12-1-2015

Biomechanical Assessment of Ertl and Burgess Transtibial Amputation Techniques

Abbie Elthea Ferris
University of Northern Colorado

UNIVERSITY OF NORTHERN COLORADO

Greeley, Colorado

The Graduate School

BIOMECHANICAL ASSESSMENT OF ERTL AND
BURGESS TRANSTIBIAL AMPUTATION
TECHNIQUES

A Dissertation Submitted in Partial Fulfillment
of the Requirements for the Degree of
Doctor of Philosophy

Abbie E. Ferris

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

December 2015

This Dissertation by: Abbie E. Ferris

Entitled: *Biomechanical Assessment of Ertl and Burgess Transtibial Amputation Techniques*

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

Accepted by the Doctoral Committee

____
Jeremy D. Smith Ph.D., Research Advisor

____
Gary D. Heise Ph.D., Committee Member

____
David S. Hydock Ph.D., Committee Member

____
Thomas M. Dunn Ph.D., Faculty Representative

Date of Dissertation Defense ____

Accepted by the Graduate School

____

Linda L. Black, Ed.D.
Associate Provost and Dean
Graduate School and International Admissions

ABSTRACT

Ferris, Abbie., E., *Biomechanical Assessment of Ertl and Burgess Transtibial Amputation Techniques*. Published Doctor of Philosophy dissertation, University of Northern Colorado, 2015.

In this dissertation, a model was developed to predict the inertial properties of the shank and foot of persons with TTA and functional differences between Ertl and Burgess amputees during curb negotiation and the sit-to-stand tasks were evaluated. The developed inertial model was able to predict the shank and foot segment mass, COM location, and MOI more accurately than using the intact limb inertial properties. Used as inputs into inverse dynamics equations, the general model predictions produced joint moments which were also similar to the subject-specific measures. Thus, this model is a better predictor than the current method of using the intact limb inertial measures for the amputated limb. The second and third studies showed differences between the Ertl and Burgess amputated limbs in functional ability. During curb negotiation the Ertl amputated limb produced net limb work (sum of ankle, knee, and hip work) similar to that of the intact limbs of both groups on the curb step. This net limb work was produced by the hip early in stance phase as a compensatory mechanism to help propel the body forward. During the sit-to-stand task, the Ertl group was able to perform the task more quickly than the Burgess group. The faster performance time was due in part to larger ground reaction forces in the Ertl amputated limb compared to the Burgess amputated limb. This suggested the Ertl limb was able to bear higher loads overall during this task. While no other differences were found between the amputated limbs, the Ertl intact limb showed

unexpected differences. Where the Burgess limbs and Ertl amputated limb adopted a hip strategy to complete the task, the Ertl intact limb adopted a knee strategy. This knee strategy is more similar to the way non-amputees complete the task. Both study 2 and 3 highlighted functional advantages of the Ertl procedure over the Burgess procedure for these tasks and is, to our knowledge, the first study of its kind. Based on these outcomes, it appears that the Ertl procedure does lead to better functional performance during prosthesis use, and further consideration should be given to using this procedure at the time of amputation. Future work needs to continue to focus on functional performance in both groups and begin to contrast the outcomes with post-operative risks following the amputation to better inform patients and clinicians about potential advantages of either technique.

ACKNOWLEDGEMENTS

I would like to thank my parents for their unwavering support and love through this entire process. I think, at times, they had more confidence in me that I did in myself. Their strength carried me through. Standing right behind my parents were my friends who always offered an ear for late night phone calls, kind words of support, and many, many laughs. Thank you to my committee members who offered valuable insights to help strengthen this dissertation and foster my growth as a scientist. Most notably is my mentor, Dr. Jeremy Smith. I could not have completed this without his patience, kindness, and motivation to keep pushing forward. Lastly, my sister kept me grounded by showing me the importance of living each day to the fullest and to never give up.

TABLE OF CONTENTS

CHAPTER I. General Tntroduction 1

CHAPTER II. Study 1: Development of Body Segment Parameter Regression Equations for Persons With Transtibial Amputation 9

- Introduction 9
- Methods 12
  - Phase I - Model development 12
  - Phase II - Model validation 19
- Results 19
- Discussion 24

CHAPTER III. Study 2: Biomechanical Analysis of Curb Ascent In Persons with Burgess and ERTL Transtibial Amputations 26

- Introduction 26
- Methods 29
  - Participants 29
  - Data collection 30
  - Data analysis 32
  - Statistical analysis 33
- Results 33
  - GROUND step results 34
  - CURB step results 36
- Discussion 38

CHAPTER IV. Study 3: Biomechanical Analysis of Sit to Stand in Persons with Burgess and ERTL Transtibial Amputations 44

- Introduction 44
- Methods 46
  - Participants 46
  - Data collection 47
  - Data analysis 49
  - Statistical analysis 50
- Results 51
  - Kinematics 51
  - Ground reaction forces 52
  - Angular impulse 53
  - Joint work 54
  - Joint powers 55

Discussion 56
CHAPTER V. General Discussion and Conclusions 61
Conclusion 68
Future directions 69
REFERENCES 72
APPENDIX A Review of Literature 95
Introduction 96
Gait 98
Spatiotemporal 99
Kinematics 101
Kinetics 102
Muscle activity 106
Prosthetic feet 108
Metabolic cost 110
Curb Negotiation 111
Sit to Stand 117
Sit to stand in non-amputees 117
Sit to stand in amputees 119
Lower Extremity Inertial Measurements 120
Estimating body segment parameters 120
Estimating inertial properties of the prosthesis 123
Amputation Techniques 124
Surgical techniques 125
Bone 128
Venous return. 134
APPENDIX B Effect Sizes 137
APPENDIX C Institutional Review Board Documents 139
Informed Consent Document 140
Approval Letter 143

LIST OF FIGURES

2.1 Oscillation rack configuration for measuring the period of oscillation of the prosthetic limb……………………………………………………………...………14

2.2 Experimental set for the reaction board test to calculate the prosthetic center of mass location…………...………………………………...…………….……..15

2.3 Example calculation for estimating prosthetic foot and shank inertial properties for a female TTA…………………..……………………………………………..18

2.4. Mean hip, knee, and ankle joint moments of 9 TTAs using INTACT, SPECIFIC, and GENERAL inertial models……….…………....……….……….23

3.1 Illustration of the curb design.……………….......……………………………...…..31

3.2 Net joint work for the ankle, knee, and hip for the GROUND step………...…..…..35

3.3 Net limb work for the entire limb for the GROUND step. ……………...……….…35

3.4 Net joint work for the ankle, knee, and hip for the CURB step……………………..37

3.5 Net joint work for entire limb for the curb step.………...………………………......37

3.6 Ankle power for the GROUND step. …...……………………………………..…39

3.7 Joint powers for the ankle (A), knee (B) and hip (C) for the curb step.……….....…41

4.1 Mean sagittal hip and knee joint angles during the STAND task.…………………..…52

4.2 Sagittal plane joint powers for the hip and knee joints during the STAND task..…..56

LIST OF TABLES

2.1 Developed GENERAL model estimates of inertial properties for prosthetic shank and foot based on SPECIFIC measures …………...……………….…...……....16

2.2 Means and standard deviations of inertial properties calculated using the SPECIFIC, GENERAL and INTACT models……………..21

2.3 Peak joint moments for the ankle, knee, and hip for all three models ………......….22

3.1 Positive and negative joint work on the GROUND step………………...…….……34

3.2 Positive and negative joint work on the CURB step……………………...……....36

4.1 Peak ground reaction forces as a percentage of body weight (BW) and symmetry index between the amputated and intact limbs. …..…….…………..…..53

4.2 Angular momentum of the ankle, knee, and hip during the stand task……..…...…..53

4.3 Total work for the knee and hip joints……………………………………….……...54

4.4 Peak Power for the knee and hip joints………………………………….....…….55

CHAPTER I

GENERAL INTRODUCTION

There are over 20 million people in the US living with disabilities which limit ambulation (Bureau, 2012). An estimated 2 million of these Americans live with limb loss resulting from dysvascular disease, trauma, or cancer (Ziegler-Graham, MacKenzie, Ephraim, Travison, & Brookmeyer, 2008). Roughly 25% of these individuals have undergone a transtibial amputation. Lifetime costs associated with lower limb amputation are over a half a million dollars including prosthetic costs (Dillingham, Pezzin, & MacKenzie, 2002; MacKenzie et al., 2007). In addition to these increased healthcare costs, these individuals must also learn to adapt to numerous challenges associated with limb loss.

The ability to walk unassisted is one of the defining cornerstones of mobility independence (Killey & Watt, 2006). Although not always apparent with the unaided eye, there are asymmetries between the intact and amputated limb of transtibial amputees (TTA). After amputation, a prosthetic foot/ankle is attached to the socket to enable the amputee to ambulate. Although these devices allow for ambulation, they do not fully replicate the physiological or mechanical structure they replace. Walking velocities of people with unilateral, TTA are significantly slower than those for non-amputees of similar age (Boonstra, Fidler, & Eisma, 1993; Isakov, Burger, Krajnik, Gregoric, & Marincek, 1997; Nolan et al., 2003; Powers, Rao, & Perry, 1998) which many limit independence. Asymmetries in step length, double limb support time, and stance and swing times between the intact limb and amputated limb have been consistently reported throughout the literature (Isakov et al., 1997; Mattes, Martin, & Royer, 2000; Royer & Wasilewski, 2006; Sadeghi, Allard, & Duhaime, 2001). Compared to the intact ankle, the prosthetic ankle has a reduced range of motion and only produces 50% of the intact ankle power during the push-off phase of walking (Bateni & Olney, 2002; Sadeghi, Allard, et al., 2001; Silverman et al., 2008; Ventura, Klute, & Neptune, 2011; Zmitrewicz, Neptune, & Sasaki, 2007). In order to compensate for this lack of ankle power, the hip contributes more to forward motion whereas the knee supports the body and maintains stability. Energy consumption is typically 20% to 30% higher in unilateral TTAs when compared to controls walking at similar speeds (Gailey et al., 1994; Molen, 1973; Torburn, Powers, Guiterrez, & Perry, 1995).

Compared to controls, the intact limb produces larger vertical ground reaction forces (GRF) during walking at comparable speeds (Engsberg, Lee, Tedford, & Harder, 1993; Nolan et al., 2003). Compared to the amputated limb, the intact limb produces significantly higher vertical GRF magnitudes during walking (Engsberg et al., 1993; Isakov, Mizrahi, Susak, & Onna, 1992; Nolan et al., 2003; Sanderson & Martin, 1997). Anterior-posterior GRFs are also significantly different between the limbs. The peak propulsive forces are reported to be significantly smaller in the amputated limb compared to the intact limb (Silverman et al., 2008). Silverman et al. (2008) found that as walking speed increases, propulsive impulse of the amputated limb does not increase with speed, and the propulsive impulse is significantly less than that produced by the intact limb or a limb of non-amputees. However, Silverman et al. (2008) found no significant difference in the braking impulse as walking speed increased. This suggests the intact limb is mainly responsible for maintaining forward momentum of the body and maintaining walking velocity and the amputated lib does not impede forward motion.

The greater vertical GRFs of the intact limb may contribute to the increased prevalence of osteoarthritis (OA) in the intact limb (Burke, Roman, & Wright, 1978). It is suggested that roughly 65% of unilateral amputees have some level of OA (Lemaire & Fisher, 1994; Melzer, Yekutiel, & Sukenik, 2001). During quiet standing the largest force applied to the body is simply body weight. However, during a dynamic task such as walking, these forces generally increase with speed compared to quiet standing values. Walking is a repetitive dynamic task thus subjecting the body to repetitive high loads compared to quiet standing. As previous literature suggests, these loads are unevenly borne between the intact and prosthetic limb and may contribute to osteoarthritis and lower back pain. Investigation into these types of dynamic situations may identify differences between the Ertl and Burgess techniques. If the Ertl decreases these loads, and increases symmetry between the limbs, the prevalence of OA in the contralateral limb may decrease.

Relative to age-matched, able-bodied individuals, persons with TTA have an increased risk of falling and fear of falling (W. Miller, Speechley, & Deathe, 2001; Vanicek, Strike, McNaughton, & Polman, 2009). Further 60% of persons with TTA report that falls affect their daily activities, work, and confidence (Kulkarni, Toole, Hirons, Wright, & Morris, 1996). While gait is important to investigate, more demanding functional tasks of daily living are also important to consider. In particular, sitting, standing, and curb negotiation are of particular interest for several reasons as they are encountered on a frequent basis, may lead to trips and falls, and the force demands are greater compared to simple walking tasks. While these are important tasks, they have been studied less often than other tasks.

Persons with TTA report that curb negotiation is more demanding than negotiating stairs even though they are encountered with the same frequency (Larsson, Johannesson, Andersson, & Atroshi, 2009; Shumway-Cook et al., 2002). However, the underlying biomechanical mechanisms contributing to a more challenging task are unclear. During curb ascent in non-amputees, the lead limb has a longer step length than during descent (Loverro, Mueske, & Hamel, 2013). During negotiation of obstacles at varied heights, analysis of GRFs showed that vertical impulse increased as a function of obstacle height (Patla & Rietdyk, 1993). During stair ascent and decent, the intact limb of persons with TTA also experiences higher vertical GRFs than the amputated limb and non-amputees (Schmalz, Blumentritt, & Marx, 2007).

Sit-to-stand is an essential activity of daily living. It is estimated that people with TTA sit-to-stand roughly 50 times per day (Bussmann, Grootscholten, & Stam, 2004; Bussmann, Schrauwen, & Stam, 2008). Researchers have found during sit-to-stand, patients transfer weight towards the unaffected leg (Agrawal, Gailey, Gaunaurd, Gailey, & O'Toole, 2011; Ozyurek, Demirbuken, & Angin, 2013). Agrawal, Gailey, O'Toole, Gaunaurd, and Dowell (2009) found that patients with TTA produced 27% more vertical GRF with the intact limb during a sit-to-stand movement compared with the prosthetic side. Non-amputee controls, however, exhibited less than 10% asymmetry in vertical GRF during the same movement. Numerous factors may contribute to these differences in symmetry between the limbs. One such factor is the type of amputation technique used to remove the limb.

The two most common TTA techniques used by surgeons to amputate a limb are the modified Burgess and osteomyoplastic amputation (Ertl) techniques (Assal, Blanck, & Smith, 2005; Commuri, Day, Dionne, & Ertl, 2010; R. Dederich, 1983; Dionne, Ertl, & Day, 2009; Ertl, Ertl, Ertl, & Stokosa, 2013). The modified Burgess technique is more frequently used than the Ertl (Dionne et al., 2009). However, this amputation technique often leads to difficulties after amputation such as pain, swelling, instability, and significant residual limb atrophy. Although less common, the Ertl has been suggested to lead to improved functional outcomes following amputation. Using a "bone bridge", the Ertl technique connects the tibia and fibula and seals the medullary canal and sutures the anterior and posterior musculatures together. This technique commonly results in a healthier residual limb, reduced incidence of bone spurs, increased vascularity, and reduced incidence of skin ulcers (Rolf Dederich, 1963; Dudek, DeHaan, & Marks, 2003; Dudek, Marks, Marshall, & Chardon, 2005; Potter, Burns, Lacap, Granville, & Gajewski, 2007).

It has also been suggested that the Ertl technique may enhance "end-bearing" capability of the residual limb compared to the Burgess (Mongon et al., 2013). This improved "end-bearing" may reduce the asymmetrical loading patterns compared to the Burgess, thus potentially reducing the risk of developing OA, low back pain, or other comorbidities.

Given the lack of data related to functional outcomes following Ertl amputations, determining whether the Ertl amputation technique has a functional advantage and is able to reduce loading asymmetries over the more common Burgess technique was needed. These asymmetries were measured using GRFs and powers. Functional tasks beyond walking where asymmetric loading patterns are likely exacerbated were investigated and included activities of daily living such as sitting and curb negotiation. To accurately measure outcome variables such as joint moments and powers during these tasks, it is important to use body segment parameters in inverse dynamics analysis that accurately reflect the amputee limb morphology.

The body segment parameters of the amputated limb and prosthesis are significantly different than the intact limb. Compared to the intact limb, the mass of the prosthetic side is consistently 30-40% less, the center of mass location is 25-35% closer to the knee joint, and the moment of inertia is 50-60% less about a transverse axis through the knee joint (Lin-Chan, Nielsen, Yack, Hsu, & Shurr, 2003b; Mattes et al., 2000; J. D. Smith, Ferris, Heise, Hinrichs, & Martin, 2014). Currently the only method to obtain the inertial measurements of the prosthesis is through the use of oscillation and reaction board testing.

J. D. Smith et al. (2014) developed an oscillation rack to approximate the inertial properties of the amputated limb and found that the mass is significantly lower in the amputated limb. Using these values, joint moments and powers were calculated during walking. J. D. Smith et al. (2014) found that these differences in inertia did not result in significant differences in kinetics during the stance phase of walking, but resulted in significantly different kinetic profiles during swing, where GRFs are not present. Thus, during swing the inertial properties of the prosthetic side contribute largely to the estimated joint kinetics and should be estimated as accurately as possible for any analyses involving the swing phase. The oscillation method of calculating subject specific body segment parameters is a very accurate method. This was assessed by testing known geometrical solids with uniform density with the oscillation rack and reaction board testing. Compared to mathematical models of those solids, the direct measures were within 5% -12% of the measured location of the moment of inertia and the center of mass (J. D. Smith et al., 2014). However, the availability of the equipment described in the study is limited. Therefore, development of regression equations for this population would eliminate the need for specialized equipment and reduce data collection times significantly.

This dissertation consisted of three studies. In the first study, regression equations were created to predict the body segment parameters of persons with TTA. These effects of these parameters were compared through a sensitivity analysis to traditional methods and direct measurement of the body segment parameters. In the second and third studies, functional tasks were biomechanically evaluated in persons with TTA resulting from both a traditional and Ertl amputations. In the second study, amputee subjects (Ertl and Burgess) were compared while they ascended a curb. During stair ambulation, previous research has found the GRFs of the intact limb increase compared to the amputated and control limbs. The third study was similar to study two, except sit-to-stand in these groups was evaluated. It has been shown that amputees shift their weight to the non-amputated limb during this task, it is important to identify differences in amputation techniques compared to control subjects.

Study One Hypothesis – Inertia Properties

H01 It was hypothesized that the regression equations we developed for persons with TTA would accurately predict the body segment parameters of the residual limb. That is, it was expected that these regression equations would results in similar joint kinetics estimates compared to using direct measures of the prosthesis inertial properties.

Study Two Hypothesis – Curb Negotiation

H02 It was hypothesized that those with an Ertl amputation would be able to take advantage of the greater end-load bearing capability of the amputated limb, which would be evidenced by greater joint power magnitudes at the ankle, knee, and hip in the amputated limb. Thus, it was expected that a greater kinetic symmetry between the amputated side and intact side would occur in Ertl amputees during curb ascent.

Study Three Hypothesis – Sit-to-Stand

H03 It was hypothesized that those with an Ertl amputation would be able to take advantage of the greater end-load bearing capability of the amputated limb, which would be evidenced by greater vertical GRFs and peak joint powers at the knee and hip in the amputated limb. Thus, it was expected that a greater kinetic symmetry between the amputated side and intact side would occur in Ertl amputees during sit-to-stand.

CHAPTER II

STUDY 1: DEVELOPMENT OF BODY SEGMENT PARAMETER REGRESSION EQUATIONS FOR PERSONS WITH TRANSTIBIAL AMPUTATION

Introduction

Inverse dynamics analysis requires three primary inputs from experimental measures: 1) motion data, 2) ground reaction forces, and 3) body segment parameters or segment inertial properties. Motion data and GRFs are reliably captured with motion capture systems and force plates, respectively. However, body segment parameters can be determined based on a variety of tools from the literature and these parameters are dependent on the model the researcher chooses. Inertial properties of a body segment include mass, center of mass location, and moment of inertia. Researchers have developed equations to calculate the percent mass of each body segment, location of the center of mass as a percentage of segment length, and the location of the moment of inertia relative to the axis of rotation.

Many investigators have relied on regression equations based on cadaveric data to estimate body segment parameters (Chandler, 1975; Clauser, McConville, & Young, 1969; Dempster, 1955; Hinrichs, 1985, 1990). Newer methods such as dual x-ray absorptiometry scans (Durkin & Dowling, 2003; Durkin, Dowling, & Andrews, 2002), gamma radiation (de Leva, 1996; Zatsiorsky, 1983), MRI (Cheng, Chen, Chen, Chen, & Chen, 2000; Martin, Mungiole, Marzke, & Longhill, 1989; Mungiole & Martin, 1990;

Pearsall, Reid, & Ross, 1994), kinematic models (Drillis, Contini, & Bluestein, 1964; Herbert Hatze, 1975), and geometric models based on geometric principles (Hanavan, 1964; H. Hatze, 1980; Jensen, 1978) have been developed.

However, because of the nature of measurements of cadaveric specimens, there are several limitations including: an older population, pooling of body fluids, tissue loss, segmentation error, and measurement error. Although accurate, gamma radiation scanning has not been extensively used due to the health risks associated with radiation exposure. Many of the newer techniques are preferred over the cadaveric studies since they are non-invasive, can be performed on living subjects, and measured on an individual basis. Given the expense of some of these tools (e.g., MRI) and limited availability, regression equations are still widely used. Few regression equations exist for children (Ganley & Powers, 2005) and women (de Leva, 1996).

The variability in estimates of segment inertial properties has generally been accepted to have little influence on the joint moments during the stance phase of walking. This is primarily due to larger influences of GRFs and center of pressure locations used within the equations of motion (Challis, 1996; Challis & Kerwin, 1996). Thus, research questions focused on stance phase portion of the movement are not generally considered to depend on the specific inertial model used. However, recently, J. D. Smith et al. (2014) illustrated in a group of unilateral, transtibial amputees that research questions focused on periods when the limb is not in contact with ground are influenced by the inertial model chosen. This is important because in lower extremity amputees inertial properties of the prosthesis are needed as inputs into the equations of motion. Investigators often estimate the inertial properties of amputees based on estimates for intact body segments (Czerniecki, Gitter, & Munro, 1991; D. I. Miller, 1987).

After the loss of a lower limb below the knee, a prosthesis is fabricated using lightweight materials. Compared to the intact limb, the mass of the prosthetic side is consistently 30-40% less, the center of mass location is 25-35% closer to the knee joint, and the moment of inertia is 50-60% less about a transverse axis through the knee joint (Lin-Chan et al., 2003b; Mattes et al., 2000; J. D. Smith et al., 2014). Using regression equations based on intact body segments to estimate the inertial properties of a prosthesis results in inertial estimates that are inaccurate for the amputated limb. These inaccuracies are larger than the typical variability that would be found between different regression models in the literature. Thus, one might question results from the literature that have used intact estimates to predict inertial properties of the prosthesis, particularly when the research question focuses on the swing phase (J. D. Smith et al., 2014).

