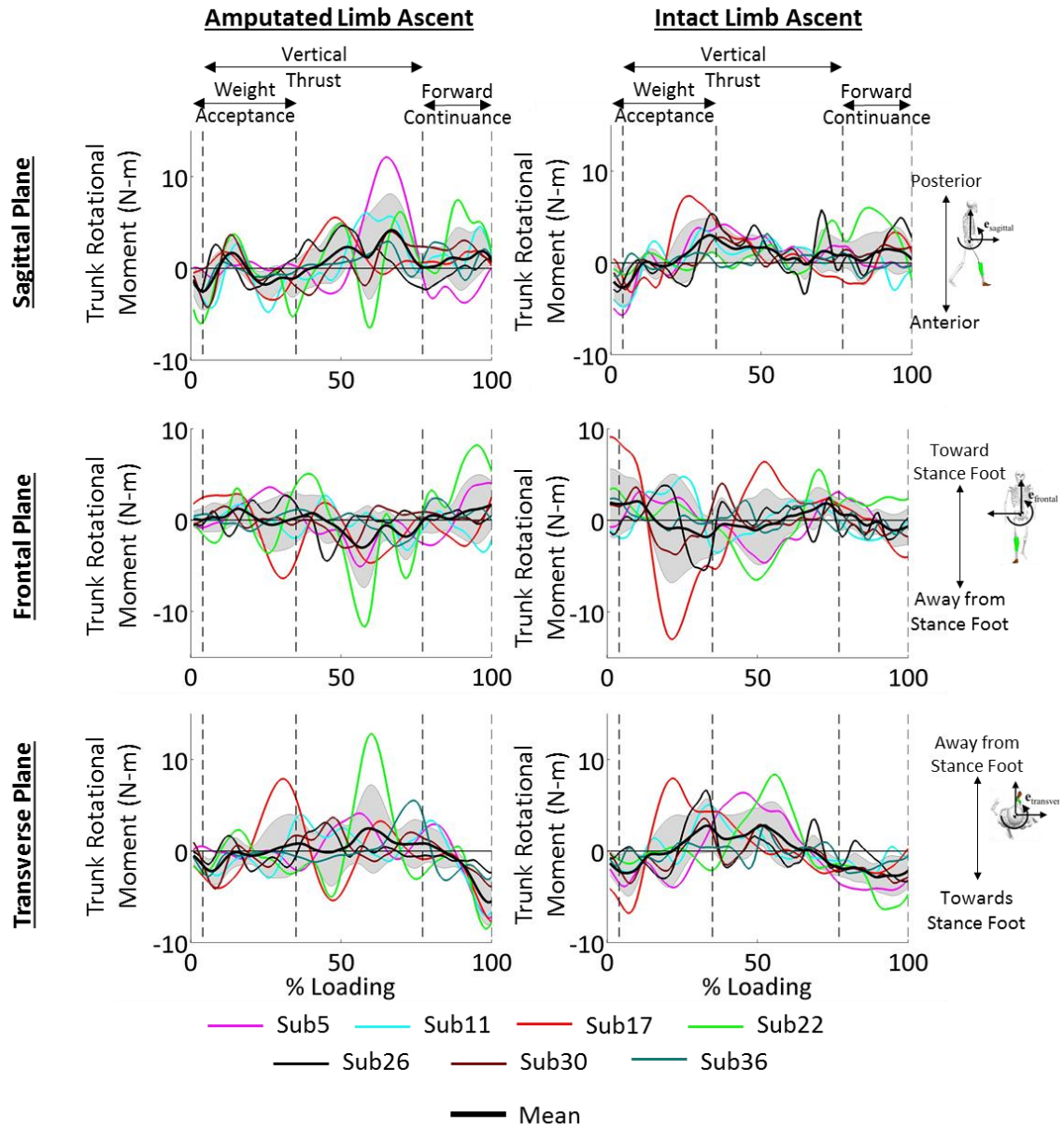


Figure C.4. Individual ensemble averages of the trunk rotational moment during the step ascent in the sagittal, frontal, and transverse planes in the TTA group. Grey shaded region indicates the group average (1 SD).