J. D. Smith et al. (2014) developed an oscillation rack to directly measure the inertial properties of the prosthesis and residual limb of persons with transtibial amputation (TTA). These values are then used to calculate the appropriate joint moments and powers. They found that these differences in inertia do not result in significant differences in kinetics during the stance phase of walking (J. D. Smith et al., 2014). This is likely due to the significantly larger GRFs overriding the differences in inertial properties during stance. However, during swing, significant differences were found at the hip and knee. Non-significant differences were noted in the ankle. However, these small differences were propagated up the kinematic chain and became significant at the knee and hip. As a result, it is important to investigate these differences during swing where they are more likely to cause differences in the joint kinetics. To date, no regression equations have been developed to assist with these calculations.

Thus, the purpose of this study was two-fold: 1) develop a general method (GENERAL) for estimating prosthesis inertial properties of the amputated limb based on subject-specific (SPECIFIC) measures in a group of individuals with TTA; and 2) evaluate the validity of the GENERAL approach compared to SPECIFIC and intact limb inertial measures. Thus, there were two phases to this study. Phase I addresses model development and Phase II examines the validity of the model. Additionally, we provide an example of the utility of the model using an inverse dynamics approach to calculate lower extremity joint moments during walking. Our specific hypotheses were as follows:

H01 No significant differences will be found between the direct measures of body segment parameters and those obtained from the regression equations on a subset of participants.

H02 No significant differences will be found between the joint moments using the direct measures of body segment parameters and those obtained using regression equations on a subset of participants.

Methods

Participants in both phases were between the ages of 18 and 65 years old, had amputations resulting from trauma, wore their prostheses daily, were very active, and were free from other comorbidities that would influence their walking ability. IRB approval and written informed consent were obtained prior to data collections.

Phase I - Model Development

The GENERAL model was developed from subject-specific (SPECIFIC) inertial properties estimated for 11 persons with TTA (9 males, 2 females, measured body mass =

95.5 ± 16.3 kg, height = 1.78 ± 0.07 m). SPECIFIC data were collected from participants in our lab (n = 5) and pooled with data from the literature (n = 6) obtained through personal communication with the authors (Smith et al, 2014). Total body mass and height were measured while participants wore their prostheses and shoes. The type of prosthetic foot was limited to energy storing and releasing feet in an effort to limit the influence of other foot types on predictive measures. Prosthetic foot types included: Flex Foot, Re-flex, College Park, and Veriflex. From our experience, these types of feet are the most commonly prescribed prosthetic foot types. Participants used various socket suspension systems including: suction, lock and pin, and elevated vacuum. The entire suspension system (including liner and ply) was included in the SPECIFIC measures of the prosthesis inertia.

For SPECIFIC measures, prosthesis mass (with shoe), center of mass location (COM), and moment of inertia (MOI) about a mediolateral axis through the prosthesis COM were determined using a standard scale, reaction board, and oscillation techniques, respectively (J. D. Smith et al., 2014). Briefly, a specially designed oscillation rack (Figure 2.1) was used to determine the moment of inertia of the prosthesis (socket, foot, and shoe).

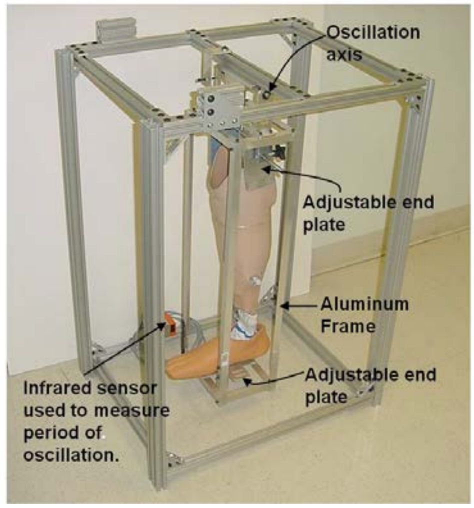

Figure 2.1. Oscillation rack configuration for measuring the period of oscillation of the prosthetic limb. Adapted from J. D. Smith et al. (2014)

Moment of inertia was measured using an oscillation technique. The segment was suspended as a pendulum where the arc of the pendulum is known and the oscillation period (τ) is measured:

$$\tau = \sqrt{\left(\frac{I_{axis}}{mgd}\right)}$$

Where m was the mass of the segment, g was the constant acceleration due to gravity (-9.8 $m \cdot s^{-2}$), and d was the distance from the axis of rotation to the center of mass location. The range of motion for this osculation technique was 5°. The inner cage with the prosthesis was then removed and the center of mass is determined via a reaction board technique (Figure 2.2).

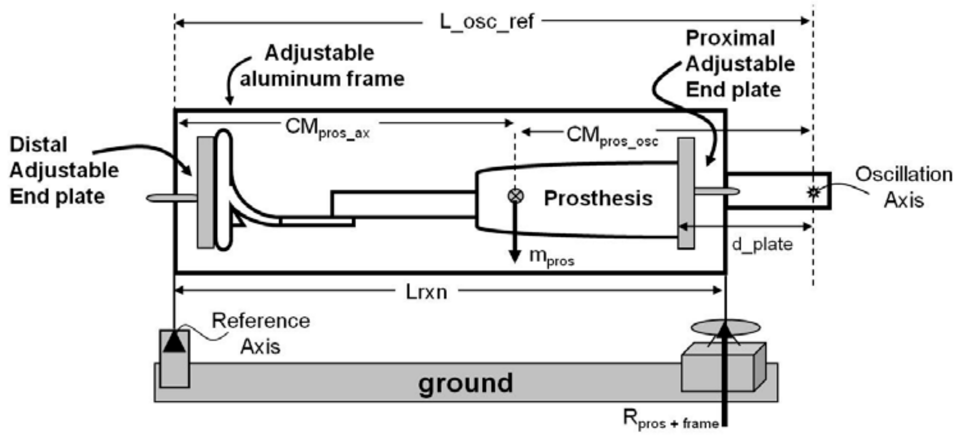

Figure 2.2. Experimental set for the reaction board test to calculate the prosthetic center of mass location. Adapted from J. D. Smith et al. (2014)

The cage with the prosthesis was placed on a board which rested on two “knife edges” and one end rested on a scale. The center of mass location was calculated:

$$x = \left(\frac{R_2\text{-}R_1}{W}\right) * d$$

Where x was the center of mass location, d was the distance between the two knife edges, R_1 was the reaction of the board without the segment, R_2 was the reaction of the board with the segment, W was the weight of the segment. Then the center of mass of the cage alone is measured. Using the parallel axis theorem, the center of mass of the prosthesis was calculated. The center of mass location was expressed as a percentage of the segment length.

Inertial properties of the prosthesis were combined with those of the residual limb that were estimated by modeling the residual limb as a frustum of a right circular cone. This overall limb inertia was then distributed into separate foot and shank segments. First, the overall mass was divided into shank (66%) and foot (34%) masses based on proportions of these masses from a dismantled prosthesis (J. D. Smith et al., 2014). Using percentages from de Leva (1996) for an intact foot segment, the COM and radius of gyration (ROG) of the prosthetic foot were then estimated. These foot estimates were subtracted from the overall inertia of the limb (prosthesis + residual limb) to leave inertial estimates for the combined residual limb and prosthetic shank, which we refer to from this point forward as the prosthetic shank segment. These methods have been described in greater detail in previous papers (J. D. Smith et al., 2014; J. D. Smith & Martin, 2011, 2013).

After SPECIFIC measures for this group of 11 TTAs were obtained, a GENERAL model was created to estimate prosthetic shank and foot inertial properties from mean values of the group (Table 2.1). Prosthetic foot and shank masses can be estimated as a percentage of adjusted body mass. Prosthetic shank COM and ROG lengths were expressed as a percentage of the prosthetic shank length relative to the knee joint. Prosthetic foot COM and ROG were based on percentages reported by de Leva (1996) for an intact foot segment.

Table 2.1

Developed GENERAL model estimates of inertial properties for prosthetic shank and foot based on SPECIFIC measures (n = 11). These data are from the model development phase (Phase I) of the study and should be used to estimate the inertial properties of the amputated limb.

	Mass (%body mass)	COM (% segment length)	ROG (% segment length)
Shank	3.3	21.0	17.1
Foot	1.4	[a]44.15m 40.14w	[a]27.9m 24.5w

Note. [a] Estimates based on de Leva (1996) were gender specific (m: men, w: women); prosthesis inertia values were not.

The computational steps used to estimate GENERAL prosthetic foot and shank inertial properties for participants are illustrated in Figure 2.3 using data for a female

TTA. Steps 1-3 in Figure 2.3 illustrate how to incorporate subject-specific measurements into the GENERAL model for better estimates. There are seven subject-specific measures that are required for application of the GENERAL model: 1) mass of the participant with shoes, 2) mass of the prosthesis with shoe and any liners and ply tucked into the socket, 3) proximal residual limb circumference, 4) distal residual limb circumference, 5) residual limb length, 6) prosthetic foot length (without the shoe), and 7) prosthetic shank length. Steps 4 – 7 in Figure 2.3 illustrate the application of the percentages from the GENERAL model (Table 2.1) to estimate individual foot and shank inertial properties.

Example Calculations
Required measurements for the GENERAL model:

Measure	Subject Data	Symbol
Gender	Female	
Mass of participant with shoes	61.36 kg	MBM
Mass of the prosthesis (with shoe and liner/ply)	2.04 kg	M_{pros}
Proximal residual limb circumference	0.255 m	C_{prox}
Distal residual limb Circumference	0.210 m	C_{dist}
Residual limb length	0.175 m	ℓ_{res}
Prosthetic foot length	0.260 m	ℓ_{foot}
Prosthetic shank length	0.360 m	ℓ_{shank}

Step 1: Volume of the residual limb

Calculate radii: $\boldsymbol{r} = \left(\frac{\boldsymbol{C}}{\boldsymbol{2*\pi}}\right)$

Proximal radius: $R = \frac{0.255\ m}{2*\pi} = 0.041\ m$ Distal radius: $r = \frac{0.210\ m}{2*\pi} = 0.033\ m$

$$\boldsymbol{Volume} = \frac{\boldsymbol{\pi * \ell_{res}}}{\boldsymbol{3}} * (\boldsymbol{R^2 + Rr + r^2})$$

$$Volume = \frac{\pi * 0.175\ m}{3} * (0.041\ m^2 + (0.041\ m * 0.033\ m) + 0.033\ m^2) = .00075\ m^3$$

Step 2: Mass of the Residual limb ($\boldsymbol{M_{residual}}$)

$\boldsymbol{d}$***** = 1100 kg·m^{-3} $\boldsymbol{M_{residual} = d * Volume}$

$$M_{residual} = 1100\ \text{kg} \cdot \text{m}^{-3} * .00075\ m^3 = 0.831\ kg$$

*An assumed tissue density of 1100 kg·m^{-3} (Mungiole and Martin, 1990)

Step 3: Calculate Adjusted Body Mass (ABM)

$$\boldsymbol{ABM} = \frac{\boldsymbol{MBM - M_{pros} - M_{residual}}}{\boldsymbol{1 - c}}$$

$$ABM = \frac{61.36\ kg - 2.04\ kg - 0.8305\ kg}{1 - 0.061} = 62.29\ kg$$

The constant (c) (men = 0.057, women = 0.061) represents the percentage of ABM accounted for by the intact shank and foot (de Leva, 1996). Apply the ABM when estimating intact inertial measures.

Step 4: Calculate segmental masses

Shank % mass: 3.3% ABM
$Shank\ mass = 0.035 * 62.29\ kg = 2.18\ kg$

Foot % mass: 1.4% ABM
$Foot\ mass = 0.014 * 62.29\ kg = 0.87\ kg$

Step 5: Calculate segment COM locations

Shank COM: 21% shank length (ℓ_{shank})
$Shank\ length = 0.21 * 0.36\ m = 0.08\ m$

Foot COM: 40.14% foot length (ℓ_{foot})
$Foot\ length = 0.4014 * 0.26\ m = 0.10\ m$

Step 6: Calculate segment radius of gyration (*k*)

Shank ROG = 19.4% segment length (ℓ_{shank})
$Shank\ ROG = 0.194 * 0.36\ m = 0.07\ m$

Foot ROG = 24.5% segment length (ℓ_{foot})
$Foot\ ROG = 0.245 * 0.26\ m = 0.06\ m$

Step 7: Calculate segment moment of inertia (about the COM axis)

MOI: $\boldsymbol{I = m \cdot k^2}$

$Shank\ MOI = 2.18\ kg * (0.072\ m)^2 = 0.011\ kg \cdot m^2$
$Foot\ MOI = 0.87\ kg * (0.062\ m)^2 = 0.003\ kg \cdot m^2$

Figure 2.3. Example calculation for estimating prosthetic foot and shank inertial properties for a female TTA. Required subject measurements are shown along with calculation steps and equations.

Phase II - Model Validation

Nine individuals with TTA (6 males, 3 females, measured body mass = 78.2 ± 18.8 kg, height = 1.76 ± 0.10 m), who were not included in the model development portion of the study, participated in the model validation phase. Inertial properties of the amputated limb were calculated using three approaches: (1) INTACT – prosthetic leg inertial properties were assumed to match those predicted for the intact leg estimated using de Leva (1996), (2) SPECIFIC – subject-specific measures as described above, and (3) GENERAL – using the model developed above.

To provide an example of the utility of our GENERAL model and its influence on commonly reported joint moments, an inverse dynamics model of walking was used with this second group of participants. Participants walked at 1.5 $m \cdot s^{-1}$ along a 10 m walkway with embedded force plates. Ground reaction forces (2000 Hz) and motion data (100 Hz) were collected. Using a three segment inverse dynamics model, joint moments at the hip, knee, and ankle were computed using the three different inertial models: (1) INTACT, (2) SPECIFIC, and (3) GENERAL. To test for differences between approaches, a single factor, three level (Inertial Model) MANOVA with repeated measures was performed on inertial properties and peak joint moments ($\alpha = .05$).

Results

A significant main effect was found for the overall MANOVA ($p < .0001$, $F(12, 38) = 17.32$). During the *Model Validation* phase of the study, no statistically significant differences were found in estimates of shank mass and COM location between the SPECIFIC and GENERAL models of the shank (Table 2.2). For the prosthetic shank estimates, the SPECIFIC and GENERAL models resulted in ~22% lower mass ($p < .05$)

and a COM location ~55% closer ($p < .05$) to the knee compared to the INTACT approach. Individual variability in shank mass between SPECIFIC and GENERAL models averaged ~14% and ranged between 5 to 34%. Similarly, the shank COM location between the SPECIFIC and GENERAL models averaged a ~30% difference and ranged between 2 – 95%. The largest percent difference between the SPECIFIC and GENERAL models was found in the MOI of the shank where the mean individual error was ~460%. However, this error was likely driven by two factors. First, three participants were above 100% percent error whereas the other 6 participants were on average 50% different between models. The second source of large percent error was due to the relatively small values for MOI. For the prosthetic foot, only the mass of the foot was significantly greater using INTACT compared with SPECIFIC and GENERAL models. All other measures for the prosthetic foot were not significantly different between models given that all measures were based on percentages reported by de Leva (1996).

Table 2.2

Means ± SD of inertial properties calculated using the SPECIFIC, GENERAL and INTACT models (n = 9). These data are from the model validation phase (Phase II) of the study.

		Shank COM			**Shank Mass**			**Shank MOI**		
		SPECIFIC*****	GENERAL*****	INTACT	SPECIFIC*****	GENERAL*****	INTACT	SPECIFIC	GENERAL*****	INTACT
ID	Gender	m	m	m	kg	kg	kg	kg·m^2	kg·m^2	kg·m^2
1	f	0.08	0.09	0.18	1.94	1.76	2.31	0.064	0.009	0.023
2	m	0.06	0.09	0.20	3.35	2.70	3.54	0.001	0.016	0.044
3	m	0.09	0.08	0.18	2.83	2.21	2.90	0.031	0.011	0.029
4	f	0.08	0.08	0.16	2.18	2.05	2.69	0.003	0.008	0.021
5	m	0.07	0.10	0.21	3.30	3.61	4.72	0.023	0.023	0.063
6	m	0.04	0.08	0.16	2.24	2.40	3.15	0.013	0.009	0.025
7	f	0.10	0.08	0.18	2.64	2.44	3.19	0.054	0.012	0.032
8	m	0.06	0.09	0.18	3.12	2.70	3.54	0.001	0.014	0.037
9	m	0.06	0.08	0.16	2.69	3.61	4.73	0.041	0.015	0.040
Mean ± SD		**0.07 ± 0.02**	**0.08 ± 0.01**	**0.18 ± 0.02**	**2.70 ± 0.51**	**2.61 ± 0.64**	**3.42 ± 0.84**	**0.026 ± 0.024**	**0.013 ± 0.005**	**0.035 ± 0.013**

		Foot COM			**Foot Mass**			**FOOT MOI**		
		SPECIFIC	GENERAL	INTACT	SPECIFIC*****	GENERAL*****	INTACT	SPECIFIC	GENERAL	INTACT
ID	Gender	m	m	m	kg	kg	kg	kg·m^2	kg·m^2	kg·m^2
1	f	0.04	0.04	0.04	0.91	0.67	1.04	0.005	0.004	0.004
2	m	0.05	0.05	0.05	1.36	1.02	1.43	0.008	0.006	0.006
3	m	0.05	0.05	0.05	1.19	0.84	1.29	0.006	0.005	0.005
4	f	0.04	0.04	0.04	0.94	0.78	1.09	0.005	0.004	0.004
5	m	0.06	0.06	0.06	1.53	1.37	1.93	0.011	0.009	0.010
6	m	0.05	0.05	0.05	0.95	0.91	1.33	0.004	0.004	0.005
7	f	0.04	0.04	0.04	0.88	0.92	1.25	0.005	0.006	0.006
8	m	0.06	0.06	0.06	1.24	1.02	1.37	0.006	0.005	0.006
9	m	0.05	0.05	0.05	1.12	1.37	1.82	0.019	0.023	0.026
Mean ± SD		**0.05±0.01**	**0.05±0.01**	**0.05±0.01**	**1.12 ± 0.23**	**0.99 ± 0.24**	**1.40 ± 0.27**	**0.008 ± 0.005**	**0.007 ± 0.006**	**0.008 ± 0.007**

Note. *Significant difference from INTACT ($p < .05$)

When the three models were applied to the inverse dynamics model of walking, moment magnitudes were not significantly different at the ankle regardless of inertia model (*Model Validation* group; n = 9). No significant differences were found among the three models during stance. Peak joint moments at the knee and hip during late swing were significantly smaller ($p < .05$) for the SPECIFIC and GENERAL models compared with the INTACT model (Table 2.3). There were no significant differences in moment magnitudes between SPECIFIC and GENERAL models (Figure 2.4).

Table 2.3

Peak joint moments ($Nm \cdot kg^{-1}$) for the ankle, knee, and hip for all three models (n = 9). These data are from the model validation phase (Phase II) of the study.

Joint	**Peak**	**SPECIFIC**	**GENERAL**	**INTACT**
Ankle	Push-off	1.32 ± 0.32	1.32 ± 0.32	1.32 ± 0.32
	Heel Strike	-0.27 ± 0.22	-0.27 ± 0.22	-0.27 ± 0.22
Knee	Heel Strike	-0.42 ± 0.12	-0.40 ± 0.13	-0.42 ± 0.13
	Early Stance	0.54 ± 0.31	0.54 ± 0.31	0.53 ± 0.31
	Mid-Stance	-0.16 ± 0.21	-0.15 ± 0.21	-0.15 ± 0.21
	Push-off	0.43 ± 0.17	0.43 ± 0.17	0.42 ± 0.18
	Terminal Swing	-0.28 ± 0.07*	-0.27 ± 0.07*	-0.40 ± 0.09
Hip	Early Stance	0.87 ± 0.40	0.86 ± 0.40	0.90 ± 0.37
	Mid-Stance	-1.32 ± 0.42	-1.32 ± 0.42	-1.32 ± 0.44
	Terminal Swing	0.34 ± 0.10*	0.38 ± 0.22*	0.69 ± 0.15

Note. *Significant difference from INTACT ($p < .05$)

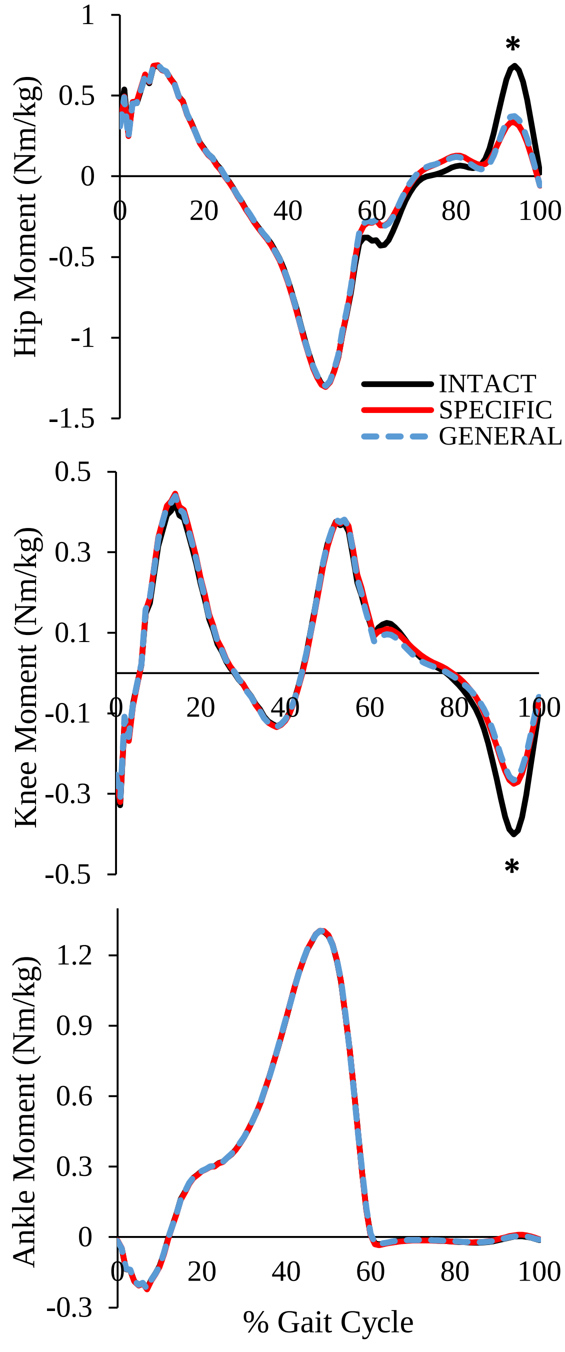

Figure 2.4. Mean hip, knee, and ankle joint moments of 9 TTAs using INTACT, SPECIFIC, and GENERAL inertial models. *Indicates INTACT measures were significantly different from the SPECIFIC and GENERAL models ($p < .05$). No Significant differences were found between SPECIFIC and GENERAL models.

Discussion

In the first phase of the study, we developed a model based on subject-specific measures from 11 persons with TTA and their prostheses to estimate the inertial properties of the prosthetic limb (Table 2.1). In the second phase we applied our developed GENERAL model to a separate group of participants to determine the validity of our model. The results of our validation process suggest our GENERAL model reasonably estimates body segment parameters of the amputated limb using the means of subject-specific measures. For the shank, the SPECIFIC and GENERAL models for the COM location were not different; whereas the SPECIFIC and GENERAL models resulted in the COM location being ~55% closer to the knee than the INTACT model. Additionally, the difference in shank mass between the SPECIFIC and GENERAL models was ~3% whereas the mass was ~21% larger using INTACT measures than the SPECIFIC. For MOI estimates at the shank, GENERAL model estimates were significantly smaller than the INTACT model. However, no significant differences were found between the SPECIFIC and INTACT models. This is likely due to the large variability in the SPECIFIC model. The large variations in MOI estimates are not likely to have a large impact on the joint moments (Challis & Kerwin, 1996). Therefore, our GENERAL model produces inertial property estimates of the prosthetic side that are much more consistent with subject-specific measures than assuming these measures are similar to those of intact segments.

Our inverse dynamics analyses in the second phase also illustrated that our GENERAL model did not result in significantly different joint moments at the knee and hip compared with the SPECIFIC measures during the swing phase of walking. This suggests that as inputs into an inverse dynamics analysis, the GENERAL measures are reasonable inputs for the COM location, MOI, and segment mass. In addition, the GENERAL model is less time consuming (~ 5 min) than SPECIFIC measures (~ 30 min) and does not require specialized equipment to complete. Future researchers working with individuals with below-knee prostheses now have a method for estimating inertial properties of the prosthetic side when subject-specific measures of the prosthesis are unavailable.

CHAPTER III

STUDY 2: BIOMECHANICAL ANALYSIS OF CURB ASCENT IN PERSONS WITH BURGESS AND ERTL TRANSTIBIAL AMPUTATIONS

Introduction

An estimated 2 million Americans live with limb loss resulting from dysvascular disease, trauma, or cancer (Ziegler-Graham et al., 2008). Roughly 25% of these individuals have undergone a transtibial amputation (TTA). These individuals must adapt to various challenges associated with limb loss.

Slips, trips, and falls (STFs) in a community based environment pose a public safety concern. Falls on level or uneven surfaces are the 7th leading cause of death in the US according to the National Safety Council (2011). To avoid STFs the body must accommodate varying surface conditions during normal gait. This task is especially difficult for those with lower extremity amputations. Relative to age-matched, able-bodied individuals, persons with TTA have an increased risk of falling and fear of falling (W. Miller et al., 2001; Vanicek et al., 2009). As a result, 60% of these individuals report falls affect their daily activities, work, and confidence (Kulkarni et al., 1996).

While level over ground walking is important to investigate, more demanding functional tasks of daily living are also important to consider. Persons with TTA report curb negotiation is more challenging than negotiating stairs even though they are encountered with the same frequency (Larsson et al., 2009; Shumway-Cook et al., 2002).

However, the underlying mechanisms contributing to this being a more challenging task to accomplish are less known. To date, we are aware of only one study to investigate curb negotiation in persons with TTA. Barnett, Polman, and Vanicek (2014) investigated curb negotiation following below knee amputation at 1, 3, and 6 months post-surgery. They found persistent asymmetries between the intact and amputated limbs across all time points. Specifically, the intact limb spent more time in stance, produced more power, and had a larger range of motion than the amputated limb while ascending the curb. Since the amount of literature for this task is limited, drawing on tasks similar to stepping up a curb such as obstacle negotiation, stair ascent, and curb negation in non-amputees may guide research in persons with TTA.

During curb ascent, a variety of kinematic variables can be used as indicators to predict the ability to safely traverse obstacles, including a curb. An example of such a measure is foot or toe clearance (Patla, Prentice, Rietdyk, Allard, & Martin, 1999). During curb ascent in non-amputees, minimum foot clearance occurs at the edge of the curb for the lead limb, while minimum foot clearance occurred equally at the edge and surface of the curb for the trail limb (Loverro et al., 2013). Additionally (in non-amputees), while negotiating a curb individuals are able to make adjustments in both step length and step time in order to avoid placing their foot in a potentially hazardous position near the curb (Crosbie & Ko, 2000). Schulz (2011) found that toe clearance increased by altering joint kinematics during swing while negotiating obstacles. Patla and Rietdyk (1993) found as obstacle height increases, toe velocity and hip velocity decreases during the obstacle crossing. The alterations in these kinematic variables (joint angles, velocities, toe clearance) may help reduce fall risk by decreasing the possible impact velocity with the obstacle

In addition to kinematic adaptations to obstacle and curb negotiation, the GRFs involved in these tasks are also altered. During stair ascent, the intact limb of persons with TTA also experiences higher vertical GRFs than the amputated limb and non-amputees (Schmalz et al., 2007) resulting in an asymmetric loading pattern between the limbs. In non-amputees, vertical GRF impulse during double limb support decreased as a function of obstacle height which was attributed to contralateral knee flexion to aid in controlling limb elevation (Patla & Rietdyk, 1993). Additionally, the anterior-posterior impulse during the braking phase increased which coincided with the decrease in forward hip velocity (Patla & Rietdyk, 1993). All of the above adaptations allow the body to rapidly respond to changes in the environment and safely navigate obstacles in non-amputees.

It is also unknown if amputation technique influences an amputee's ability to negotiate a curb. The most commonly used transtibial amputation technique is the modified Burgess (Dionne et al., 2009). A less common amputation technique is the transtibial osteomyoplastic amputation (Ertl). The Ertl technique has been suggested to improve the overall physiology of the residual limb by maintaining the medullary canal pressures and improving vascularization of the remaining tissues (Dionne et al., 2009; Ertl et al., 2013). Using a "bone bridge", the Ertl connects the tibia and fibula sealing the medullary canal and sutures the anterior and posterior residual musculatures together. The Ertl has been suggested to promote greater load-bearing on the distal end of the residual limb (Dionne et al., 2009). Greater residual limb load-bearing has the potential to positively impact long-term outcomes by increasing symmetrical loading between limbs; thereby reducing the increased risk of osteoarthritis in the intact limb joints (Hurley, McKenney, Robinson, Zadravec, & Pierrynowski, 1990). However, limited data related to functional outcomes following Ertl amputations exist. By investigating a more challenging task, such as curb negotiation, it is more likely that any underlying functional differences between surgical techniques would be highlighted.

Therefore, the purpose of this study was to determine whether Ertl amputations lead to a better functional ability to step up onto a curb compared to more traditional Burgess amputations. Since a suggested biomechanical advantage of the Ertl amputation is an increased capability to bear loads on the end of the residual limb, it was hypothesized that those with Ertl amputations would produce greater joint work at the ankle, knee, and hip with the amputated limb while negotiating a curb compared to the Burgess amputated limb. Joint work takes into account kinematic and kinetic variables and provides insight into each joint's individual contribution to the motion. Further joint and net limb work can provide insight into limb asymmetries and differences between amputation techniques.

Methods

Participants

Participants were recruited from the Northern Colorado region. Two groups of transtibial amputees were recruited: traditional (Burgess) (n = 7; 88.3 ± 16.0 kg, 1.78 ± 0.08 m; 55 ± 5 years) and osteomyoplastic (Ertl) transtibial amputees (n = 5; 79.8 ± 15.5 kg, 1.79 ± 0.08 m; 55 ± 8 years). Inclusion criteria included: amputation resulting from trauma, no concomitant musculoskeletal injuries, neurological, or visual impairments, able to understand directions and comprehend the requirements of the study in English. Additionally, all participants were physically active 3 days a week including activities such as long walks, resistance training, and aerobic training.

Institutional Review Board at the University of Northern Colorado provided approval and oversight for the study. Upon arrival at the Biomechanics Lab, informed consent was presented verbally and written consent obtained. A copy of the informed consent was given to the participant.

Data Collection

Participants were asked to change into tight-fitting clothing for data collection purposes. Additional anthropometrics were taken from each participant for use as inputs into the biomechanical model of the person during data analysis. These measures included various segment widths, breadths, lengths, and circumferences. Retroreflective markers were placed with toupee tape on anatomical landmarks on the upper and lower body and trunk. A 10-camera motion capture system (100Hz) was used to capture motion data (VICON, Oxford, UK).

To recreate a curb in the laboratory, two existing force plates embedded in tandem in the regular walking surface of the laboratory were used. Mounted on a steel frame and placed in line with the force plates in the floor, a third force plate was used to create the step of the curb (Figure 3.1). By placing the force plates in this manner, we were able to collect a step on the ground (GROUND) prior to the curb and the step up the curb (CURB). During each trial ground reaction forces from each force plate was collected (2000 Hz).

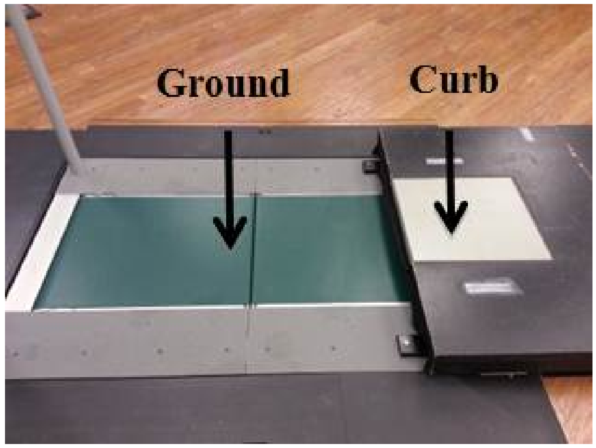

Figure 3.1. Illustration of the curb design. Two forces plates were embedded in the ground and a single force place mounted above the ground level created the curb.

The top of the CURB force plate was 16 cm higher than the GROUND level of the laboratory. A wooden skirt was placed around the curb force plate and extended 3 m beyond the trailing edge of the curb to continue a level walking surface. This simulated a curb similar to what would be encountered on a daily basis. According to the U.S. Department of Transportation Federal Highway Administration, curb heights can vary between 10 and 25 cm (Administration, 2014). The height of 16 cm was model based on the height of the curb outside of the lab where data collections occurred.

Participants were asked to walk up the curb at their self-selected walking velocity. Prior to data collection, the participant was allowed to practice walking up the curb. During these practice trials, the starting position of the participant was adjusted to ensure successful foot strikes on the GROUND and CURB. This ensured that the amputated limb contacted the CURB while the intact limb contacted the GROUND steps. The participant was also adjusted to ensure the intact limb contacted the CURB and the amputated limb contacted the GROUND step on subsequent trials. Data were collected until at least three successful trials were captured with each leg hitting the CURB force plate and each leg hitting the GROUND plate for a total of six successful trials. Walking velocity was measured using a timing system whose measurement area (5 m) encompassed the curb. For a trial to be included for analysis, the walking velocity had to be within ±5% of the participant's preferred walking velocity. Participants traversed ~3 m before they reached the data collection area ensuring a steady state walking pattern emerged before the curb was reached.

Data Analysis

The GROUND and CURB steps were analyzed separately. Marker trajectories (F_c = 6 Hz) and GRFs (F_c = 50 Hz) were low-pass filtered using a fourth-order, zero-lag, recursive Butterworth digital filter. A subject-specific model was created using subject specific anthropometrics. Specifically, for the amputated limb, the inertial properties of the shank and foot were found using oscillation rack and reaction board techniques (J. D. Smith et al., 2014). This model was used to determine 3D angular kinematics were combined with the GRFs using inverse dynamics to estimate joint moments, and powers using Visual3D (C-motion, Germantown, MD).

Participants were placed into two groups based on the surgical technique used for their amputation (Ertl and Burgess). Further, the amputated and intact limbs were also grouped and compared (intact and amputated).

To identify individual joint contributions to the curb negotiation task, joint work of the ankle, knee, and hip for the intact and amputated limbs was estimated as the integral of the power curve for each joint. The joint power curve was not rectified prior to integration in order to characterize both positive and negative joint work. The positive and negative work at each joint was summed to determine the net joint work performed at each joint. Further, the total limb joint work was computed as the sum of the net joint work produced at each joint.

Statistical Analysis

A t-test was used to test for significant differences in walking speed between the groups ($\alpha = .05$, SAS 9.4, Cary, NC). A single factor MANOVA with $\alpha = .05$ (SAS 9.4, Cary, NC) was used to evaluate whether differences in joint work existed between the groups and limbs. Predetermined orthogonal contrasts were performed to assess the following hypotheses:

H01 For the Ertl group, there will be no significant differences between intact and amputated limbs.

H02 For the Burgess group, there will be no significant differences between intact and amputated limbs

H03 Contrasting groups, there will be no significant differences between amputated limbs.

H04 Contrasting groups, there will be no significant differences between intact limbs.

H05 No significant differences will be found between the Ertl amputated limb and the Burgess intact limb.

H06 No significant differences will be found between the Ertl intact limb and the Burgess amputated limb.

Results

All participants were able to negotiate the curb safely with both limbs as the lead limb up onto the curb step and both groups walked at similar speeds (1.28 ± 0.20 m/s Ertl vs. 1.28 ± 0.19 m/s Burgess) up the curb. To simplify the presentation of the remainder of the results, data will be presented below in two main section: results for the steps occurring on the GROUND force plate and results for the steps occurring on the CURB force plate.

GROUND Step Results

A significant model effect was found for the GROUND step (p < .0008, $F(33,30.166) = 3.24$). The intact limbs of both groups produced significantly ($p < .05$) more positive ankle work than both of the amputated limbs during steps on the ground force plate (Table 3.1). Positive work was not significantly different at any other joint. The negative work (Table 3.1) performed at the ankle of the Ertl amputated limb was significantly smaller than the Ertl intact limb ($p = .0489$, $F(1, 20) = 4.40$). The Burgess ampuated limb ankle negative work although not significantly smaller than the Ertl intact limb, did approach significance ($p = .0897$, $F(1, 20) = 3.18$). However, all other negative work performed was similar across groups and limbs.

Table 3.1

Positive and negative joint work ($J \cdot kg^{-1}$) on the GROUND step. Data are mean ± SD.

Group	**Amputated**			**Intact**		
	Ankle	**Knee**	**Hip**	**Ankle**	**Knee**	**Hip**
			Positive Work			
Ertl	0.14 ± 0.05	0.22 ± 0.06	0.53 ± 0.27	0.85 ± 0.18* ‡	0.32 ± 0.14	0.52 ± 0.13
Burgess	0.12 ± 0.05	0.22 ± 0.08	0.63 ± 0.18	0.86 ± 0.28* ‡	0.23 ± 0.07	0.46 ± 0.19
			Negative Work			
Ertl	-0.19 ± 0.08	-0.36 ± 0.17	-0.15 ± 0.11	-0.31 ± 0.12*	-0.42 ± 0.15	-0.22 ± 0.14
Burgess	-0.21 ± 0.08	-0.29 ± 0.08	-0.17 ± 0.12	-0.26 ± 0.11	-0.34 ± 0.05	-0.25 ± 0.24

Note. *Significant difference from Ertl amputated limb.
‡Significantly different from Burgess amputated limb

Net joint work was significantly larger ($p < .05$) in the intact limb of both groups at the ankle compared with the respective amputated limb during the ground step (Figure

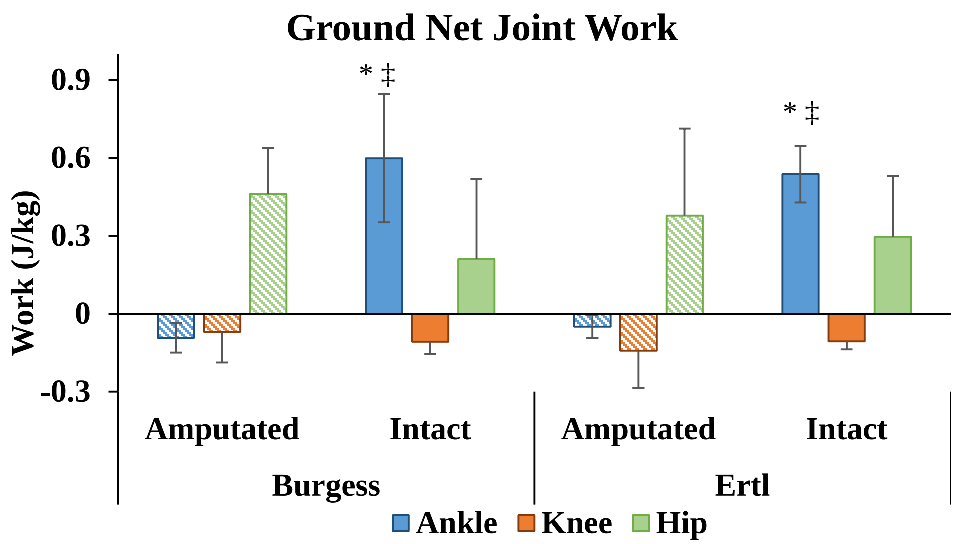

Figure 3.2. Net joint work for the ankle, knee, and hip for the GROUND step. *Significantly different from Ertl amputated limb; ‡Significantly different from Burgess amputated limb.

3.2). Net work at the knee and hip was not significantly different across limbs. Total limb work (the sum of net work at the ankle, knee, and hip) was significantly larger ($p < .05$) in both of the intact limbs compared to both of the amputated limbs (Figure 3.3).

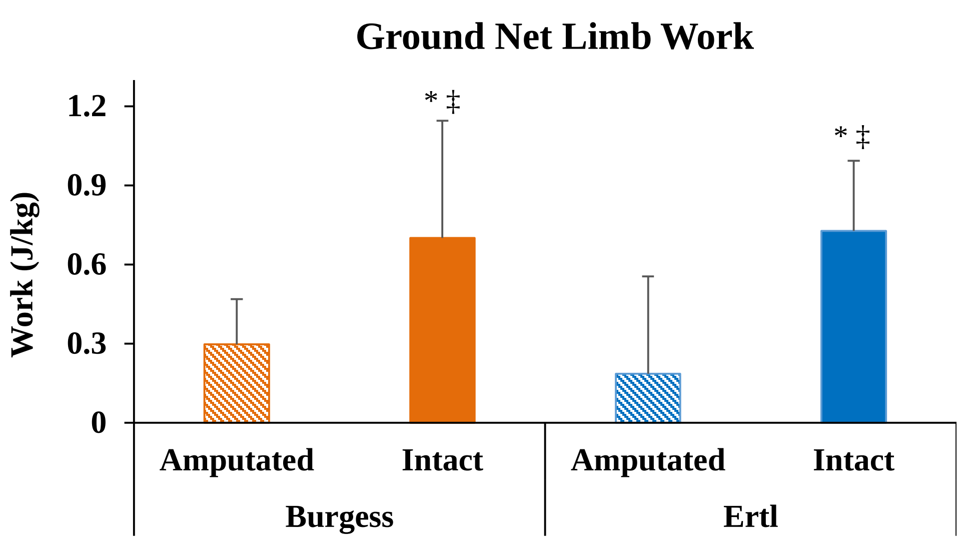

Figure 3.3. Net limb work for the entire limb for the GROUND step. *Significantly different from Ertl amputated limb; ‡Significantly different from Burgess amputated limb

CURB Step Results

A significant model effect was found for the CURB step (Wilk's Lambda: p = .0550, $F(33,30.166) = 1.79$, Hoteling-Lawley Trace: p = .0356, $F(33, 16.19) = 2.34$). As with the ground step, both amputated limbs produced significantly less positive work at the ankle than the intact limbs ($p < .05$). Additionally, the Burgess amputated limb produced significantly less ($p < .05$) positive knee work than both intact limbs (Table 3.2). Yet the Ertl amputated limb produced a positive knee work similar to the Burgess intact limb, but significantly less than the Ertl intact limb ($p = .0401$, $F(1, 20) = 4.82$). No significant differences were found in negative work production between limbs or groups (Table 3.2).

Table 3.2

Positive and negative joint work ($J \cdot kg^{-1}$) on the CURB step. Data shown mean ± SD.

Group	Amputated			Intact		
	Ankle	Knee	Hip	Ankle	Knee	Hip
			Positive Work			
Ertl	0.31 ± 0.21	0.28 ± 0.14	1.23 ± 0.27	0.81 ± 0.55* ‡	0.54 ± 0.29* ‡	1.01 ± 0.10
Burgess	0.22 ± 0.13	0.13 ± 0.07	1.07 ± 0.41	0.77 ± 0.19* ‡	0.47 ± 0.20‡	0.88 ± 0.26
			Negative Work			
Ertl	-0.21 ± 0.08	-0.69 ± 0.38	-0.30 ± 0.20	-0.25 ± 0.14	-0.67 ± 0.28	-0.38 ± 0.26
Burgess	-0.21 ± 0.07	-0.69 ± 0.28	-0.40 ± 0.17	-0.20 ± 0.11	-0.56 ± 0.24	-0.53 ± 0.31

Note. *Significant difference from Ertl amputated limb.
‡Significantly different from Burgess amputated limb

Net work at the ankle was significantly ($p < .05$) smaller for both amputated limbs compared to both intact limbs (Figure 3.4). The Burgess amputated limb produced significantly more negative net knee work than both the intact limbs ($p < .05$). The Ertl amputated limb also produced significantly ($p = .0419$, $F(1, 20) = 4.73$) more negative net knee work than the Burgess intact limb, but not the Ertl intact limb. The Ertl amputated limb differed further from the Burgess intact limb by producing significantly (p = .0164, $F(1, 20)$ = 6.86) more positive net work at the hip.

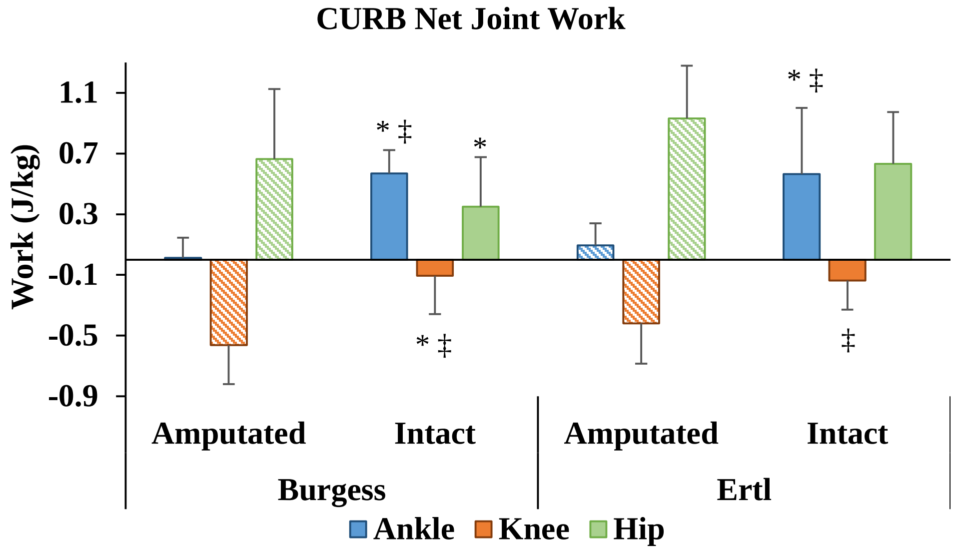

Figure 3.4. Net joint work for the ankle, knee, and hip for the CURB step. *Significantly different from Ertl amputated limb; ‡Significantly different from Burgess amputated limb.