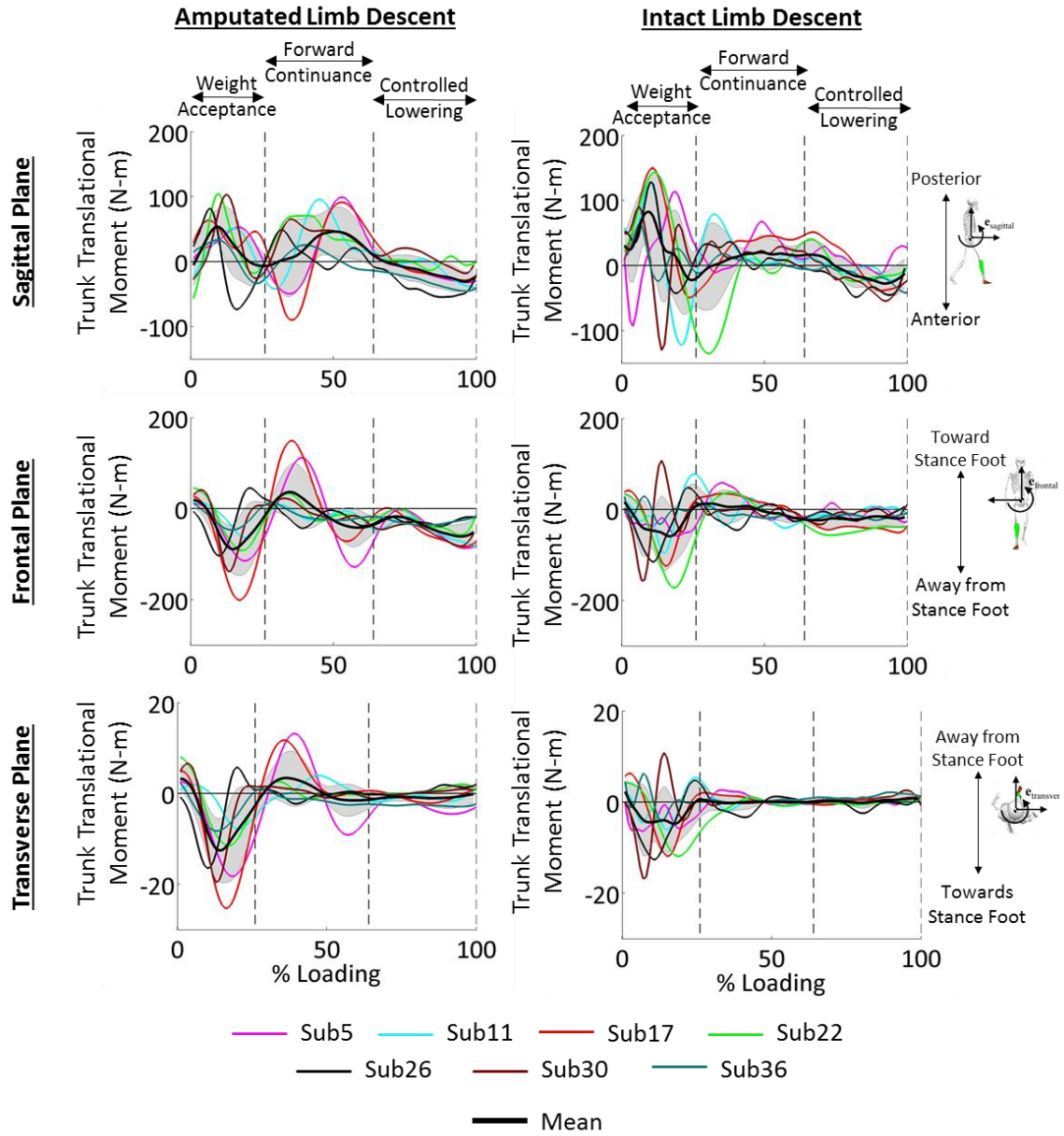


Figure C.5. Individual ensemble averages of the trunk translational moment during the step *descent* with respect to the stance foot in the sagittal, frontal, and transverse planes in the TTA group. Grey shaded region indicates the group average (1 SD).