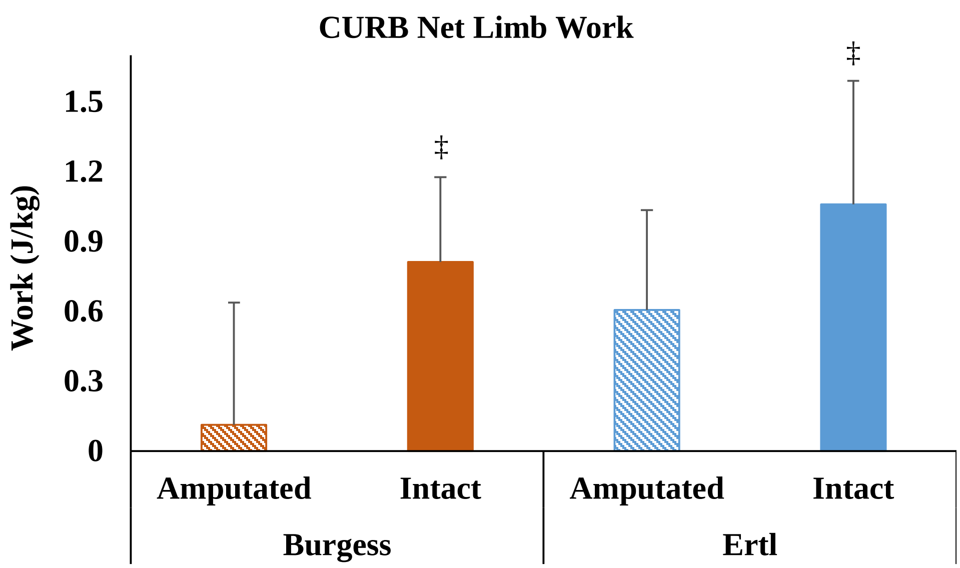

Figure 3.5. Net joint work for entire limb for the curb step. ‡Significantly different from Burgess amputated limb.

Total limb work showed the Burgess amputated limb produced significantly ($p < .05$) less work compared to each of the intact limbs (Figure 3.5). However, no significant differences were found between the Ertl amputated limb and each intact limb. The contrast of total limb work between the Burgess amputated limb and Ertl amputated limb approached significance ($p = .0845$, $F(1, 20) = 3.30$) suggesting that the Ertl limb overall was performing more net joint work than the Burgess limb.

Discussion

The purpose of this study was to determine whether biomechanical differences existed between individuals with Ertl and Burgess amputees during curb ascent. The steps on the GROUND and the CURB were analyzed separately for this task. Overall, the main finding on the GROUND was that the amputated limb of both groups produced significantly less work than the intact limb. The large amount of work produced by the intact limbs is likely attributed to the large amount of power produced (from which work was derived) at the ankle (Figure 3.6) during the push-off phase of the gait cycle (~45 – 60%). The amputated limbs produced ~78% less peak power than the intact limbs during this time. These observations are similar to those reported by Barnett et al. (2014) during a similar curb negotiation task in persons with TTA.

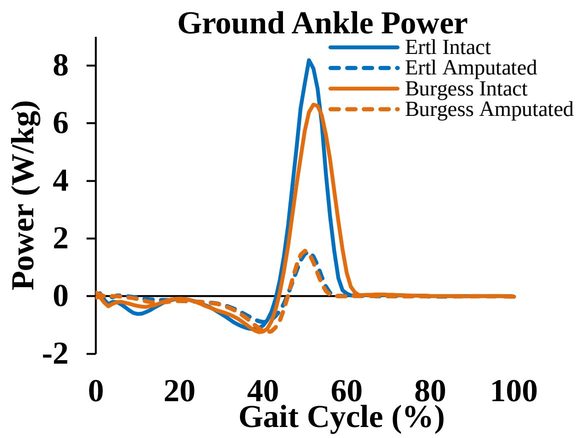

Figure 3.6. Ankle power for the GROUND step. Data are shown from heel strike to heel strike.

Relative to level over-ground walking, the overall pattern of the power curves for the hip, knee, and ankle are similar to the GROUND step for people with TTA (Bateni & Olney, 2002; Sadeghi, Allard, et al., 2001; Winter & Sienko, 1988). However, the overall magnitude of the ankle power is ~50% greater on the GROUND compared to published values of level walking. This suggests that the initial stance phase of the GROUND step is similar to over ground walking, whereas the push-off portion of the stance phase differs due to the transition from the GROUND to the CURB. The increase in push-off power assists with the translation of the body vertically onto the CURB step rather than a continuation of forward movement as in level over-ground walking.

Results from the CURB step suggest that the Ertl amputated limb produced overall limb work of a similar magnitude to the intact limbs of the Ertl and Burgess groups. However, the way in which the total work was produced is different for the Ertl amputated limb and both intact limbs. The intact limbs produced a large amount of net work at the ankle. In contrast, the Ertl amputated limb produced more net hip work and very little net ankle work. The amount of work produced by the Ertl amputated limb at the hip was significantly larger than the Burgess intact limb.

While joint work did provide insight into the net limb contributions to the movement, timing of work production during the gait cycle is unknown. To understand the timing of the work production, qualitative assessment of the power profiles was performed (Figure 3.7). Although both amputated limbs produced similar power profiles at the ankle (Figure 3.7), the power profile at the hip and knee showed a different trend between the two amputated groups. The hip power profile (Figure 3.7 C), shows the Ertl amputated limb produced larger peak hip power in early stance (~15% of gait cycle) compared to the Burgess amputated limb and both intact limbs.

This increase in hip power production of the Ertl amputated limb is consistent with results published for traversing a curb similar to the current task (Barnett et al., 2014). Although Barnett et al. (2014) used a lower height for curb height (7.5 cm vs 16 cm in the current study), there is a clear trend for the amputated limb (of both groups) hip joint to generate a larger peak power during early stance than the intact limb. In contrast, the intact limb of both groups generated more knee power during early stance (Figure 3.7). The inter-limb differences suggest that the amputated limbs adopt more of a hip strategy than a knee strategy (used by the intact limb) during early stance. In the current study, this hip strategy is even more exaggerated by the Ertl amputated limb where it produced more hip power than the Burgess amputated limb during early stance.

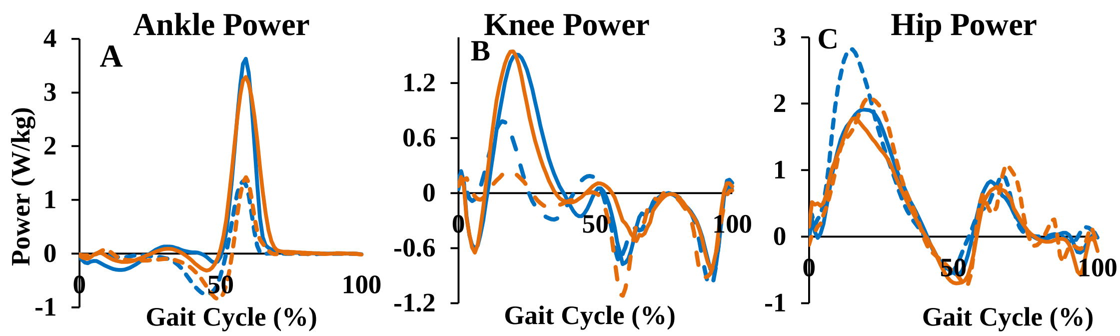

Figure 3.7. Joint powers for the ankle (A), knee (B) and hip (C) for the curb step. (—) Ertl intact, (—) Burgess intact, (--) Ertl amputated, (--) Burgess amputated.

In comparison to Barnett et al. (2014), the peak power production at the hip is smaller than the Ertl amputated limb; however, the Burgess amputated limb was similar in magnitude. There are several possible reasons for these differences: 1) there may be a surgical technique effect, 2) Barnett et al. (2014) included dysvascular amputees. All of these factors can influence the comparisons between these two studies.

The hip and knee power profiles of the CURB step also resemble published hip and knee power profiles for stair ambulation (Aldridge, Sturdy, & Wilken, 2012; Alimusaj, Fradet, Braatz, Gerner, & Wolf, 2009; Powers, Boyd, Torburn, & Perry, 1997). However, the overall magnitude of the hip power is lower during the CURB step than stair ambulation (Aldridge et al., 2012; Yack, Nielsen, & Shurp, 1999). This lower power production is also associated with less work produced on the CURB step than stair ambulation for all limbs (Yack et al., 1999). Thus, these differences suggest the demands of curb negotiation is different than stair ascent. Where stair ascent requires larger quantities of power generation to assist with vertical movement, the CURB only requires moderate levels of power generation to ascend the single step and progress forward. Finally, curb negotiation differs further from stair negotiation due to the lack of a handrail to use for support. When participants are allowed to use a handrail for stair ascent, joint moments decrease (Reeves, Spanjaard, Mohagheghi, Baltzopoulos, & Maganaris, 2008) and dynamic stability increases in older adults (Reid, Novak, Brouwer, & Costigan, 2011). This suggests that the presence and use of a handrail can influence how stairs are negotiated. In this task, and curb negotiation in general, a handrail is not available and may produce results which further differ from stair ambulation.

Although we were able to identify differences between limbs and groups, limitations existed in this current study. Socket type or prosthetic foot type was not controlled which may have affected the way in which the amputated limb was used. The Ertl group were not prescribed sockets which would have allowed for the distal end of the residual limb to bear loads. Sample size was also a limiting factor of this study. Given that some comparisons approached significance, adding more subjects may lead due these observed differences becoming significant.

This work contributes to the overall body of literature by investigating curb ascent in Ertl and Burgess below knee amputees and evaluates biomechanical differences between the two groups. Both amputated limbs behaved in similar ways on the GROUND step and produced less work than the intact limb due to smaller ankle power production. Overall comparisons between functional outcomes of the Ertl and Burgess groups suggest the Ertl amputated limb may behave differently than the Burgess limb by producing more net limb work on the CURB step than the Burgess amputated limb which was produced at the hip during early stance. The increased work performed by the Ertl amputated limb may be a result of increased end-load bearing ability of the limb compared to the Burgess amputated limb. This suggests supports the hypothesis that the Ertl and Burgess amputated limbs behaved dissimilarly while on the CURB step.

CHAPTER IV

STUDY 3: BIOMECHANICAL ANALYSIS OF SIT TO STAND IN PERSONS WITH BURGESS AND ERTL TRANSTIBIAL AMPUTATIONS

Introduction

People with transtibial amputation (TTA) sit-to-stand (STAND) roughly 50 times per day (Bussmann et al., 2004; Bussmann et al., 2008), which suggests that this task is an important activity of daily living. While standing from a seated position, unilateral TTA patients rely more on the unaffected leg to produce force against the ground (Agrawal et al., 2011; Ozyurek, Demirbuken, & Angin, 2014). More specifically, Agrawal et al. (2009) and Ozyurek et al. (2014) found patients with TTA produced 27% more peak vertical GRF with the intact limb during a sit-to-stand movement compared with the prosthetic side.

Similar weight shifts to the non-involved limb during the STAND movement have been observed in other patient populations including patients who have undergone total knee replacement (Farquhar, Kaufman, & Snyder-Mackler, 2009; Mizner & Snyder-Mackler, 2005), hip replacement (Talis et al., 2008), or have hemiparesis (Roy et al., 2006). Non-amputee controls, however, exhibited less than 10% asymmetry in vertical GRF during the STAND movement (Ozyurek et al., 2014). Additionally, there appears to be no evidence to support a limb preference ("dominance") in non-amputees (Schofield et al., 2014) which may have accounted for the symmetry differences. However, the same group did identify timing differences in peak joint moments and powers between limbs during the sit-to-stand task (Schofield et al., 2013).

Two amputation techniques are frequently used to amputate a limb below the knee: Burgess amputation and osteomyoplastic (Ertl) amputation (Assal et al., 2005; Commuri et al., 2010; R. Dederich, 1983; Dionne et al., 2009; Ertl et al., 2013). The traditional and most commonly used transtibial amputation technique is the modified Burgess (Dionne et al., 2009). However, this amputation technique often leads to difficulties after amputation such as pain, swelling, instability, and significant residual limb atrophy. Although less common, the Ertl technique has been suggested to lead to improved functional outcomes following amputation. Using a “bone bridge”, the Ertl technique connects the tibia and fibula, seals the medullary canal, and sutures the anterior and posterior musculatures together. This technique leads to improved physiological function of the residual limb, reduced incidences of bone spurs, increased vascularity, and reduced incidence of skin ulcers (Rolf Dederich, 1963; Dudek et al., 2003; Dudek et al., 2005; Potter et al., 2007).

It has also been suggested that the Ertl technique may enhance “end-bearing” capability of the residual limb compared to the Burgess (Mongon et al., 2013). This improved “end-bearing” may reduce loading asymmetries compared to the Burgess; thus, potentially reducing the risk of developing osteoarthritis (OA), low back pain, or other comorbidities (Burke et al., 1978; Lemaire & Fisher, 1994; Melzer et al., 2001). To date, only one study has investigated the distal end loading within the socket of an Ertl amputee wearing a total surface bearing prosthesis (Commuri et al., 2010). The distal sensors indicated weight was borne on the distal end of the residual limb. However, these results were not compared to a Burgess amputation and thus conclusions between the techniques are inconclusive. Given the lack of data related to functional outcomes following Ertl amputations, determining whether the Ertl amputation technique has a functional advantage and is able to reduce loading asymmetries over the more common Burgess technique is needed.

The purpose of this study was to determine whether there are functional differences during a STAND task between individuals who had amputations performed using a Burgess technique or Ertl technique. It was hypothesized that those with an Ertl amputation would be able to take advantage of the greater end-load bearing capability of the residual limb, which will be evidenced by greater vertical GRF production and joint power magnitudes at the knee and hip. Thus, it was hypothesized that a greater kinetic symmetry between the amputated side and intact side would occur in Ertl amputees during the STAND task.

Methods

Participants

Participants were recruited from the Northern Colorado region. Two groups of transtibial amputees were recruited: traditional (Burgess) (n = 7) and osteomyoplastic (Ertl) transtibial amputees (n = 11). Participants were between 43-65 years of age. Inclusion criteria included: amputation resulting from trauma, no concomitant musculoskeletal injuries, neurological, or visual impairments, able to understand directions and comprehend the requirements of the study in English. Additionally, they were physically active 3 days a week including activities such as long walks, resistance training, and aerobic training.

Although no pilot data were used to establish our sample size, previous sit-to-stand literature suggested a sample population of ten individuals is sufficient to identify differences between groups during this task. In studies comparing tasks in young adults and young adults to older adults, sample sizes ranged from 6-10 individuals per group (Agrawal et al., 2011; Bussmann et al., 2004). Additionally, using data from multiple sources and multiple tasks, such as walking, sit-to-stand, and stair ambulation, a power analysis was performed (Agrawal et al., 2011; Ferris, Aldridge, Rabago, & Wilken, 2012; Schmalz et al., 2007). The comparisons for these studies were between control subjects and amputees and between two prostheses. It was found that on average, a sample size of 5-7 individuals would be sufficient to identify significant differences between these groups. Effect sizes for these data were on average 2.7; ranging from 1-7.1.

The Institutional Review Board at the University of Northern Colorado provided approval and oversight for the study. Upon arrival at the Biomechanics Lab, informed consent was presented verbally and written consent obtained. A copy of the informed consent was given to the participant.

Data Collection

Eleven individuals with Ertl amputation (79.4 ± 16.7 kg, 1.77 ± 0.08 m) and seven individuals with a Burgess amputation (88.3 ± 16.0 kg, 1.78 ± 0.08 m,) were recruited. Participants were asked to change into tight-fitting clothing for data collection purposes. Additional anthropometrics were taken from each participant for use as inputs into the biomechanical model of the person during data analysis. These measures included various segment widths, breadths, lengths, and circumferences. Retroreflective markers were placed with toupee tape on anatomical landmarks on the upper and lower body and trunk. A 10-camera motion capture system (100Hz) was used to capture motion data (VICON, Oxford, UK).

Each participant’s fibular head height (measured from fibular head to floor) was measured and an adjustable seat was adjusted to this height. The seat was placed directly behind two force plates embedded in the floor. These force plates (AMTI, Waterford, MA) were used to measure ground reaction forces (2000Hz) during the sit-to-stand task. The participant was verbally instructed to stand comfortably such that each foot was placed on separate force plates with the seat behind them so that they could sit down on the seat. The legs of the seat did not contact the force plates, so GRFs that were measured only reflected the force under each foot. Each participant was allowed to determine the best foot placement for him or her to be able to complete the task safely.

A five times sit-to-stand task was used, where participants completed five sequential repetitions of sitting to standing and standing to sitting as fast as possible. The participant was instructed to begin standing then “sit” down on the seat behind them (their bottom had to make contact with the seat) then stand up quickly. This completed one repetition. During the task, the participant was told not to use his/her hands to push off the legs or chair in order to stand. However, arm position was not otherwise restrained. Prior to data collection, the participant was allowed to practice the task and adjust foot placement until they felt comfortable with the task. Once the participant was comfortable, data collection began. One five time sit-to-stand task was recorded. A researcher counted aloud how many repetitions the participant completed while using a stopwatch to record how long it took to complete the five repetitions. This time was recorded on the data collection sheet.

Data Analysis

Participants were placed into two groups based on the surgical technique used for their amputation (Ertl and Burgess). Further, the amputated and intact limbs were also grouped and compared (intact and amputated). The middle three repetitions of the STAND motion were biomechanically analyzed. Each repetition was analyzed separately. Marker trajectories (F_c = 6 Hz) and GRFs (F_c = 50 Hz) were low-pass filtered using a fourth-order, zero-lag, recursive Butterworth digital filter. A subject specific model was created using Visual3D (C-motion, Germantown, MD) and the anthropometric data collected previously. The inertial measures of the amputated limb were estimates based on the intact limb. Although this over estimates the inertial properties of the prosthesis, pervious literature has shown that during stance phase, the ground reaction forces are significantly larger than the inertial forces of the segment and the results are equivocal (J. D. Smith et al., 2014). The subject model was used to determine 3D angular kinematics as inputs into inverse dynamic equations.

Peak vertical GRFs for each limb were calculated for each repetition. The peak force was used to calculate a symmetry index (SI) between the two limbs (intact and amputated):

$$SI = 100 - 100 * \frac{I - P}{(I + P)}$$

Where *I* represents the peak vertical GRF for the intact limb and *P* represents the peak vertical GRF for the amputated limb of the same individual (Agrawal et al., 2011).

Hip and knee peak joint angles were compared to understand if there were differences in how each group completed the task kinematic ally. To identify individual joint kinetic contributions, joint powers were calculated for the hip and knee joints. Joint powers were calculated as the product of the specific joint moment (knee and hip) and the joint's angular velocity. To further understand the results from the joint powers, total joint work was calculated as the integral of the rectified power curve for the knee and hip joints. Joint angular impulse was calculated as the integral of the rectified joint moment curve for the knee and hip joints.

Statistical Analysis

A t-test was used to test for significant differences in sit-to-stand time between the groups ($\alpha = .05$, SAS 9.4, Cary, NC). A single factor MANOVA with $\alpha = .05$ (SAS 9.4, Cary, NC) was used to evaluate whether differences in joint work existed between the groups and limbs. Predetermined orthogonal contrasts were performed to assess the following hypotheses:

H01 For the Ertl group, there will be no significant differences between intact and amputated limbs.

H02 For the Burgess group, there will be no significant differences between intact and amputated limbs

H03 Contrasting groups, there will be no significant differences between amputated limbs.

H04 Contrasting groups, there will be no significant differences between intact limbs.

H05 No significant differences will be found between the Ertl amputated limb and the Burgess intact limb.

H06 No significant differences will be found between the Ertl intact limb and the Burgess amputated limb.

Results

All participants were able to complete the five time sit-to-stand task. From a functional perspective, the ERTL group was able to perform the sit-to-stand task significantly faster (p = .0052, $t(16) = 2.901$) than the Burgess group (9.33 ± 2.66 s vs 13.27 ± 2.83 s). Below, mechanical differences between groups which led to this functional difference will be presented. An overall model effect was found between the limbs (p = .0103, $F(60, 39.619) = 2.02$).

Kinematics

At the beginning of the STAND task, the hips and knees were positioned at ~68-88° of flexion (Figure 4.1). As the participant rose, the flexion angle decreased to ~0° flexion (full extension). The maximum knee flexion angle of the Ertl amputated limb was significantly larger (~10°) than the Burgess amputated limb ($p = .009$, $F(1, 32) = 7.74$) and intact limb ($p = .0432$, $F(1, 32) = 4.43$). Additionally, the Ertl intact limb was significantly more flexed at the knee (~8°) than the Burgess amputated limb ($p = .0399$, $F(1, 32) = 4.59$). No significant differences were found between the peak knee flexion angle of the intact and amputated limbs of the same group. This suggests the Burgess group adopted a less flexed knee than the Ertl group (~80° Burgess vs ~ 88° Ertl). However, overall range of motion for the knee and hip was not significantly different between groups or limbs. No significant differences were found in the peak hip angle.

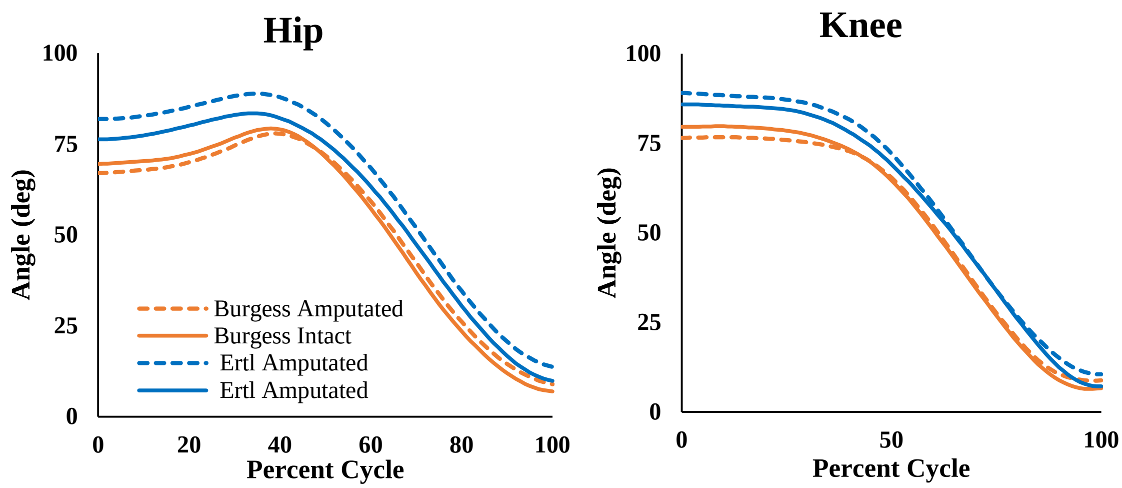

Figure 4.1. Mean sagittal hip and knee joint angles during the STAND task. 0 degrees indicates full extension at the hip and knee.