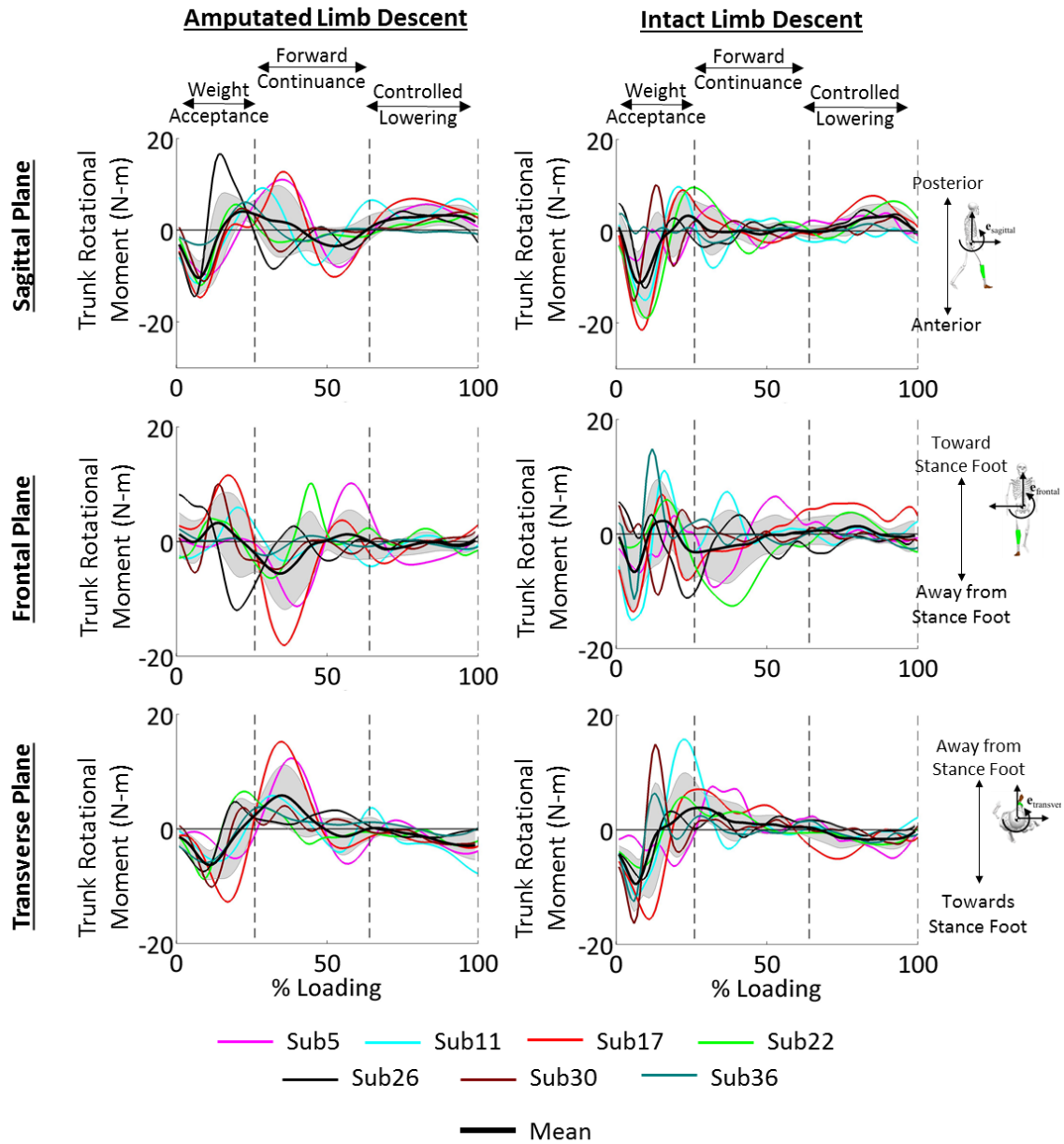


Figure C.6. Individual ensemble averages of the trunk rotational moment during the step descent in the sagittal, frontal, and transverse planes in the TTA group. Grey shaded region indicates the group average (1 SD).