Ground Reaction Forces

Peak GRFs, normalized to body weight, were significantly smaller for the amputated limb compared to the intact limb for both groups (Table 4.1). Additionally, the Ertl amputated limb produced significantly (p = .033, F(1, 32) = 4.97) more peak force at a percentage of body weight than the Burgess amputated limb (Table 4.1). The SI between limbs was less than 100 for both group (88.35 ± 11.9 Ertl, 85.15 ± 7.31 Burgess) indicating that the intact limb produced more force than the amputated limb. However, Ertl and Burgess groups exhibited a similar level of loading asymmetry as the SI for both groups were not significantly different.

Table 4.1

Peak ground reaction forces as a percentage of body weight (BW) and symmetry index between the amputated and intact limbs. Data shown are mean ± SD.

Group	Amputated GRF (%BW)	Intact GRF (%BW)	SI
Ertl	63.3 ± 8.4	80.3 ± 11.6*‡	88.34 ± 11.92
Burgess	52.5 ± 8.0*	71.1 ± 11.0‡	85.15 ± 7.31

Note. *Significantly different from Ertl amputated limb (p < .05)
‡Significantly different from Burgess amputated limb (p < .05)

Angular Impulse

The intact limbs had significantly larger knee angular impulse than the amputated limbs (Table 4.2). The Ertl intact limb had significantly smaller hip angular impulse than the Burgess intact limb (p = .0033, $F(1, 32) = 10.11$). Further, another trend was noted between the Burgess limbs where the intact limb showed a trend towards a larger hip angular impulse than the amputated limb (p = .0691, $F(1, 32) = 3.54$). These trends suggest the Ertl intact limb is acting differently than the other limbs.

Table 4.2

Angular impulse (Nm·s·BW^{-1}) of the ankle, knee, and hip during STAND. Data are shown mean ± SD.

Group	Amputated			Intact		
	Ankle	Knee	Hip	Ankle	Knee	Hip
Ertl	0.011 ± 0.004	0.009 ± 0.005	0.023 ± 0.009	0.008 ± 0.004	0.024 ± 0.006*‡	0.019 ± 0.007
Burgess	0.008 ± 0.004	0.013 ± 0.008	0.022 ± 0.003	0.012 ± 0.004	0.020 ± 0.008*‡	0.029 ± 0.005†

Note. *Significantly different from Ertl amputated limb (p < .05)
‡Significantly different from Burgess amputated limb (p < .05)
†Significantly different from Ertl intact limb (p < .05)

The large angular impulse (compared to the ankle and hip) for the amputated limb for both groups suggests that the primary contributor to the motion is the hip (Table 4.2).

This observation also holds true for the Burgess intact limb. However, for the Ertl intact limb, the primary contributor is the knee. Of the 11 Ertl participants, 8 chose to adopt the knee as the primary contributor to the motion for the intact limb. In contrast, of the 7 Burgess participants, only 1 chose to use the knee as a primary contributor for the intact limb. With one exception (in the Burgess group) the amputated limb of both groups relied on the hip as a primary contributor to the motion.

Joint Work

At the knee, the amputated limb for both groups generated significantly (all *p-values* $< .0332$) less total work than the knee of the intact limbs (Table 4.3). The Ertl intact limb also produced more work at the knee than at the knee of the Burgess intact limb ($p = .0104$, $F(1, 32) = 7.40$). Total work was not significantly different at the hip between groups or between limbs. Thus, one factor that contributed to the Ertl group's ability to complete the five time sit-to-stand task factor was the greater amount of work performed by the intact limb's knee of the Ertl group.

Table 4.3

Total work for the knee and hip joints.

Group	**Amputated**		**Intact**	
	Knee	**Hip**	**Knee**	**Hip**
Ertl	0.20 ± 0.14	0.65 ± 0.28	$0.76 \pm 0.25^{*\ddagger}$	0.51 ± 0.33
Burgess	0.26 ± 0.24	0.42 ± 0.21	$0.51 \pm 0.21^{*\ddagger\dagger}$	0.62 ± 0.24

Note. *Significantly different from Ertl amputated limb ($p < .05$)
‡Significantly different from Burgess amputated limb ($p < .05$)
†Significantly different from Ertl intact limb ($p < .05$)

Joint Powers

Peak power for both the knee and hip occurred at ~50% of the cycle. Peak knee power of the amputated limb for both groups was significantly smaller (~65% smaller) than their respective intact limb (Table 4.4). However, the Ertl intact limb produced significantly greater peak knee power (~50% greater) than the Burgess intact knee. No significant differences were found between limbs or groups at the hip. However, the Ertl amputated limb showed a trend toward producing more peak hip power than the Burgess amputated limb ($p = .078$, $F(1, 32) = 3.32$). Thus, the faster time of the Ertl group appears to be driven by greater power production at the knee intact limb and possibly the hip of the Ertl amputated limb.

Table 4.4

Peak Power for the knee and hip joints ($W \cdot kg^{-1}$). Data are shown mean ± SD.

Group	**Amputated**		**Intact**	
	Knee	**Hip**	**Knee**	**Hip**
Ertl	1.02 ± 0.53	2.06 ± 0.85	2.92 ± 1.25*‡	1.95 ± 1.15
Burgess	0.71 ± 0.34	1.25 ± 0.73	1.99 ± 0.67 *‡†	1.86 ± 0.72

Note. *Significantly different from Ertl amputated limb ($p < .05$)
‡Significantly different from Burgess amputated limb ($p < .05$)
†Significantly different from Ertl intact limb($p < .05$)

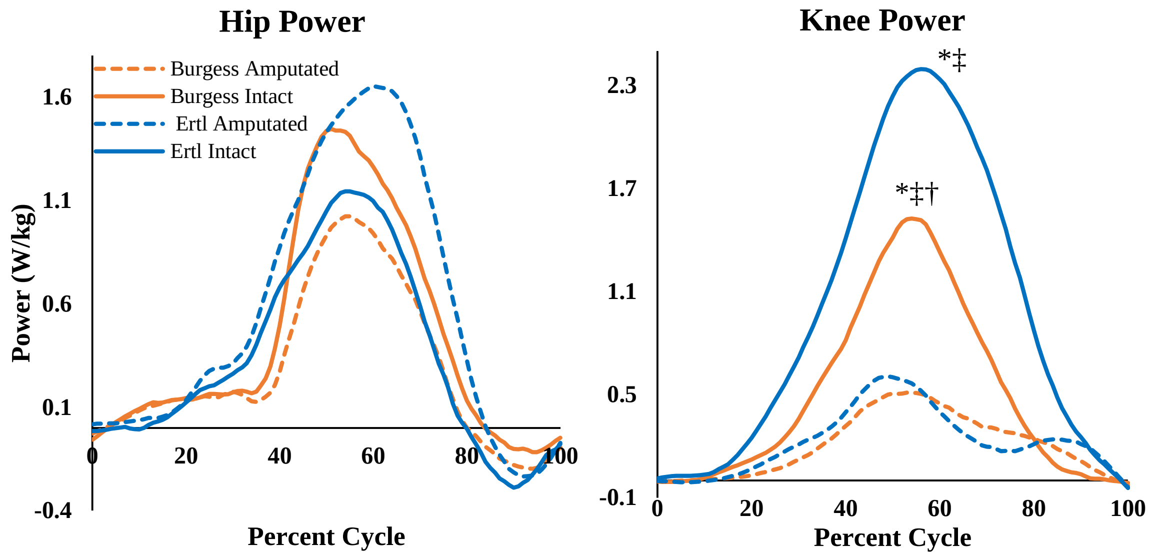

Figure 4.2. Sagittal plane joint powers for the hip and knee joints during the STAND task. At 0 % of the cycle the person is sitting and at 100% of the cycle, the person is fully standing. *Significantly different from Ertl amputated limb; ‡Significantly different from Burgess amputated limb; †Significantly different from Ertl intact limb ($p < .05$).

Discussion

This investigation aimed to identify differences in the functional performance of the STAND task in two groups of TTAs (Burgess and Ertl). We used several methods to evaluate differences in the groups via a timed clinical measure and several biomechanical outcomes. The timed measure found that the Ertl group was able to perform the five time sit-to-stand task more quickly than the Burgess group. Since both groups were similarly active and wore similar prostheses, this suggests that the Ertl group was able to perform the task in a manner that differs from the Burgess group which resulted in better clinical performance.

The faster time to complete the task in the Ertl group differs from previous reports in the literature (Dougherty, 2001; Keeling et al., 2013; Pinzur, Beck, Himes, & Callaci, 2008). These studies relied on the responses from questionnaires answered by the participants (e.g.: Short Form 36 (SF-36), Prosthesis Evaluation Questionnaire (PEQ)) to identify whether functional differences existed between Burgess and Ertl patients. Although these questionnaires are powerful tools and have been validated, they are still subjective measures. In contrast, the data reported in our study, are the first quantitative data that point to measureable differences in a functional task between the Ertl and Burgess groups. These functional measures lend support for the Ertl procedure over the Burgess. This is true despite increased tourniquet times during surgery, longer post-operative healing and potential for increased post-operative complications (e.g., lack of bone-bridge ossification) that are often cited as reasons not to perform an Ertl procedure (Taylor, French, Poka, Blint, & Mehta, 2010; Taylor & Poka, 2011).

Kinematically, the Ertl group adopted a knee position that placed the shank perpendicular to the ground, whereas the Burgess group placed their feet slightly more forward. It is interesting that each group freely chose to adopt two different foot placements. The placement of the feet closer to the chair with increased knee flexion has been shown to reduce stand time in non-amputees (Khemlani, Carr, & Crosbie, 1999). Though, Khemlani et al. (1999) showed differences in rise times, the magnitude of differences between foot placements was ~0.1s. Rise time was shorter when the feet were placed in a more flexed position compared to being placed at 90°. Although it is unclear how foot placement affects movement in this population, it is clear that foot placement will likely influence lower extremity joint kinetics during the STAND task. We did not control foot placement for our task, we only required that one foot be positioned on each force plate while the person was seated and we made an effort to position the seat as close as possible to force plates for both groups. It would be worth investigating whether the burgess group would perform faster on this task if they were forced to begin the task with the shank positioned perpendicular to ground.

Kinetic evaluation showed differences between limbs and groups. We observed similar inter-limb GRF asymmetries between the amputated and intact limbs regardless of group. Our SI results concur with those previously reported (Ranging from ~75% - 85% SI) for below knee amputees performing the stand task (Agrawal et al., 2011; Ozyurek et al., 2014; Slajpah, Kamnik, Burger, Bajd, & Munih, 2013). However, the absolute vertical GRF data show that the Ertl amputated limb was able to produce significantly more force than the Burgess amputated limb (~62% BW vs. ~53% BW respectively). The Ertl amputated limb may have been able to tolerate higher loads on the amputated limb and contribute more to the overall motion and faster time of the Ertl group.

With the exception of the symmetry index, the Ertl amputated limb acted in a similar manner to the Burgess amputated limb. Joint power, work, and moments were similar between the amputated limbs. Further, the relationship between the intact and amputated limb was also similar for both amputation techniques. These results did not support our hypothesis that the Ertl amputated limb would behave in a manner more like the intact limb.

However, we did observe an unexpected and interesting outcome in the Ertl intact limb. In addition to the dissimilarities to the Ertl amputated limb, the Ertl intact limb behaved dissimilarly to the Burgess intact limb. The Ertl intact limb produced significantly larger peak knee joint powers, knee moments, knee angular impulse, and total knee work than the Burgess intact limb. In contrast, both limbs of the Burgess group and the Ertl amputated limb adopt a hip strategy (increased hip work, hip power, and hip angular momentum) to accomplish the task. The shift from the hip to the knee suggest that the Ertl intact limb adopts a knee strategy to accomplish the stand task.

Shifts from a knee to hip strategy has been seen in multiple patient populations along with an increase in trunk flexion during standing (Doorenbosch, Harlaar, Roebroeck, & Lankhorst, 1994; Gross, Stevenson, Charette, Pyka, & Marcus, 1998; Roebroeck, Doorenbosch, Harlaar, Jacobs, & Lankhorst, 1994). Although hip flexion was not different between the Ertl and Burgess groups, this does not mean that the trunk angle was not different. Due to the flexibility of the spine, there may have been group differences that were not measured using the current methodology.

Further, with added mass to the trunk, a shift to an increase reliance on the hip to assist in the sit-to-stand movement has also been observed (Savelberg, Fastenau, Willems, & Meijer, 2007; Van der heijden, Meijer, Willems, & Savelberg, 2009). More specifically, Van der heijden et al. (2009) found that as the demands become too great, due to decreased muscular strength, the hip and ankle increase their contributions to the overall movement by up to ~60%. They suggested a decrease in knee extensor strength may decrease the ability to perform the sit-to-stand task without additional assistance from the hip and ankle. This explanation may provide a reason as to why the Ertl adopted a knee strategy. It may suggest that the Ertl intact leg knee extensors were stronger than the Ertl amputated limb and the Burgess intact limb. The implication of the Ertl intact limb adopting a strategy that is more akin to a non-amputee suggests there may be benefits to strength training for both limbs.

This study was limited in a few ways. Each person (regardless of amputation technique) wore a different prosthetic socket suspension type which included: lock and pin, vacuum, and elevated suction. Moreover, none of the Ertl participants wore a socket designed to allow for more end loading-bearing of the residual limb. As a result, it may not be surprising that we did not observe more symmetrical loading in the Ertl group. Additionally, foot placement may have played an important role in the time to stand in these groups. We chose to allow the participants to choose their foot placement to reproduce a more natural movement. However, by normalizing foot placement and knee angle, the variability between groups would have been controlled and may have offered insight into the time to complete the task.

To our knowledge, this is the first study to evaluate the functional differences and similarities between the Ertl and Burgess amputation techniques. We found the Ertl group was able to perform the stand task more quickly than the Burgess group which indicates that there is a functional difference between these groups. Additionally, the Ertl amputated limb was able to produce more vertical GRF than the Burgess amputated limb which may have facilitated in performing the task more quickly. Surprisingly, we found the Ertl intact limb used a knee strategy compared to the hip strategy used by the Burgess group and the Ertl amputated limb. Although asymmetries persisted between the Ertl intact and amputated limbs, these results suggest that differences do exist between the Ertl and Burgess procedures which differs from previously published work. From this research, it is abundantly clear more research is warranted in this area.

CHAPTER V

GENERAL DISCUSSION AND CONCLUSIONS

The purpose of this dissertation was two-fold: to create a method to predict the inertial properties of the shank and foot segments of transtibial amputees (TTAs), and to assess functional differences between two below-knee amputation techniques. The findings of this dissertation support the hypothesis that functional differences exist between the Ertl and Burgess amputations. Specifically, the Ertl amputees appear to adopt strategies that are more like the intact limb and are able to perform the sit to stand task more quickly than the Burgess group. Further, the Ertl amputated limb was able to preferentially support more loads during the sit to stand task than the Burgess amputated limb. The developed inertial model was also able to produce inertial measures that are more similar to specific measures than the intact limb. This suggests that when specific measures are not available, the GENERAL model should be used.

Through the use of a prosthetic limb, persons with TTA are able to be successful community ambulators. This remains true even though passive elastic feet cannot produce power at the ankle at a level similar to that seen in the intact limb (Bateni & Olney, 2002; Sadeghi, Allard, et al., 2001; Silverman et al., 2008; Ventura et al., 2011; Zmitrewicz et al., 2007). Therefore, through three studies in this dissertation, a generic model was developed to predict inertial properties of the prosthetic side (Study 1), functional differences between Ertl and Burgess amputee groups during curb negotiation were identified (Study 2), and functional differences between Ertl and Burgess amputee groups during sit-to-stand were identified (Study 3).

In study 1, it was again reported that the body segment parameters of the amputated limb and prosthesis are significantly different than the intact limb. When including these outcomes with those in the literature, the mass of the prosthetic side is consistently 30-40% less, the center of mass location is 25-35% closer to the knee joint, and the moment of inertia is 50-60% less about a transverse axis through the knee joint compared with the intact limb (Lin-Chan et al., 2003b; Mattes et al., 2000; J. D. Smith et al., 2014). These inertial properties are important inputs into model simulations and inverse dynamic equations for biomechanical analysis. The most common practice in the literature to-date is to use the intact limb inertial properties for the amputated limb inertial properties. For those with the necessary equipment, which is very limited in number, a complex process based on reaction board and oscillation techniques is used to predict these subject-specific properties (Czerniecki et al., 1991; D. I. Miller, 1987; J. D. Smith et al., 2014). Use of the intact limb inertial properties has previously been shown (J. D. Smith et al., 2014) to produce inaccurate joint moments and powers during the swing phase of walking. Although reaction board and oscillation techniques are able to estimate the inertial properties of the amputated limb reasonably well, the process is lengthy and requires specialized equipment. Thus, there was a need to develop a general model which would produce similar results to the subject-specific measure of the oscillation technique. The development of such equations became the aim of Study 1.

It was hypothesized that outputs from the general model would produce results that would not differ from the subject-specific measures. Subject-specific measures were obtained from an initial population of individuals with TTA. The mass, COM, and MOI of the shank and foot of the amputated limb were determined using an oscillation rack and a reaction board (prosthetic limb) and geometric modeling (residual limb) (J. D. Smith et al., 2014). The means of these measures became the basis for the GENERAL model to predict these measures in the absence of an oscillation rack and reaction board. The model was validated using a separate unique population whose SPECIFIC measures were also measure using oscillation reaction board techniques. The results of the validation process suggested the GENERAL model estimated reasonably well the body segment parameters of the amputated limb. The GENERAL model predicted shank COM location and mass that more closely resembled subject-specific measures compared to INTACT measures. However, MOI between the SPECIFIC and INCACT models did not differ significantly. This was driven by the large variability in the SPECIFIC model likely due to the assumption that the prosthetic shank mass is 66% of the total prosthesis mass. This assumption may not hold true for all individuals based on prosthesis prescription. We have found the shank mass can range between ~55 - 75%. The large variations in MOI estimates are not likely to have a large impact on the joint moments (Challis & Kerwin, 1996). These results suggest that the GENERAL model successfully produces inertial property estimates of the prosthetic side that are much more consistent with subject-specific measures than assuming these measures are similar to those of intact segments. Therefore, it is suggested that when subject specific measures are not available, the GENERAL model should be used rather than intact limb inertial measures.

The second goal of study 1 aimed to understand how using the outputs from these models influence joint moments and powers. The inverse dynamics analyses in the second phase also illustrated that our GENERAL model did not result in significantly different joint moments at the knee and hip compared with the SPECIFIC measures during the swing phase of walking. This suggested that as inputs into an inverse dynamics analysis, the GENERAL measures were reasonable inputs for the COM location, MOI, and segment mass. Further, in addition to providing similar outputs as the SPECIFIC model, the GENERAL model is less time consuming (~ 5 min) than SPECIFIC measures (~ 30 min). Additionally, the use of the GENERAL model does not require specialized equipment to complete. Given the availability to predict amputated limb inertial properties and the ease and time saving benefits of the GENERAL model this study adds to the current body of literature and increases the accuracy of future research.

There also was a need to identify if surgical technique affects functional performance of persons with TTA (Studies 2 & 3). The two most common TTA techniques used by surgeons to amputate a limb are the modified Burgess and osteomyoplastic amputation (Ertl) techniques (Assal et al., 2005; Commuri et al., 2010; R. Dederich, 1983; Dionne et al., 2009; Ertl et al., 2013). The modified Burgess technique is more frequently used than the Ertl (Dionne et al., 2009). However, the Ertl procedure has the potential to lead to improved functional performance in persons with TTA due to the increased capability to bear loads on the end of the residual limb provided by the bone-bridge created in this technique. (Mongon et al., 2013).

In study 2, differences in mechanical work between the Ertl and Burgess groups were investigated during a curb negotiation task. The curb negotiation task is often reported as a challenging task by lower extremity amputees (Shumway-Cook et al., 2002); thus, it was investigated because it had greater potential to highlight functional differences between groups than less challenging task such as over-ground walking. The ground step and curb steps were analyzed separately for both the amputated and intact limbs of both groups. It was hypothesized that the Ertl amputated limb would behave in a manner similar to the Ertl intact limb due to the ability to bear greater loads on the end of the residual limb. On the GROUND step, both of the amputated limbs behaved similarly by producing less work at the ankle compared to the intact limbs. This was due to the decrease in ankle power production during the push-off phase. These results were similar to those seen during level over-ground walking in persons with TTA. With the loss of the intact ankle, the passive elastic prosthetic foot cannot actively generate mechanical power during push-off. However, on the CURB step, differences between the Ertl and Burgess groups emerged. The Ertl amputated limb produced similar net limb work as that observed in the intact limb of both groups. Although the ankle power remained diminished as seen with the GROUND step, the Ertl amputated limb produced significantly more hip work than the Burgess intact limb. This hip work was produced during the early stance phase. Further this hip power production was larger than both intact limbs and the Burgess amputated limb.

Changes from an ankle to a hip strategy has been noted in curb (Barnett et al., 2014) and stair (Aldridge et al., 2012; Alimusaj et al., 2009; Powers et al., 1997) negotiation in persons with TTA. Although it is tempting to draw similarities between curb and stair negotiation, the current study suggests significant differences exist between the two tasks. Most notably, the magnitudes of the joint powers are much smaller in curb negotiation than stair ambulation. Beyond biomechanical measures, curbs generally do not have a handrail for support whereas staircases are required to have a handrail for support (OSHA). Since persons with TTA report that curb negotiation is more challenging as stair ambulation (Larsson et al., 2009; Shumway-Cook et al., 2002), the lack of a handrail for support may be one contributing factor to this perception. There is evidence to support that the use of a handrail is actually beneficial from a mechanical perspective. When older adults were able to use a handrail in stair negotiation, joint moments decrease (Reeves et al., 2008) and dynamic stability increases (Reid et al., 2011). Thus, it appears the Ertl procedure lead to greater functional ability of individuals with TTA when negotiating a curb.