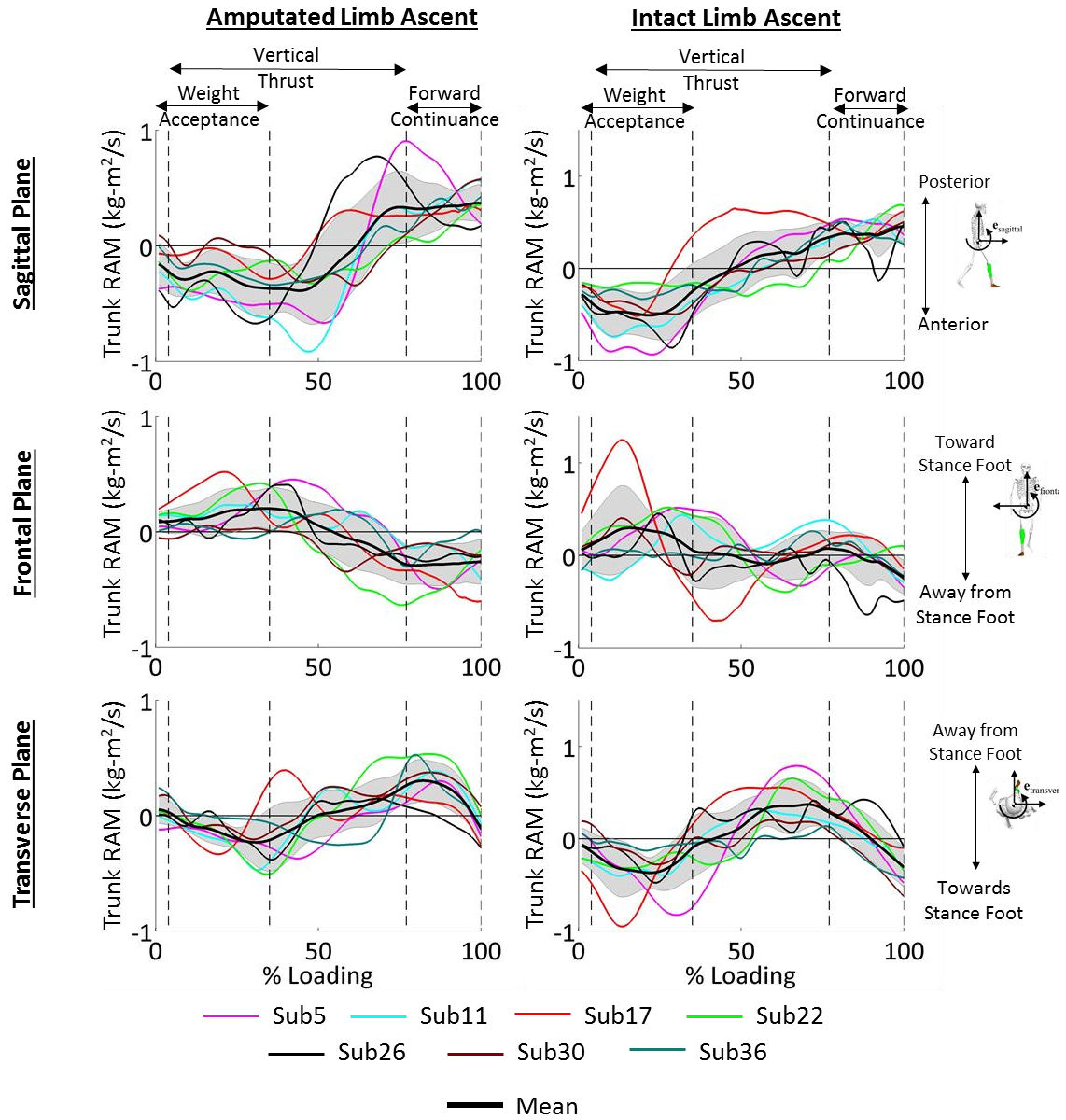


Figure C.7. Individual ensemble averages of the trunk rotational angular momentum (RAM) during the step ascent. Grey shaded region indicates the group average (1 SD).