Study 3 aimed to identify differences in the functional performance of the sit-to-stand task in two groups of TTAs (Burgess and Ertl). From a clinical perspective, the Ertl group was able to perform the five time sit-to-stand task significantly more quickly than the Burgess group (9.33 ± 2.66 s vs 13.27 ± 2.83 s). Since both groups were similarly active and wore similar prostheses, this suggested that the Ertl group performed the task in a manner that differed from the Burgess group. The faster time may have been attributable to the Ertl group preferentially placing their feet closer to their body than the Burgess group (Khemlani et al., 1999). However, the Ertl amputated limb produced more force than the Burgess amputated limb, which was likely a strong reason why the Ertl group was able to perform the task faster. The higher load borne by the Ertl amputated limb does suggest that the Ertl group was able to preferentially increase the loads placed on the Ertl amputated limb.

Contrary to our hypothesis that the Ertl amputated limb would behave differently than the Burgess amputated limb, joint moments, powers, and work patterns were similar between the limbs. However, the Ertl intact limb behaved differently than the Burgess intact limb. The Ertl intact limb produced significantly larger peak knee joint powers, knee moments, and total knee work than the Burgess intact limb. Further, the Ertl intact limb produced significantly less hip angular momentum than the Burgess intact limb. These differences in joint mechanics suggest the limbs of the Burgess group and the amputated Ertl limb adopt a hip strategy whereas the Ertl intact limb adopts a knee strategy to accomplish the task. These trends show a clear shift in strategy adopted by the Ertl intact limb. Shifts from a knee to hip strategy have been seen in multiple patient populations along with an increase in trunk flexion during standing (Doorenbosch et al., 1994; Gross et al., 1998; Roebroeck et al., 1994). Van der heijden et al. (2009) found as the demands become too great, due to decreased muscular strength, the hip and ankle increase their contributions to the overall movement by ~60%. They suggested a decrease in knee extensor strength may decrease the ability to perform the sit-to-stand task without additional assistance from the hip and ankle. This explanation may provide a reason as to why the Ertl adopted a knee strategy. It may suggest that the Ertl intact leg knee extensors were stronger than the amputated limb and also compared to the Burgess intact limb. Thus, study 3 also suggests there is a functional advantage of the Ertl procedure over the Burgess procedure in individuals with TTA.

In summary, the current studies developed a model to predict the inertial properties of the shank and foot of persons with TTA and evaluated the functional differences in Ertl and Burgess amputees during curb negotiation and the sit-to-stand task. The developed inertial model was able to predict the shank and foot segment mass, COM location, and MOI more accurately than using the intact limb inertial properties. Used as inputs into inverse dynamics equations, the general model predictions produced joint moments which were also similar to the subject-specific measures. Thus, this model is a better predictor than the current method of using the intact limb inertial measures for the amputated limb. The second and third studies showed differences between the Ertl and Burgess amputated limbs in functional ability. The curb task showed that the Ertl amputated limb produced net limb work similar to the intact limbs of both groups on the curb step. This work was produced by the hip early in stance phase as a compensatory mechanism to help propel the body forward. The sit to stand task showed that the Ertl group was able to perform the task more quickly than the Burgess group. The faster performance time resulted in higher GRF in the Ertl amputated limb compared to the Burgess amputated limb which suggests the Ertl limb is able to bear higher loads during this task. While no other differences were found between the amputated limbs, the Ertl intact limb showed unexpected differences. Where the Burgess limbs and Ertl amputated limb adopted a hip strategy to complete the task, the Ertl intact limb adopted a knee strategy. This knee strategy is more similar to the way non-amputees complete the task. Both study 2 and 3 show functional advantage of the Ertl procedure over the Burgess procedure for these tasks and is, to our knowledge, the first study of its kind. These results lend support for the Ertl procedure over the Burgess since the functional abilities of these individuals is improved.

Conclusion

Study I determined that we were able to create a valid model to predict the inertial properties of the amputated shank and foot based on subject-specific measures. The inertial outputs from the GENERAL model did not differ statistically from the SPECIFIC measures. Moreover, when these inertial measures were used as inputs to the equations of motion, no significant differences were found between the SPECIFIC and GENERAL models. Thus, when SPECIFIC measures are not available, the GENERAL model should be applied to calculate the inertial properties of the amputated shank and foot.

For Studies II and III, the overall hypothesis that functional differences exist between the Ertl and Burgess amputees was supported. Ertl amputees were able to perform the five time sit-to-stand task more quickly than the Burgess group. This finding alone shows a clear functional difference between the two groups. Further, during the STAND task, the Ertl amputated limb was preferentially loaded more than the Burgess amputated limb. This lends support to the hypothesis that Ertl amputees are able to load the residual limb more than the traditional Burgess amputation.

Further, during curb negotiation, the Ertl amputated limb produced net limb work similar in magnitude to the Burgess and Ertl intact limbs. Again this supports the hypothesis that the Ertl amputated limb does act differently than the Burgess amputated limb which is more akin to an intact limb.

Future Directions

With the conclusion of this dissertations, there are several unanswered questions which are worthy of future work. Although insightful, inverse dynamic analysis does not provide clear insight into the motor control strategies adopted by these amputation techniques. In future studies we plan to evaluate the muscle activation patterns of the intact and amputated limbs which may provide more depth of understanding to the current level of knowledge. More specifically, the sit-to-stand task muscle activation patterns may help to explain more of the differences we noted between the Ertl intact limb and the Burgess intact limb. Changes in muscle recruitment patterns may show an increase in muscle activation in the quadriceps muscle group of the Ertl intact limb in comparison to increased muscle activity of the gluteal muscles of the Burgess intact limb.

Further, the differences between the Ertl intact limb and the Burgess intact limb may have also been driven by strength differences in the extensor muscle of the thigh. The investigation into the strength differences between the Burgess and Ertl groups may lend more insight into possible mechanistic differences between these two groups.

Lastly, we also did not evaluate trunk movement and how it contributes to the initiation of the sit-to-stand motion. It has been shown that the trunk position can change the upward initiation of the sit-to-stand movement. This change in strategy may also shift the demands from the knee to the hip in the Burgess group.

In addition to investigating muscle activation patterns during the curb step, the curb study should also include an evaluation of more kinematic variables including step length, stance time, and toe clearance. These variables have been linked to the incidence of falls in older adults. To gain even more insight beyond these measures, an analysis of dynamic stability between the two groups may also highlight differences in how these two groups negotiate the curb.

Finally, it is important to note that the choice of socket suspension system may be a contributing factor to the result of studies 2 and 3. Although prosthetists are aware of the Ertl procedure, none of the participants wore a socket designed to allow for total end-bearing of the residual limb. Each person (regardless of amputation technique) wore a different suspension type which included: lock and pin, vacuum, and elevated suction. Beyond the sit-to-stand and curb negotiation, it is important to investigate if a prosthetic socket which has been designed specifically for the Ertl amputation can help to reduce the inter-limb asymmetries during a variety of tasks.

REFERENCES

Administration, U. S. D. o. T. F. H. (2014). Designing Sidewalks and Trails for Access. Retrieved from Chapter 4 - Sidewalk Design Guidelines and Existing Practices website:http://www.fhwa.dot.gov/environment/bicycle_pedestrian/publications/sidewalks/chap4b.cfm

Agrawal, V., Gailey, R., Gaunaurd, I., Gailey, R., 3rd, & O'Toole, C. (2011). Weight distribution symmetry during the sit-to-stand movement of unilateral transtibial amputees. *Ergonomics, 54*(7), 656-664. doi: 10.1080/00140139.2011.586060

Agrawal, V., Gailey, R., O'Toole, C., Gaunaurd, I., & Dowell, T. (2009). Symmetry in external work (SEW): a novel method of quantifying gait differences between prosthetic feet. *Prosthetics and Orthotics International, 33*(2), 148-156. doi: 10.1080/03093640902777254

Aldridge, J. M., Sturdy, J. T., & Wilken, J. M. (2012). Stair ascent kinematics and kinetics with a powered lower leg system following transtibial amputation. *Gait and Posture, 36*(2), 291-295. doi: 10.1016/j.gaitpost.2012.03.013

Alimusaj, M., Fradet, L., Braatz, F., Gerner, H. J., & Wolf, S. I. (2009). Kinematics and kinetics with an adaptive ankle foot system during stair ambulation of transtibial amputees. *Gait and Posture, 30*(3), 356-363. doi: 10.1016/j.gaitpost.2009.06.009

Assal, M., Blanck, R., & Smith, D. G. (2005). Extended posterior flap for transtibial amputation. *Orthopedics, 28*(6), 542-546.

Au, S., Berniker, M., & Herr, H. (2008). Powered ankle-foot prosthesis to assist level-ground and stair-descent gaits. *Neural Netw, 21*(4), 654-666. doi: 10.1016/j.neunet.2008.03.006

Barclay-Goddard, R., Stevenson, T., Poluha, W., Moffatt, M. E., & Taback, S. P. (2004). Force platform feedback for standing balance training after stroke. *Cochrane Database Syst Rev*(4), CD004129.

Barnett, C., Polman, R., & Vanicek, N. (2014). Longitudinal changes in transtibial amputee gait characteristics when negotiating a change in surface height during continuous gait. *Clinical Biomechanics*.

Barth, D. G., Schumacher, L., & Thomas, S. S. (1992). Gait Analysis and Energy Cost of Below- Knee Amputees Wearing Six Different Prosthetic Feet. *J Prosthetics Orthotics, 4*(2), 63-75.

Bateni, H., & Olney, S. J. (2002). Kinematic and Kinetic Variations of Below-Knee Amputee Gait. *J Prosthetics Orthotics, 14*(1), 2-12.

Begg, R. K., & Sparrow, W. A. (2000). Gait characteristics of young and older individuals negotiating a raised surface: implications for the prevention of falls. *Journals of Gerontology. Series A, Biological Sciences and Medical Sciences, 55*(3), M147-154.

Boonstra, A. M., Fidler, V., & Eisma, W. H. (1993). Walking speed of normal subjects and amputees: aspects of validity of gait analysis. *Prosthetics and Orthotics International, 17*(2), 78-82.

Bureau, U. S. C. (2012). American FactFinder. from U.S. Department of Commerce http://factfinder2.census.gov

Burke, M. J., Roman, V., & Wright, V. (1978). Bone and joint changes in lower limb amputees. *Annals of the Rheumatic Diseases, 37*(3), 252-254.

Bussmann, J. B., Grootscholten, E. A., & Stam, H. J. (2004). Daily physical activity and heart rate response in people with a unilateral transtibial amputation for vascular disease. *Archives of Physical Medicine and Rehabilitation, 85*(2), 240-244.

Bussmann, J. B., Schrauwen, H. J., & Stam, H. J. (2008). Daily physical activity and heart rate response in people with a unilateral traumatic transtibial amputation. *Archives of Physical Medicine and Rehabilitation, 89*(3), 430-434. doi: 10.1016/j.apmr.2007.11.012

Challis, J. H. (1996). Accuracy of Human Limb Moment of Inertia Estimations and Their Influence on Resultant Joint Moments. [016508]. *Journal of Applied Biomechanics, 12*(4), 517-530.

Challis, J. H. (1999). Precision of the Estimation of Human Limb Inertial Parameters. *Journal of Applied Biomechanics, 15*, 418-428.

Challis, J. H., & Kerwin, D. G. (1996). Quantification of the uncertainties in resultant joint moments computed in a dynamic activity. *Journal of Sports Sciences, 14*(3), 219-231.

Chandler, R. F., Clauser, C.E., McConville, J.T., Reynolds, H.M. and Young, S.W. . (1975). Investigation of the inertial properties of the human body *Pamphlets DOT HS-801 430 and AMRL TR-74-137, Wright Patterson Air Force Base, OH*.

Cheng, C. K., Chen, H. H., Chen, C. S., Chen, C. L., & Chen, C. Y. (2000). Segment inertial properties of Chinese adults determined from magnetic resonance imaging. *Clinical Biomechanics (Bristol, Avon), 15*(8), 559-566.

Clauser, C. E., McConville, J. T., & Young, J. W. (1969). Weight, Volume, and Center of Mass of Segments of the Human Body. *AMRL Technical Report. Wright-Patterson Air Force Base, OH*, 60-70.

Commuri, S., Day, J., Dionne, C. P., & Ertl, W. J. (2010). Assessment of pressures within the prosthetic socket of a person with osteomyoplastic amputation during varied walking tasks. *Journal of Prosthetics and Orthotics, 22*(2), 127.

Council, N. S. (2011). *lnjury Facts* Vol. 2011 Edition.

Crosbie, J., & Ko, V. (2000). Changes in the temporal and distance parameters of gait evoked by negotiation of curbs. *Aust J Physiother, 46*(2), 103-112.

Czerniecki, J. M., Gitter, A., & Munro, C. (1991). Joint moment and muscle power output characteristics of below knee amputees during running: the influence of energy storing prosthetic feet. *Journal of Biomechanics, 24*(1), 63-75.

Dall, P. M., & Kerr, A. (2010). Frequency of the sit to stand task: An observational study of free-living adults. *Applied Ergonomics, 41*(1), 58-61.

de Leva, P. (1996). Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters. *Journal of Biomechanics, 29*(9), 1223-1230.

DeCoster, T. A., & Homedan, S. (2006). Amputation osteoplasty. *Iowa Orthopaedic Journal, 26*, 54-59.

Dederich, R. (1963). Plastic treatment of the muscles and bone in amputation surgery. *Journal of Bone & Joint Surgery, British Volume, 45*(1), 60-66.

Dederich, R. (1983). [Indications for amputation as well as stump care in bone infection of the lower extremity]. *Orthopade, 12*(4), 235-255.

Dempster, W. (1955). Space requirements of the seated operator. *WADC Technical Report, Wright Patterson Air Force Base, OH*, 55–159.

Dillingham, T. R., Pezzin, L. E., & MacKenzie, E. J. (2002). Limb amputation and limb deficiency: epidemiology and recent trends in the United States. *Southern Medical Journal, 95*(8), 875-883.

Dionne, C., Ertl, W., & Day, J. (2009). Rehabilitation for Those With Transtibial Osteomyoplastic Amputation. *Journal of Prosthetics and Orthotics, 21*(1), 64.

Donn, J. M., Porter, D., & Roberts, V. C. (1989). The effect of footwear mass on the gait patterns of unilateral below-knee amputees. *Prosthetics and Orthotics International, 13*(3), 140-144.

Doorenbosch, C. A., Harlaar, J., Roebroeck, M. E., & Lankhorst, G. J. (1994). Two strategies of transferring from sit-to-stand; the activation of monoarticular and biarticular muscles. *Journal of Biomechanics, 27*(11), 1299-1307.

Dougherty, P. J. (2001). Transtibial amputees from the Vietnam War. Twenty-eight-year follow-up. *Journal of Bone & Joint Surgery - American Volume, 83-A*(3), 383-389.

Drillis, R., Contini, R., & Bluestein, M. (1964). Body Segment Parameters; a Survey of Measurement Techniques. *Artificial Limbs, 25*, 44-66.

Dudek, N. L., DeHaan, M. N., & Marks, M. B. (2003). Bone overgrowth in the adult traumatic amputee. *American Journal of Physical Medicine and Rehabilitation, 82*(11), 897-900. doi: 10.1097/01.PHM.0000087459.94599.2D

Dudek, N. L., Marks, M. B., Marshall, S. C., & Chardon, J. P. (2005). Dermatologic conditions associated with use of a lower-extremity prosthesis. *Archives of Physical Medicine and Rehabilitation, 86*(4), 659-663. doi: 10.1016/j.apmr.2004.09.003

Durkin, J. L., & Dowling, J. J. (2003). Analysis of body segment parameter differences between four human populations and the estimation errors of four popular mathematical models. *Journal of Biomechanical Engineering, 125*(4), 515-522.

Durkin, J. L., Dowling, J. J., & Andrews, D. M. (2002). The measurement of body segment inertial parameters using dual energy X-ray absorptiometry. *Journal of Biomechanics, 35*(12), 1575-1580. doi: S0021929002002270 [pii]

Engsberg, J. R., Lee, A. G., Tedford, K. G., & Harder, J. A. (1993). Normative ground reaction force data for able-bodied and below-knee-amputee children during walking. *Journal of Pediatric Orthopedics, 13*(2), 169-173.

Ertl, J., Ertl, J., Ertl, W., & Stokosa, J. (2013, February 15, 2013). The Ertl osteomyoplastic transtibial amputation reconstruction description of technique and long term results.   Retrieved February 15, 2013, from http://ertlreconstruction.com/documents/108.html

Farquhar, S. J., Kaufman, K. R., & Snyder-Mackler, L. (2009). Sit-to-Stand 3 months after unilateral total knee arthroplasty: Comparison of self-selected and constrained conditions. *Gait and Posture, 30*(2), 187-191. doi: http://dx.doi.org/10.1016/j.gaitpost.2009.04.007

Ferris, A. E., Aldridge, J. M., Rabago, C. A., & Wilken, J. M. (2012). Evaluation of a powered ankle-foot prosthetic system during walking. *Archives of Physical*

Medicine and Rehabilitation, 93(11), 1911-1918. doi: 10.1016/j.apmr.2012.06.009

Firth, G. B., Masquijo, J. J., & Kontio, K. (2011). Transtibial Ertl amputation for children and adolescents: a case series and literature review. *J Child Orthop*, *5*(5), 357-362. doi: 10.1007/s11832-011-0364-0

Fleckenstein, S. J., Kirby, R. L., & MacLeod, D. A. (1988). Effect of limited knee-flexion range on peak hip moments of force while transferring from sitting to standing. *Journal of Biomechanics, 21*(11), 915-918.

Fradet, L., Alimusaj, M., Braatz, F., & Wolf, S. I. (2010). Biomechanical analysis of ramp ambulation of transtibial amputees with an adaptive ankle foot system. *Gait and Posture*, *32*(2), 191-198. doi: 10.1016/j.gaitpost.2010.04.011

Fridman, A., Ona, I., & Isakov, E. (2003). The influence of prosthetic foot alignment on trans-tibial amputee gait. *Prosthetics and Orthotics International*, *27*(1), 17-22.

Gailey, R. S., Wenger, M. A., Raya, M., Kirk, N., Erbs, K., Spyropoulos, P., & Nash, M. S. (1994). Energy expenditure of trans-tibial amputees during ambulation at self-selected pace. *Prosthetics and Orthotics International, 18*(2), 84-91.

Galli, M., Cimolin, V., Crivellini, M., & Campanini, I. (2008). Quantitative analysis of sit to stand movement: Experimental set-up definition and application to healthy and hemiplegic adults. *Gait and Posture, 28*(1), 80-85. doi: http://dx.doi.org/10.1016/j.gaitpost.2007.10.003

Ganley, K. J., & Powers, C. M. (2005). Gait kinematics and kinetics of 7-year-old children: a comparison to adults using age-specific anthropometric data. *Gait and Posture*, *21*(2), 141-145.

Gross, M. M., Stevenson, P. J., Charette, S. L., Pyka, G., & Marcus, R. (1998). Effect of muscle strength and movement speed on the biomechanics of rising from a chair in healthy elderly and young women. *Gait and Posture, 8*(3), 175-185. doi: http://dx.doi.org/10.1016/S0966-6362(98)00033-2

Gurkan, U. A., & Akkus, O. (2008). The mechanical environment of bone marrow: a review. *Annals of Biomedical Engineering, 36*(12), 1978-1991. doi: 10.1007/s10439-008-9577-x

Hanavan, E. P., Jr. (1964). A Mathematical Model of the Human Body. Amrl-Tr-64-102. *AMRL-TR, 18*, 1-149.

Hansen-Leth, C. (1979). The vascularization in the amputation stumps of rabbits. A microangiographic study. *Acta Orthopaedica Scandinavica, 50*(4), 399-406.

Hantula, D. A., Bragger, J. L. D., & Rajala, A. K. (2001). Slips and falls in stores and malls: Implications for community-based injury prevention. *Journal of Prevention & Intervention in the Community, 22*(1), 67-79.

Hatze, H. (1975). A new method for the simultaneous measurement of the moment of inertia, the damping coefficient and the location of the centre of mass of a body segmentin situ. *European Journal of Applied Physiology and Occupational Physiology, 34*(1), 217-226.

Hatze, H. (1980). A mathematical model for the computational determination of parameter values of anthropomorphic segments. *Journal of Biomechanics, 13*(10), 833-843. doi: 0021-9290(80)90171-2 [pii]

Herr, H. M., & Grabowski, A. M. (2012). Bionic ankle-foot prosthesis normalizes walking gait for persons with leg amputation. *Proc Biol Sci, 279*(1728), 457-464. doi: 10.1098/rspb.2011.1194

Hillery, S. C., Wallace, E. S., McIlhagger, R., & Watson, P. (1997). The effect of changing the inertia of a trans-tibial dynamic elastic response prosthesis on the kinematics and ground reaction force patterns. *Prosthetics and Orthotics International, 21*(2), 114-123.

Hinrichs, R. N. (1985). Regression equations to predict segmental moments of inertia from anthropometric measurements: an extension of the data of Chandler et al. (1975). *Journal of Biomechanics, 18*(8), 621-624. doi: 0021-9290(85)90016-8 [pii]

Hinrichs, R. N. (1990). Adjustments to the segment center of mass proportions of Clauser et al. (1969). *Journal of Biomechanics, 23*(9), 949-951. doi: 0021-9290(90)90361-6 [pii]

Hitt, J. K., Sugar, T. G., Holgate, M., & Bellman, R. (2010). An Active Foot-Ankle Prosthesis With Biomechanical Energy Regeneration. *Journal of Medical Devices-Transactions of the Asme, 4*(1).

Huang, S., & Ferris, D. P. (2012). Muscle activation patterns during walking from transtibial amputees recorded within the residual limb-prosthetic interface. *Journal of NeuroEngineering and Rehabilitation, 9*(55), 1-16. doi: Doi 10.1186/1743-0003-9-55

Hughes, M., Weiner, D., Schenkman, M., Long, R., & Studenski, S. (1994). Chair rise strategies in the elderly. *Clinical Biomechanics*, *9*(3), 187-192.

Hurley, G. R., McKenney, R., Robinson, M., Zadravec, M., & Pierrynowski, M. R. (1990). The role of the contralateral limb in below-knee amputee gait. *Prosthetics and Orthotics International*, *14*(1), 33-42.

Hussainy, H., Goesling, T., Datta, D., & Saleh, M. (2004). Osteomyoplasty for transtibial amputation: a case report and review of the literature. *Journal of Prosthetics and Orthotics*, *16*(1), 2-5.

Isakov, E., Burger, H., Krajnik, J., Gregoric, M., & Marincek, C. (1996). Influence of speed on gait parameters and on symmetry in trans-tibial amputees. *Prosthetics and Orthotics International*, *20*(3), 153-158.

Isakov, E., Burger, H., Krajnik, J., Gregoric, M., & Marincek, C. (1997). Double-limb support and step-length asymmetry in below-knee amputees. *Scandinavian Journal of Rehabilitation Medicine*, *29*(2), 75-79.

Isakov, E., Burger, H., Krajnik, J., Gregoric, M., & Marincek, C. (2001). Knee muscle activity during ambulation of trans-tibial amputees. *J Rehabil Med*, *33*(5), 196-199.