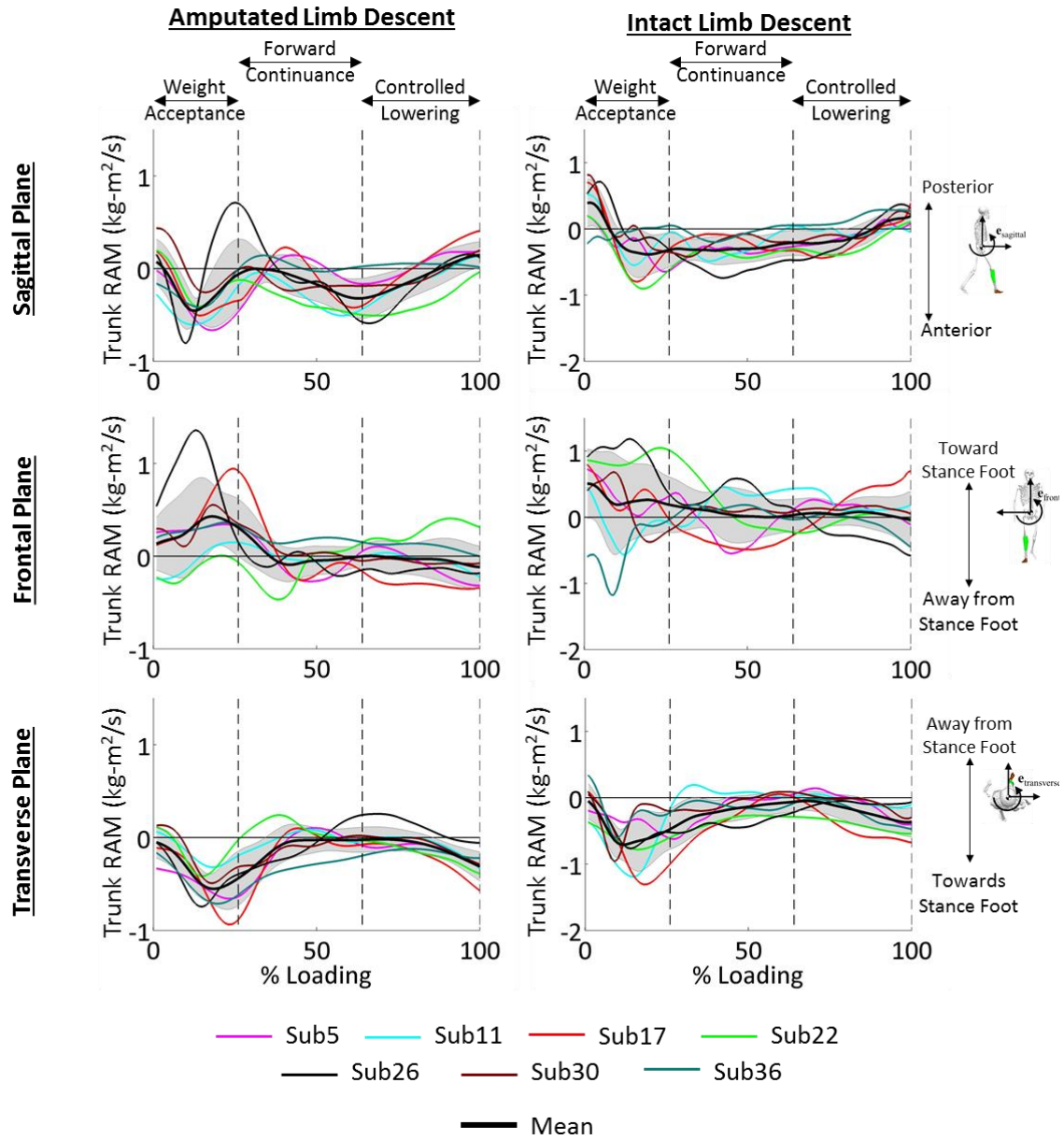


Figure C.8. Individual ensemble averages of the trunk rotational angular momentum (RAM) during the step *descent*. Grey shaded region indicates the group average (1 SD).