Isakov, E., Keren, O., & Benjuya, N. (2000). Trans-tibial amputee gait: time-distance parameters and EMG activity. *Prosthetics and Orthotics International*, *24*(3), 216-220.

Isakov, E., Mizrahi, J., Susak, Z., & Onna, I. (1992). A Swedish knee-cage for stabilizing short below-knee stumps. *Prosthetics and Orthotics International*, *16*(2), 114-117.

Jensen, R. K. (1978). Estimation of the biomechanical properties of three body types using a photogrammetric method. *Journal of Biomechanics, 11*(8-9), 349-358. doi: 0021-9290(78)90069-6 [pii]

Kawagoe, S., Tajima, N., & Chosa, E. (2000). Biomechanical analysis of effects of foot placement with varying chair height on the motion of standing up. *Journal of Orthopaedic Science, 5*(2), 124-133.

Keeling, J. J., Shawen, S. B., Forsberg, J. A., Kirk, K. L., Hsu, J. R., Gwinn, D. E., & Potter, B. K. (2013). Comparison of Functional Outcomes Following Bridge Synostosis with Non-Bone-Bridging Transtibial Combat-Related Amputations. *The Journal of Bone & Joint Surgery, 95*(10), 888-893. doi: 10.2106/jbjs.l.00423

Kerr, K. M., White, J. A., Barr, D. A., & Mollan, R. A. (1997). Analysis of the sit-stand-sit movement cycle in normal subjects. *Clinical Biomechanics, 12*(4), 236-245.

Khemlani, M. M., Carr, J. H., & Crosbie, W. J. (1999). Muscle synergies and joint linkages in sit-to-stand under two initial foot positions. *Clinical Biomechanics, 14*(4), 236-246.

Killey, B., & Watt, E. (2006). The effect of extra walking on the mobility, independence and exercise self-efficacy of elderly hospital in-patients: a pilot study. *Contemporary Nurse, 22*(1), 120-133.

Kim, H.-D. (2009). The Influence of Aging on the Center of Pressure Trajectory: a Comparison of Crossing an Obstacle and Stepping onto a Curb from a Position of Quiet Stance. *Journal of Physical Therapy Science, 21*(2), 183-188.

Kulkarni, J., Toole, C., Hirons, R., Wright, S., & Morris, J. (1996). Falls in patients with lower limb amputations: prevalence and contributing factors. *Physiother*, *82*(2), 130-136.

Larsson, B., Johannesson, A., Andersson, I. H., & Atroshi, I. (2009). The Locomotor Capabilities Index; validity and reliability of the Swedish version in adults with lower limb amputation. *Health Qual Life Outcomes, 7*, 44. doi: 10.1186/1477-7525-7-44

Lee, H.-J., & Chou, L.-S. (2006). Detection of Gait Instability Using the Center of Mass and Center of Pressure Inclination Angles. *Archives of Physical Medicine and Rehabilitation, 87*(4), 569-575.

Lemaire, E. D., & Fisher, F. R. (1994). Osteoarthritis and elderly amputee gait. *Archives of Physical Medicine and Rehabilitation, 75*(10), 1094-1099.

Lin-Chan, S. J., Nielsen, D. H., Yack, H. J., Hsu, M. J., & Shurr, D. G. (2003a). The effects of added prosthetic mass on physiologic responses and stride frequency during multiple speeds of walking in persons with transtibial amputation. *Archives of Physical Medicine and Rehabilitation, 84*(12), 1865-1871.

Lin-Chan, S. J., Nielsen, D. H., Yack, H. J., Hsu, M. J., & Shurr, D. G. (2003b). The effects of added prosthetic mass on physiologic responses and stride frequency during multiple speeds of walking in persons with transtibial amputation. *Archives of Physical Medicine and Rehabilitation, 84*(12), 1865-1871.

Loon, H. E. (1962). Below-knee amputation surgery. *Artificial Limbs, 6*(1), 86-99.

Loverro, K. L., Mueske, N. M., & Hamel, K. A. (2013). Location of minimum foot clearance on the shoe and with respect to the obstacle changes with locomotor task. *Journal of Biomechanics, 46*(11), 1842-1850.

Lythgo, N., Begg, R., & Best, R. (2007). Stepping responses made by elderly and young female adults to approach and accommodate known surface height changes. *Gait and Posture, 26*(1), 82-89. doi: DOI 10.1016/j.gaitpost.2006.07.006

MacKenzie, E. J., Jones, A. S., Bosse, M. J., Castillo, R. C., Pollak, A. N., Webb, L. X., Burgess, A. R. (2007). Health-care costs associated with amputation or reconstruction of a limb-threatening injury. *Journal of Bone and Joint Surgery, 89*(8), 1685-1692. doi: 10.2106/JBJS.F.01350

Martin, P. E., Mungiole, M., Marzke, M. W., & Longhill, J. M. (1989). The use of magnetic resonance imaging for measuring segment inertial properties. *Journal of Biomechanics, 22*(4), 367-376.

Mattes, S. J., Martin, P. E., & Royer, T. D. (2000). Walking symmetry and energy cost in persons with unilateral transtibial amputations: matching prosthetic and intact limb inertial properties. *Archives of Physical Medicine and Rehabilitation, 81*(5), 561-568.

McFadyen, B. J., & Prince, F. (2002). Avoidance and accommodation of surface height changes by healthy, community-dwelling, young, and elderly men. *Journals of Gerontology. Series A, Biological Sciences and Medical Sciences, 57*(4), B166-174.

Melzer, I., Yekutiel, M., & Sukenik, S. (2001). Comparative study of osteoarthritis of the contralateral knee joint of male amputees who do and do not play volleyball. *Journal of Rheumatology, 28*(1), 169-172.

Mena, D., Mansour, J. M., & Simon, S. R. (1981). Analysis and synthesis of human swing leg motion during gait and its clinical applications. *Journal of Biomechanics, 14*(12), 823-832.

Miller, D. I. (1987). Resultant lower extremity joint moments in below-knee amputees during running stance. *Journal of Biomechanics, 20*(5), 529-541.

Miller, W., Speechley, M., & Deathe, B. (2001). The prevalence and risk factors of falling and fear of falling among lower extremity amputees. *Archives of Physical Medicine and Rehabilitation, 82*(8), 1031-1037. doi: 10.1053/apmr.2001.24295

Mizner, R. L., & Snyder-Mackler, L. (2005). Altered loading during walking and sit-to-stand is affected by quadriceps weakness after total knee arthroplasty. *Journal of Orthopaedic Research, 23*(5), 1083-1090. doi: 10.1016/j.orthres.2005.01.021

Molen, N. H. (1973). Energy/speed relation of below-knee amputees walking on motor driven treadmill. *Int Z Angrew Physiol, 31*, 173-185.

Mongon, M. L., Piva, F. A., Mistro Neto, S., Carvalho, J. A., Belangero, W. D., & Livani, B. (2013). Cortical tibial osteoperiosteal flap technique to achieve bony bridge in transtibial amputation: experience in nine adult patients. *Strategies Trauma Limb Reconstr, 8*(1), 37-42. doi: 10.1007/s11751-013-0152-0

Moosabhoy, M. A., & Gard, S. A. (2006). Methodology for determining the sensitivity of swing leg toe clearance and leg length to swing leg joint angles during gait. *Gait and Posture*.

Mungiole, M., & Martin, P. E. (1990). Estimating segment inertial properties: comparison of magnetic resonance imaging with existing methods. *Journal of Biomechanics, 23*(10), 1039-1046.

Murray, M., & Clarkson, B. (1966). The vertical pathways of the foot during level walking. I. Range of variability in normal men. *Physical Therapy, 46*(6), 585-589.

Nolan, L., Wit, A., Dudzinski, K., Lees, A., Lake, M., & Wychowanski, M. (2003). Adjustments in gait symmetry with walking speed in trans-femoral and trans-tibial amputees. *Gait and Posture, 17*(2), 142-151. doi: 10.1016/s0966-6362(02)00066-8

Norrish, K., O'Reilly, K., Whitney, K., Campbell, R., & MacDonell, K. (2009). A step in the right direction: how does curb negotiation affect dynamic stability?

OSHA, U. D. o. L. Stiarways and Ladders. *Safety and Health Regulations for Construction.* Retrieved 10/22, 2015

Ozyurek, S., Demirbuken, I., & Angin, S. (2013). Altered movement strategies in sit-to-stand task in persons with transtibial amputation. *Prosthetics and Orthotics International*, 0309364613497742. doi: 10.1177/0309364613497742

Ozyurek, S., Demirbuken, I., & Angin, S. (2014). Altered movement strategies in sit-to-stand task in persons with transtibial amputation. *Prosthetics and Orthotics International, 38*(4), 303-309. doi: 10.1177/0309364613497742

Papa, E., & Cappozzo, A. (2000). Sit-to-stand motor strategies investigated in able-bodied young and elderly subjects. *Journal of Biomechanics, 33*(9), 1113-1122.

Patla, A. E., Prentice, S. D., Rietdyk, S., Allard, F., & Martin, C. (1999). What guides the selection of alternate foot placement during locomotion in humans. *Experimental Brain Research, 128*(4), 441-450.

Patla, A. E., & Rietdyk, S. (1993). Visual Control of Limb Trajectory over Obstacles during Locomotion - Effect of Obstacle Height and Width. *Gait and Posture, 1*(1), 45-60. doi: Doi 10.1016/0966-6362(93)90042-Y

Patla, A. E., & Shumway-Cook, A. (1999). Dimensions of mobility: Defining the complexity and difficulty associated with community mobility. *JOURNAL OF AGING AND PHYSICAL ACTIVITY, 7*(1), 7-19.

Pavol, M., Owings, T., Foley, K., & Grabiner, M. (1999). The sex and age of older adults influence the outcome of induced trips. *Journal of Gerontology, Series A: Biological Sciences and Medical Sciences*.

Pearsall, D., Reid, J., & Ross, R. (1994). Inertial properties of the human trunk of males determined from magnetic resonance imaging. *Annals of Biomedical Engineering, 22*(6), 692-706.

Pilegard, H. K., Rorbaek Madsen, M., Hansen-Leth, C., & Terp, A. (1985). Muscle blood flow after amputation. Increased flow with medullary plugging. *Acta Orthopaedica Scandinavica, 56*(6), 500-502.

Pinzur, M. S., Beck, J., Himes, R., & Callaci, J. (2008). Distal tibiofibular bone-bridging in transtibial amputation. *The Journal of Bone & Joint Surgery, 90*(12), 2682-2687.

Potter, B. K., Burns, T. C., Lacap, A. P., Granville, R. R., & Gajewski, D. A. (2007). Heterotopic ossification following traumatic and combat-related amputations. Prevalence, risk factors, and preliminary results of excision. *Journal of Bone and Joint Surgery*, *89*(3), 476-486. doi: 10.2106/JBJS.F.00412

Powers, C. M., Boyd, L. A., Torburn, L., & Perry, J. (1997). Stair ambulation in persons with transtibial amputation: an analysis of the Seattle LightFoot. *Journal of Rehabilitation Research and Development, 34*(1), 9-18.

Powers, C. M., Rao, S., & Perry, J. (1998). Knee kinetics in trans-tibial amputee gait. *Gait and Posture, 8*(1), 1-7.

Rao, S. S., Boyd, L. A., Mulroy, S. J., Bontrager, E. L., Gronley, J. K., & Perry, J. (1998). Segment velocities in normal and transtibial amputees: prosthetic design implications. *IEEE Transactions on Rehabilitation Engineering*, *6*(2), 219-226.

Reeves, N. D., Spanjaard, M., Mohagheghi, A. A., Baltzopoulos, V., & Maganaris, C. N. (2008). Influence of light handrail use on the biomechanics of stair negotiation in old age. *Gait and Posture, 28*(2), 327-336. doi: http://dx.doi.org/10.1016/j.gaitpost.2008.01.014

Reid, S. M., Novak, A. C., Brouwer, B., & Costigan, P. A. (2011). Relationship between stair ambulation with and without a handrail and centre of pressure velocities during stair ascent and descent. *Gait and Posture*, *34*(4), 529-532. doi: http://dx.doi.org/10.1016/j.gaitpost.2011.07.008

Roebroeck, M. E., Doorenbosch, C. A., Harlaar, J., Jacobs, R., & Lankhorst, G. J. (1994). Biomechanics and muscular activity during sit-to-stand transfer. *Clinical Biomechanics (Bristol, Avon), 9*(4), 235-244. doi: 10.1016/0268-0033(94)90004-3

Roy, G., Nadeau, S., Gravel, D., Malouin, F., McFadyen, B. J., & Piotte, F. (2006). The effect of foot position and chair height on the asymmetry of vertical forces during sit-to-stand and stand-to-sit tasks in individuals with hemiparesis. *Clinical Biomechanics (Bristol, Avon), 21*(6), 585-593. doi: 10.1016/j.clinbiomech.2006.01.007

Royer, T. D., & Wasilewski, C. A. (2006). Hip and knee frontal plane moments in persons with unilateral, trans-tibial amputation. *Gait and Posture, 23*(3), 303-306. doi: 10.1016/j.gaitpost.2005.04.003

Sadeghi, H., Allard, P., & Duhaime, P. M. (2001). Muscle power compensatory mechanisms in below-knee amputee gait. *American Journal of Physical Medicine and Rehabilitation, 80*(1), 25-32.

Sadeghi, H., Sadeghi, S., Prince, F., Allard, P., Labelle, H., & Vaughan, C. L. (2001). Functional roles of ankle and hip sagittal muscle moments in able-bodied gait. *Clinical Biomechanics, 16*(8), 688-695.

Sanderson, D. J., & Martin, P. E. (1997). Lower extremity kinematic and kinetic adaptations in unilateral below-knee amputees during walking. *Gait and Posture, 6*(2), 126-136.

Savelberg, H., Fastenau, A., Willems, P., & Meijer, K. (2007). The load/capacity ratio affects the sit-to-stand movement strategy. *Clinical Biomechanics, 22*(7), 805-812.

Schmalz, T., Blumentritt, S., & Marx, B. (2007). Biomechanical analysis of stair ambulation in lower limb amputees. *Gait and Posture, 25*(2), 267-278.

Schmalz, T., Blumentritt, S., & Reimers, C. D. (2001). Selective thigh muscle atrophy in transtibial amputees: an ultrasonographic study. *Archives of Orthopaedic and Trauma Surgery, 121*, 307-312.

Schofield, J. S., Parent, E., Lewicke, J., Carey, J. P., El-Rich, M., & Adeeb, S. (2014). Leg Dominance May Not Be a Predictor of Asymmetry in Peak Joint Moments and Ground Reaction Forces During Sit-to-Stand Movements. *Journal of Applied Biomechanics, 30*(1), 179-183.

Schofield, J. S., Parent, E. C., Lewicke, J., Carey, J. P., El-Rich, M., & Adeeb, S. (2013). Characterizing asymmetry across the whole sit to stand movement in healthy participants. *Journal of Biomechanics, 46*(15), 2730-2735. doi: 10.1016/j.jbiomech.2013.07.046

Schulz, B. W. (2011). Minimum toe clearance adaptations to floor surface irregularity and gait speed. *Journal of Biomechanics, 44*(7), 1277-1284. doi: 10.1016/j.jbiomech.2011.02.010

Shepherd, R., & Koh, H. (1996). Some biomechanical consequences of varying foot placement in sit-to-stand in young women. *Scandinavian Journal of Rehabilitation Medicine, 28*(2), 79-88.

Shumway-Cook, A., Patla, A. E., Stewart, A., Ferrucci, L., Ciol, M. A., & Guralnik, J. M. (2002). Environmental demands associated with community mobility in older adults with and without mobility disabilities. *Physical Therapy*, *82*(7), 670-681.

Sibella, F., Galli, M., Romei, M., Montesano, A., & Crivellini, M. (2003). Biomechanical analysis of sit-to-stand movement in normal and obese subjects. *Clinical Biomechanics, 18*(8), 745-750.

Silverman, A. K., Fey, N. P., Portillo, A., Walden, J. G., Bosker, G., & Neptune, R. R. (2008). Compensatory mechanisms in below-knee amputee gait in response to increasing steady-state walking speeds. *Gait and Posture, 28*(4), 602-609. doi: 10.1016/j.gaitpost.2008.04.005

Slajpah, S., Kamnik, R., Burger, H., Bajd, T., & Munih, M. (2013). Asymmetry in sit-to-stand movement in patients following transtibial amputation and healthy individuals. *International Journal of Rehabilitation Research, 36*(3), 275-283. doi: 10.1097/MRR.0b013e3283606235

Smith, D. G., & Fergason, J. R. (1999). Transtibial amputations. *Clin Orthop Relat Res*(361), 108-115.

Smith, J. D., Ferris, A. E., Heise, G. D., Hinrichs, R. N., & Martin, P. E. (2014). Oscillation and reaction board techniques for estimating inertial properties of a below-knee prosthesis. *J Vis Exp*(87), e50977-e50977. doi: 10.3791/50977

Smith, J. D., & Martin, P. E. (2011). Short and Longer Term Changes In Amputee Walking Patterns Due To Increased Prosthesis Inertia. *J Prosthet Orthot, 23*(3), 114-123.

Smith, J. D., & Martin, P. E. (2013). Effects of prosthetic mass distribution on metabolic costs and walking symmetry. *J Appl Biomech, 29*(3), 317-328. doi: 2011-0097 [pii]

Stevens, C., Bojsen-Möller, F., & Soames, R. (1989). The influence of initial posture on the sit-to-stand movement. *European Journal of Applied Physiology and Occupational Physiology, 58*(7), 687-692.

Talis, V., Grishin, A., Solopova, I., Oskanyan, T., Belenky, V., & Ivanenko, Y. (2008). Asymmetric leg loading during sit-to-stand, walking and quiet standing in patients after unilateral total hip replacement surgery. *Clinical Biomechanics, 23*(4), 424-433.

Taylor, B. C., French, B., Poka, A., Blint, A., & Mehta, S. (2010). Osteomyoplastic and traditional transtibial amputations in the trauma patient: perioperative comparisons and outcomes. *Orthopedics, 33*(6), 390. doi: 10.3928/01477447-20100429-14

Taylor, B. C., & Poka, A. (2011). Osteomyoplastic transtibial amputation: technique and tips. *J Orthop Surg Res, 6*(1), 13. doi: 10.1186/1749-799X-6-13

Torburn, L., Powers, C. M., Guiterrez, R., & Perry, J. (1995). Energy expenditure during ambulation in dysvascular and traumatic below-knee amputees: a comparison of five prosthetic feet. *Journal of Rehabilitation Research and Development, 32*(2), 111-119.

Torburn, L., Schweiger, G. P., Perry, J., & Powers, C. M. (1994). Below-knee amputee gait in stair ambulation. A comparison of stride characteristics using five different prosthetic feet. *Clin Orthop Relat Res*(303), 185-192.

Tsai, C. S., & Mansour, J. M. (1986). Swing phase simulation and design of above knee prostheses. *Journal of Biomechanical Engineering, 108*(1), 65-72.

Turner, C. H., & Robling, A. G. (2004). Mechanical loading and bone formation. *BoneKEy-Osteovision, 1*(9), 15-23.

Van der heijden, M. M. P., Meijer, K., Willems, P. J. B., & Savelberg, H. H. C. M. (2009). Muscles limiting the sit-to-stand movement: An experimental simulation of muscle weakness. *Gait and Posture, 30*(1), 110-114. doi: http://dx.doi.org/10.1016/j.gaitpost.2009.04.002

van Dieen, J. H., Spanjaard, M., Konemann, R., Bron, L., & Pijnappels, M. (2007). Balance control in stepping down expected and unexpected level changes. *Journal of Biomechanics, 40*(16), 3641-3649.

Vanicek, N., Strike, S., McNaughton, L., & Polman, R. (2009). Gait patterns in transtibial amputee fallers vs. non-fallers: biomechanical differences during level walking. *Gait and Posture, 29*(3), 415-420. doi: 10.1016/j.gaitpost.2008.10.062

Ventura, J. D., Klute, G. K., & Neptune, R. R. (2011). The effect of prosthetic ankle energy storage and return properties on muscle activity in below-knee amputee walking. *Gait and Posture, 33*(2), 220-226. doi: 10.1016/j.gaitpost.2010.11.009

Versluys, R., Lenaerts, G., Van Damme, M., Jonkers, I., Desomer, A., Vanderborght, B., . . . Lefeber, D. (2009). Successful preliminary walking experiments on a transtibial amputee fitted with a powered prosthesis. *Prosthetics and Orthotics International, 33*(4), 368-377. doi: 10.3109/03093640902984587

Wang, L., Fritton, S. P., Weinbaum, S., & Cowin, S. C. (2003). On bone adaptation due to venous stasis. *Journal of Biomechanics, 36*(10), 1439-1451.

Wang, Y., & Watanabe, K. (2008). The relationship between obstacle height and center of pressure velocity during obstacle crossing. *Gait and Posture, 27*(1), 172-175.

Waters, R. L., & Mulroy, S. (1999). The energy expenditure of normal and pathologic gait. *Gait and Posture, 9*(3), 207-231.

Welch, R. D., Johnston, C. E., 2nd, Waldron, M. J., & Poteet, B. (1993). Bone changes associated with intraosseous hypertension in the caprine tibia. *Journal of Bone and Joint Surgery, 75*(1), 53-60.

Winter, D. A. (1992). Foot trajectory in human gait: a precise and multifactorial motor control task. *Physical Therapy, 72*(1), 45-53; discussion 54-46.

Winter, D. A., & Sienko, S. E. (1988). Biomechanics of below-knee amputee gait. *Journal of Biomechanics, 21*(5), 361-367.

Yack, H. J., Nielsen, D. H., & Shurp, D. G. (1999). Kinetic Patterns During Stair Ascent in Patients with Transtibial Amputations Using Three Different Prostheses. *JPO: Journal of Prosthetics and Orthotics, 11*(3), 57-62.

Yeadon, M. R., & Morlock, M. (1989). The appropriate use of regression equations for the estimation of segmental inertia parameters. *Journal of Biomechanics, 22*(6), 683-689.

Zatsiorsky, V. M. a. S., V. N. (1983). The mass and inertia characteristics of the main segments of the human body. *Biomechanics VIII-B*, 1152-1159.

Ziegler-Graham, K., MacKenzie, E. J., Ephraim, P. L., Travison, T. G., & Brookmeyer, R. (2008). Estimating the prevalence of limb loss in the United States: 2005 to 2050. *Archives of Physical Medicine and Rehabilitation, 89*(3), 422-429. doi: 10.1016/j.apmr.2007.11.005

Zmitrewicz, R. J., Neptune, R. R., & Sasaki, K. (2007). Mechanical energetic contributions from individual muscles and elastic prosthetic feet during symmetric unilateral transtibial amputee walking: a theoretical study. *Journal of Biomechanics, 40*(8), 1824-1831. doi: 10.1016/j.jbiomech.2006.07.009

APPENDIX A

REVIEW OF LITERATURE

Introduction

There are over 20 million people in the US living with disabilities which limit ambulation (Bureau, 2012). An estimated 2 million of these Americans live with limb loss resulting from dysvascular disease, trauma, or cancer (Ziegler-Graham et al., 2008). Roughly 25% of these individuals have undergone a transtibial amputation. Lifetime costs associated with lower limb amputation are over a half a million dollars including prosthetic costs (Dillingham et al., 2002; MacKenzie et al., 2007). In addition to these increased healthcare costs, these individuals must also learn to adapt to numerous hurdles associated with losing a limb.