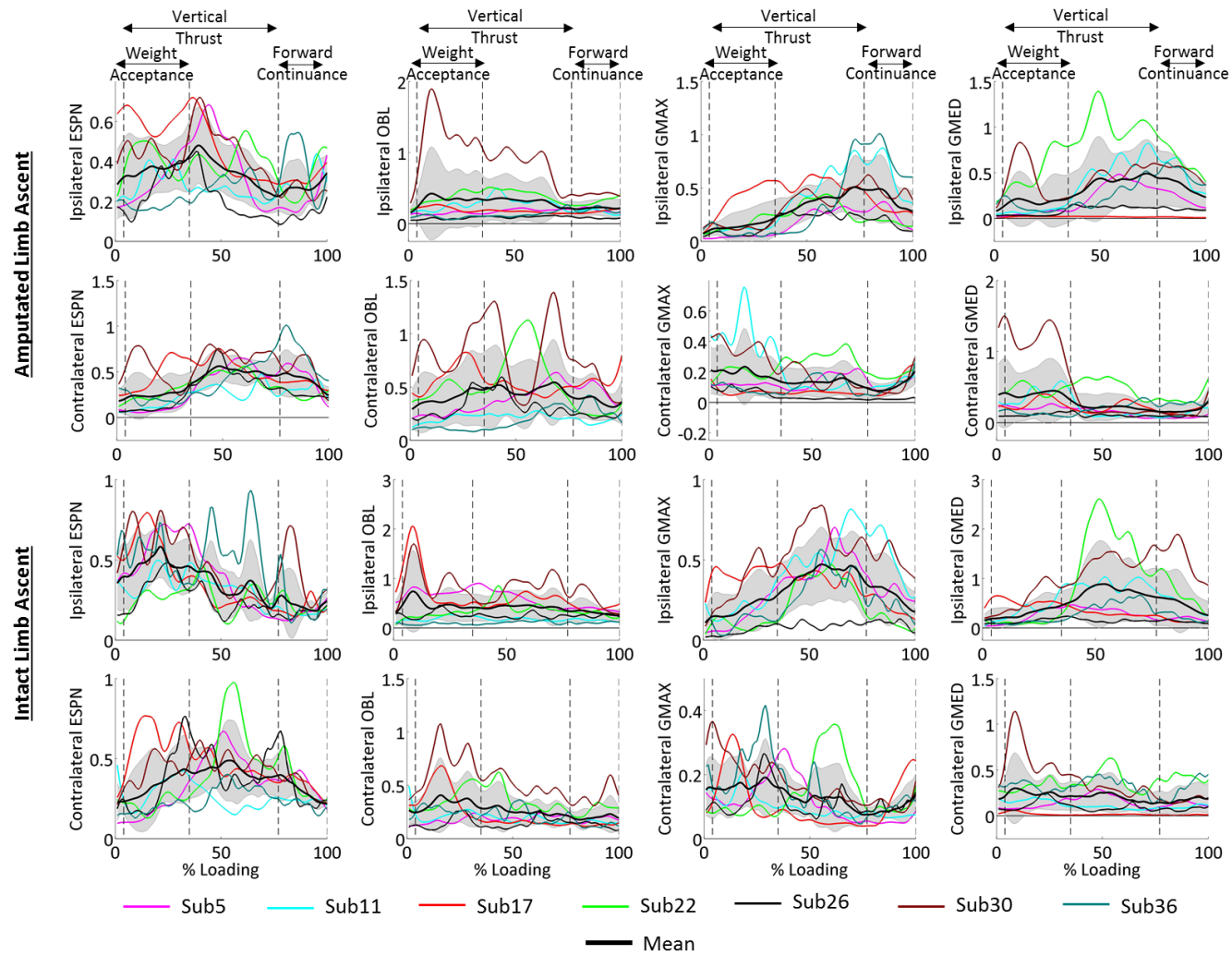


Figure C.9. Individual ensemble averages of the normalized EMG of the ipsilateral and contralateral core muscles during the step ascent. Grey shaded region indicates the group average (1 SD).

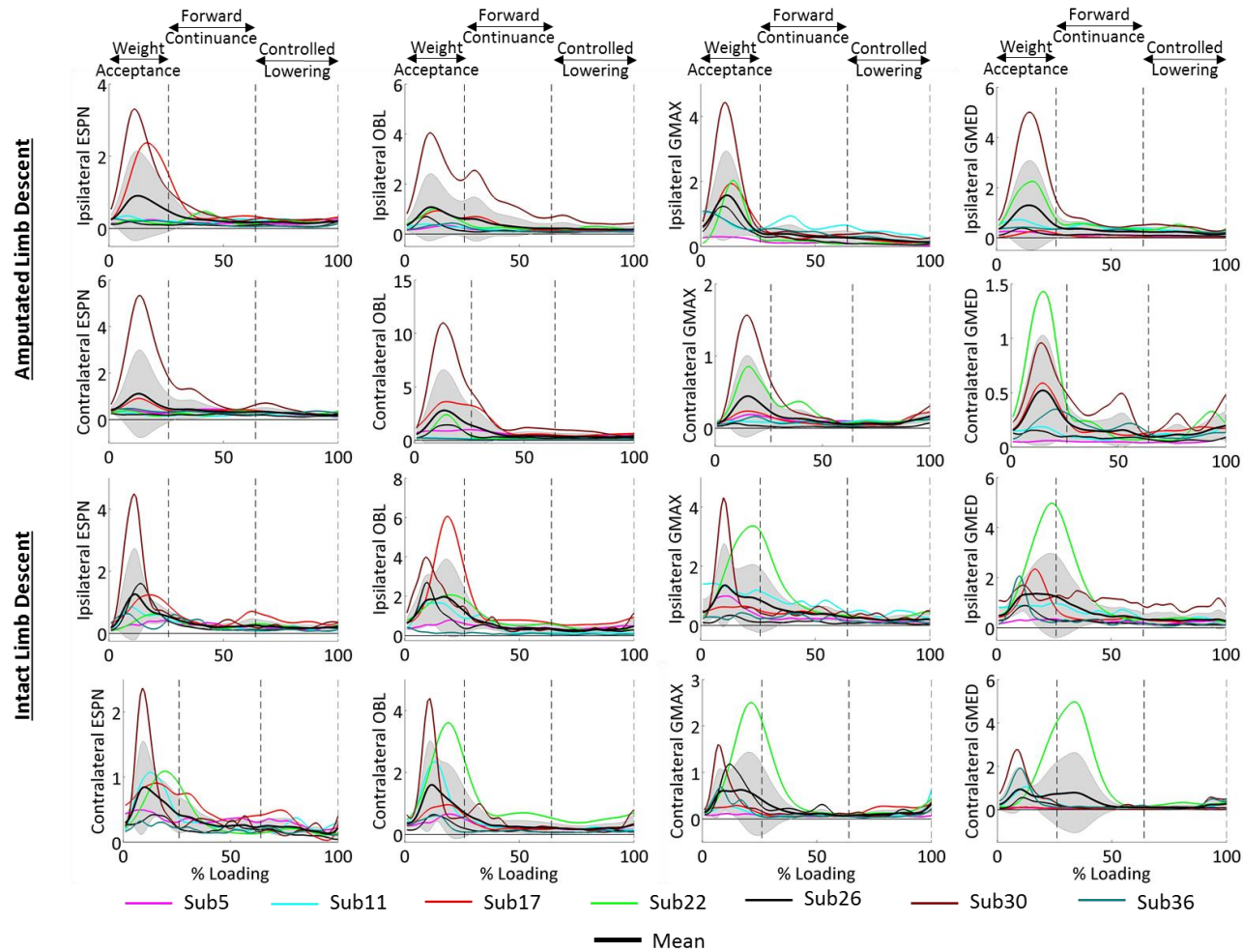


Figure C.10. Individual ensemble averages of the normalized EMG of the ipsilateral and contralateral core muscles during the step descent. Grey shaded region indicates the group average (1 SD).