The ability to walk unassisted is one of the defining cornerstones of mobility independence. Although not always apparent with the unaided eye, there are large asymmetries between the intact and amputated limb of transtibial amputees. The vertical GRFs in the intact limb are higher than those in the amputated limb which may be a compensatory mechanism to maintain forward momentum during walking to compensate for the lack of a powered push-off in the amputated limb. This increase in GRFs in the intact limb may contribute to the increased prevalence of OA in the intact limb.

Overall, persons with transtibial amputation are able to ambulate successfully and be active members of the community. They are able to competitively participate in sporting events at or near the same level as their non-amputee counterparts. Further research is required to understand how amputees interact with their environment when they are confronted with more challenging tasks beyond walking. These tasks may shed more light on the challenges faced by these individuals.

Two transtibial amputation techniques are commonly implemented: the traditional Burgess amputation and the transtibial osteomyoplastic amputation (Ertl) (Assal et al., 2005; Commuri et al., 2010; R. Dederich, 1983; Dionne et al., 2009; Ertl et al., 2013). The most commonly used transtibial amputation technique is the modified Burgess (Dionne et al., 2009). However, this amputation technique often leads to difficulties after amputation such as pain, swelling, instability, and significant residual limb atrophy, which results in reduced prosthesis use after rehabilitation. Although less common, the Ertl has been suggested to lead to improved functional outcomes following amputation. Using a “bone bridge”, the Ertl technique connects the tibia and fibula, seals the medullary canal, and sutures the anterior and posterior musculatures together. This technique commonly results in a healthier residual limb, reduced incidence of bone spurs, increased vascularity, and reduced incidence of skin ulcers (Rolf Dederich, 1963; Dudek et al., 2003; Dudek et al., 2005; Potter et al., 2007). It has also been suggested that the Ertl technique may enhance “end-bearing” capability of the residual limb compared to the Burgess (Mongon et al., 2013). This improved “end-bearing” may reduce the asymmetrical walking patterns compared to the Burgess, thus potentially reducing the risk of developing osteoarthritis, low back pain, or other comorbidities. Given the lack of data related to functional outcomes following Ertl amputations, determining whether the Ertl amputation technique has a functional advantage over the more common Burgess technique is needed.

Transtibial amputees undergo osseous and neurological changes following amputation. Bone often responds by increasing bone formation as bone spurs or bone overgrowth. This is more prevalent in the traditional technique rather than the Ertl. However, this may not fairly represent the outcomes of the traditional technique as it is more common than the Ertl. As a result more research is warranted to follow the progress of these individuals. However, the additional physiological changes of the improved vascularity sealing the medullary canal may suggest the Ertl results in a more physiological healthy limb.

Given the need for a more in depth understanding of these two amputation techniques, an analysis of functional tasks is warranted. Walking is very commonly investigated yet this task may not be sufficient to tease out differences in these two techniques. Therefore additional functional tasks should be investigated such as sit-to-stand and curb negotiation. Additionally, the inertial components of the amputated limb and prosthetic differ significantly from the intact limb. There are accepted methods to account and measure these differences. However, they are limited due to assumption and methodologies. The need for appropriate regression equations in this population are required and should be developed as part of this research.

Gait

Gait has been studied in an effort to compare differences in populations, patient groups, and disease states. Since it is the most widely used ambulation technique it also offers a “gold standard” for comparison of other movement patterns. These movement patterns can range from running, hopping, jumping, or uneven terrain. Importantly, it can also be used to evaluate abnormal walking patterns in patient populations such as below knee amputees. This is particularly important in this population for several reasons. The most common reason being unilateral amputees exhibit asymmetrical walking patterns. This has been attributed to a variety of factors ranging from prosthesis design, pain, muscle atrophy, and leg length discrepancies to name a few.

Persons with a lower extremity amputation are more likely to develop osteoarthritis (OA) in the contralateral limb (Burke et al., 1978). It is suggested that roughly 65% of unilateral amputees have some level of OA (Lemaire & Fisher, 1994; Melzer et al., 2001). It is important to note, however, unilateral, transtibial amputees are able to ambulate quite successfully without the use of assistive devices. Further, to the unaided eye, it is often difficult to identify a person with a unilateral, transtibial amputation. Thus, the underlying mechanisms which contribute to the prevalence of OA and asymmetrical walking patterns has been investigated through biomechanical techniques and are discussed below.

Spatiotemporal

Some of the most basic measures of inter-limb asymmetries are spatiotemporal measures. These measures include the time and distance measures of the limbs during walking, for example: stance time, swing time, double limb support time, and step length. Interestingly, there seem to be discrepancies within the literature regarding the trends seen in inter-limb asymmetries in these measures. Walking velocities of people with unilateral, transtibial amputation are significantly slower (1.0 to 1.3 $m \cdot s^{-1}$) than those for healthy individuals of similar age (1.3 to 1.5 $m \cdot s^{-1}$) (Boonstra et al., 1993; Isakov et al., 1997; Nolan et al., 2003; Powers et al., 1998). This one measure has been consistently reported through the literature.

When comparing unilateral, transtibial amputees walking at 0.94 $m \cdot s^{-1}$ and 1.38 $m \cdot s^{-1}$, Isakov, Burger, Krajnik, Gregoric, and Marincek (1996) found no significant differences between the intact and amputated limbs in stance time, swing time, double limb support time, step time, and step length. However, a year later, the same research group found the amputated limb spent more time in double limb support and took longer steps (Isakov et al., 1997). Other authors have supported this finding of a longer step length of the amputated limb (Mattes et al., 2000). However, Sadeghi, Allard, et al. (2001) found that step length of the intact limb was longer than the amputated limb. Further, Royer and Wasilewski (2006) found no significant differences in step lengths even though the walking speeds were similar. As is evident by these results, there appears to be some variability in the TTA literature.

Researchers have also found conflicting results in stance and swing time for each limb. Several researchers have found the intact limb spent an increased amount of time in stance and reduced time in swing (Nolan et al., 2003; Sadeghi, Allard, et al., 2001; Sanderson & Martin, 1997). However, Royer and Wasilewski (2006) and Fridman, Ona, and Isakov (2003) found no differences in swing or stance between the limbs. Interestingly, these inter-limb asymmetries (when found) have been noted to improve with walking velocity (Isakov et al., 1996; Nolan et al., 2003).

There may be multiple reasons for the inconsistent results between studies. One such explanation is the subject population in each study. Torburn et al. (1995) showed significant differences between traumatic and dysvascular amputees. Walking velocity and stride length were significantly lower in the dysvascular group than the traumatic group. The reason for these differences in the dysvascular and traumatic amputees is likely due to the activity levels of these groups. Generally, dysvascular groups are described as older, more sedentary populations whereas traumatic groups are generally younger and more active. Therefore, the decreases in performance may be attributable to differences in physical activity levels and aging processes.

Kinematics

Walking kinematics are less often reported in the literature than kinetic measures. However, as mentioned before, visually, amputees appear to ambulate similarly to healthy individuals. This is especially true at the knee and hip where peak angular differences between the intact and amputated limb are minimal, even when compared to healthy controls (Bateni & Olney, 2002; Sanderson & Martin, 1997). However, although there have been few reported differences in peak angles at the knee, knee range of motion is significantly smaller than the intact and control limbs (Ferris et al., 2012).

The largest notable difference in amputee gait kinematics is in the amputated limb at the ankle. Since the prosthetic ankle is unable to achieve plantar flexion like the intact limb, the plantar flexion angle is significantly smaller than the intact and healthy control at heel strike and push-off (Bateni & Olney, 2002; Isakov et al., 1996; Sadeghi, Allard, et al., 2001; Sanderson & Martin, 1997). Additionally, the ankle range of motion of the amputated leg ankle is smaller than the intact and control limbs (Ferris et al., 2012). As a result, the angular velocity at the ankle is significantly lower than the intact limb and healthy controls (Rao et al., 1998; Sanderson & Martin, 1997). Although the knee and hip joint changes exhibit minimal changes, they are generally more extended at heel contact with little to no change in angular velocity due to these position small changes (Donn, Porter, & Roberts, 1989; Hillery, Wallace, McIlhagger, & Watson, 1997; Isakov et al., 1996, 1997; Powers et al., 1998; Rao et al., 1998; Sanderson & Martin, 1997).

Kinetics

The GRFs of unilateral, transtibial amputees have been investigated. Most notably, the GRFs between the intact and amputated limbs differ significantly. Compared to controls, the intact limb produces larger vertical GRFs during walking at comparable speeds (Engsberg et al., 1993; Nolan et al., 2003). Compared to the amputated limb, the intact limb produces significantly higher vertical GRF magnitudes during walking (Engsberg et al., 1993; Isakov et al., 1992; Nolan et al., 2003; Sanderson & Martin, 1997). Some studies have suggested the first peak of the vertical GRF is similar between the intact and amputated limbs, but the second peak is significantly lower in the amputated limb (Nolan et al., 2003; Sanderson & Martin, 1997). This increased vertical GRF production of the intact limb has been suggested to be a contributing factor in development of OA in the intact limb (Lemaire & Fisher, 1994; Melzer et al., 2001). Although in terms of percentage, the difference in vertical GRF between legs is relatively small, the repetitive loading over many steps per day may cause micro trauma to the joints resulting in OA.

Anterior-posterior GRFs are also significantly altered between the limbs. The peak propulsive forces have been reported to be significantly smaller in the amputated limb compared to the intact limb (Silverman et al., 2008). Silverman et al. (2008) found that as walking speed increases, propulsive impulse of the amputated limb does not increase with all speed increases, and the propulsive impulse is significantly less than that produced by the intact limb or a limb of a healthy control. However, Silverman et al.

(2008) found no significant difference in the braking impulse as walking speed increased. This suggests the intact limb is mainly responsible for maintaining forward momentum of the body and maintaining walking velocity. Interestingly the amputated limb does not impede forward progress by increasing the braking impulse. An increase in braking impulse might have been expected due to the inability of the ankle to plantarflex during the first part of stance phase.

Transtibial amputees are at a functional disadvantage compared to healthy individuals. Due to the loss of the ankle/foot complex, they are unable to actively produce power at the ankle which is critical at push-off during walking. The body must respond by altering the neuromuscular control of other joints to compensate for this lack of active push-off created by the ankle plantarflexors (Winter & Sienko, 1988). These results are consistent in the literature where the participants are tested wearing elastic storage and release (ESR) prosthetic feet. Newer microprocessor controlled prosthetic feet aim to restore this power production at the ankle. Ferris et al. (2012) found these powered devices did increase power production at the ankle. However, the power produced was greater than the intact and control limbs and was temporally later in the stance phase (Ferris et al., 2012). Because research on these powered devices is limited, the emphasis will be on dynamic elastic type prosthetic feet.

This minimal power production at the ankle of the amputated limb results in kinetic differences up the kinetic chain when compared to the intact limb and controls. During early stance, the amputated limb ankle moment remains dosriflexor for longer (18% vs 6%, Winter, 1988) than controls (Bateni & Olney, 2002; Ferris et al., 2012; Winter & Sienko, 1988). Winter and Sienko (1988) contributed this increase in time of the dorsiflexor moment to the prosthetic foot design. Prosthetic feet are unable to plantarflex rapidly like an intact ankle and require the leg to rotate over the foot to create foot flat. Towards the middle of stance, the amputated ankle produces a significantly lower plantarflexor moment (60-70% of the intact ankle) compared to the intact limb and controls, which is again attributed to the inability of the ankle to actively plantarflex and (Ferris et al., 2012; Sanderson & Martin, 1997; Winter & Sienko, 1988).

At terminal stance and push-off, the energy absorbed and generated by the amputated ankle is significantly different compared to the intact and control limbs. For energy storing and releasing prosthetic feet, they are generally able to absorb as much energy as the intact limb prior to push-off (Ferris et al., 2012; Winter & Sienko, 1988). However, during push-off, the amputated limb produced significantly less power than the intact and control limbs due to the lack of powered push-off (Ferris et al., 2012; Silverman et al., 2008; Winter & Sienko, 1988). Throughout the entire stance phase, the amputated limb ankle produces less positive work than the intact and control ankles at 0.6 $m \cdot s^{-1}$, 0.9 $m \cdot s^{-1}$, 1.2 $m \cdot s^{-1}$, and 1.5 $m \cdot s^{-1}$. Further, the intact limb produced more negative work than the amputated limb at all four speeds; whereas the control only produces more negative work at 0.6 $m \cdot s^{-1}$and 0.9 $m \cdot s^{-1}$(Silverman et al., 2008)..

When compared to the intact limb and controls, the amputated knee joint moments are significantly smaller (Powers et al., 1998; Sanderson & Martin, 1997; Winter & Sienko, 1988). The intact and control limbs produce primarily extensor moments at 20% and 60% of stride. Although the amputated limb follows a similar pattern, the net moment is primarily flexor throughout the entirety of the stride. As a result, the net joint power for the amputated limb is smaller than the intact and control limbs (Ferris et al., 2012; Powers et al., 1998; Sadeghi, Allard, et al., 2001; Silverman et al., 2008; Winter & Sienko, 1988). Specifically, the net positive work performed by the amputated limb is significantly smaller than the intact and control limbs at 1.2 $m{\cdot}s^{-1}$and 1.5 $m{\cdot}s^{-1}$. Yet there is no significant difference in the net negative work performed between the limbs at any speed (Silverman et al., 2008). In addition to producing a greater amount of positive work than the amputated limb, the intact limb produced more positive work than the controls at 1.2 $m{\cdot}s^{-1}$ and 1.5 $m{\cdot}s^{-1}$ (Silverman et al., 2008).

In the frontal plane, the amputated limb produces a significantly smaller knee abduction moment through stance compared to the intact limb (Royer & Wasilewski, 2006). However, Sadeghi, Allard, et al. (2001) found only a small increase in frontal plane power generation in the amputated limb prior to push-off.

Hip moments have been reported to be highly variable between studies. Ferris et al. (2012) reported the hip moments at heel strike appear to be significantly smaller than the intact limb, but not significantly different from the control (Ferris et al., 2012). However, Sanderson and Martin (1997) reported the hip moments in early stance to be similar between the amputated and control limbs and did not differ through the entire stance phase. However, they reported the intact limb was smaller than the amputated and control limbs for the first half of stance.

Winter and Sienko (1988) reported that hip moments of the amputee are highly variable between individuals and prosthetic feet. As a result, the powers reported by these authors varied greatly. Ferris et al. (2012) reported no significant differences in hip power generation or absorption throughout the gait cycle. Winter and Sienko (1988) reported an increase in hip power generation at terminal stance. They suggested this increase in power generation was a compensatory mechanism at the hip for the lack of ankle plantarflexors. Sadeghi, Allard, et al. (2001) found significant increases in hip power in all three anatomical planes, again attributed to the lack of ankle power. Further, the authors reported two new bursts of power production and generation at heel strike not noted previously. On the intact side, they found an increase in hip extensor activity earlier in the stance phase possibly to help propel the body forward. Net positive work produced at the hip of the amputated and intact limbs was significantly higher than the control limb at 1.2 $m{\cdot}s^{-1}$ and 1.5 $m{\cdot}s^{-1}$ (Silverman et al., 2008).

In the frontal plane, the hip abduction moment of the intact limb was significantly larger than the amputated limb (Royer & Wasilewski, 2006). Sadeghi, Sadeghi, et al. (2001) reported significantly larger hip power absorption in the frontal plane during the first part of stance and reduced power absorption during the later portion of stance and swing compared to the intact limb.

Muscle Activity

Due to the loss of the limb, below-knee amputees have reduced muscle mass due to atrophy in the thigh and residual limb. The muscle excitation patterns of the thigh and hip have been studied in this group. However, muscle activation patterns of the residual limb (stump) have received considerably less attention due to the socket worn on the residual limb. During the gait cycle, the semimembranosis and vastus lateralis have been shown to increase the duration and magnitude of the muscle activity in the amputated limb compared to controls (Isakov, Burger, Krajnik, Gregoric, & Marincek, 2001; Isakov, Keren, & Benjuya, 2000; Powers et al., 1998; Schmalz, Blumentritt, & Reimers, 2001; Winter & Sienko, 1988).

Winter and Sienko (1988) found that the rectus femoris of the amputated limb showed increased activation throughout the entire gait cycle compared to controls. The biceps femoris long head demonstrates the largest increase in duration than the other hamstrings muscles, remaining activated for the majority of the gait cycle. The increase in activity of muscles acting at the knee during stance are suggested to act as knee stabilizers and likely result in underestimation of the knee joint moment or contributes to the reduced extensor function exhibited in the knee moment of the amputated side during stance (Powers et al., 1998; Sanderson & Martin, 1997; Winter & Sienko, 1988). At the hip, gluteus maximus shows an increase in activity over the entire stance phase compared to controls (Winter & Sienko, 1988). However, during swing, muscle activation between the intact and amputated limb are generally not different (Isakov et al., 2001; Isakov et al., 2000; Powers et al., 1998; Schmalz et al., 2001; Winter & Sienko, 1988).

Since there is a lack of information on the activation patterns of the muscles within the socket,Huang and Ferris (2012) attempted to use surface electrodes to measure the muscle activity of the residual musculature. Their main interest was to assess the feasibility of using surface electrode to control a myo-controlled prosthesis. They placed electrodes over medial and lateral heads of gastrocnemius and tibialis anterior within the socket. Amputees were asked to walk over a variety of walking speeds while wearing the electrodes. Their results indicated that they were able to collect repeatable electrical signal from each muscle at each speed.

Additionally, they were able to show an increase in muscle activity with increasing speed (Huang & Ferris, 2012). However, the activation patterns between each individual were highly variable and considerably different from controls. The authors suggested that this variability in individuals is due to neurological remodeling (plasticity) after amputation. They suggested the reorganization of muscle activity is primarily due to the individual use of each muscle. Some amputees may increase muscle activity within the socket to increase stability which is different from normal walking activation patterns. Although inconsistent, the authors believe these results suggest there is sufficient muscle activity to control a myo-electric prosthesis during walking. These devices would act similarly to upper extremity prostheses that use myographic inputs to control the hand mechanism (open, close, rotate). At the ankle, the activity patterns could be used to signal a powered prosthetic foot to produce powered plantarflexion at push-off.

Prosthetic Feet

Below-knee amputee gait is also influenced by the type of prosthetic foot/pylon/socket worn by the user. There are numerous types of prosthetic feet available to amputees on the market. These feet range from a Solid Ankle Cushioned Heel (SACH), Seattle Foot, many varieties of energy storing and releasing feet (ESR), and newer microprocessor controlled feet (Powerfoot and Proprio foot). The ESR feet are suggested to improve walking over SACH and Seattle feet due to their ability to store and release energy during the gait cycle. In order for the prosthetic limb to mimic the intact ankle, it must reproduce the action of the ankle during the gait cycle. The intact ankle moves through four distinct phases: controlled plantarflexion, controlled dorsiflexion, powered plantarflexion, and swing phase (Au, Berniker, & Herr, 2008).

During controlled plantarflexion, the dorsiflexors act to control the decent of the forefoot onto the ground during heel strike/early stance. Controlled dorsiflexion requires control of the leg as it rotates over the foot. It is during this portion of stance that energy is stored. In powered plantarflexion, at push-off, the energy stored during controlled dorsiflexion is released and coupled with concentric muscle action of the plantarflexors. Lastly, during swing phase, the dorsiflexors concentrically act to dorsiflex the foot to aid in toe clearance.

Gait asymmetries persist as does an increase in metabolic cost during walking regardless of foot type (ESR vs SACH). The type of prosthetic foot can increase or decrease the step length differences compared to the intact limb by up to 8cm (Barth, Schumacher, & Thomas, 1992). This change in step length varied from the SACH, Seattle Lightfoot, and several ESR type feet. Position of the prosthetic foot can also influence the inter-limb asymmetries. By increasing the external rotation of the foot from an optimal position, the symmetry between the limbs decreased measures of stance time, swing time, and step length (Fridman et al., 2003). It is important to note this is a large change in external rotation from the optimal change. Smaller changes (18°) did not result in changes in inter-limb symmetry indices.

The ESR feet have shown modest increases in the amount of energy returned during terminal stance, yet they only generate approximately 50% of the amount of power produced by an intact ankle at push-off (Au et al., 2008; Ferris et al., 2012; Herr & Grabowski, 2012; Hitt, Sugar, Holgate, & Bellman, 2010; Versluys et al., 2009). With the introduction of powered feet, specifically the Powerfoot, modest decreases in metabolic costs are seen (~8%) (Herr & Grabowski, 2012).

As mentioned previously, the Powerfoot has been shown to restore power production, yet it is temporally shifted to later in stance and is greater than the intact and control limbs (Ferris et al., 2012). However, the Powerfoot did not restore dorsiflexion during swing to aid in toe clearance. Similarly, the Proprio foot has been shown to decrease inter-limb asymmetries during slope walking and stair ambulation, but it still does not have active power production to compensate for the missing ankle (Alimusaj et al., 2009; Fradet, Alimusaj, Braatz, & Wolf, 2010).

Metabolic Cost

Walking energy consumption has been shown to increase roughly 20% in unilateral transtibial amputees when compared to controls walking at similar speeds (Gailey et al., 1994; Molen, 1973; Torburn et al., 1995). This trend towards an increase in metabolic energy expenditure has been noted regardless of the type of prosthetic foot when normalized to the distance traveled (ESR vs SACH) (Torburn et al., 1995). Additionally, when not normalized, the traumatic amputees appear to increase metabolic cost compared to dysvascular amputees and controls. However, when metabolic cost is normalized to walking distance, both the traumatic and dysvascular groups increase their metabolic cost compared to controls (Torburn, Schweiger, Perry, & Powers, 1994). Waters and Mulroy (1999) found that amputees increase metabolic cost when increasing walking velocity in a similar fashion as controls. Regardless of speed however, the amputee's metabolic cost was consistently higher than the controls (Torburn et al., 1994; Waters & Mulroy, 1999).

As mentioned previously, transtibial amputees selectively walk at velocities slower than control subjects. However, like control subjects, they are most metabolically efficient at speeds around 1.3 $m \cdot s^{-1}$ which is often faster than their self-selected walking speed (Molen, 1973). Although these trends towards increased metabolic cost in transtibial amputees are consistent throughout the literature, the causality behind the increase is unknown.

Given the above review of the literature of TTA gait and the factors associated with this task, we have a sufficient basis to evaluate tasks beyond gait. For the purposes of this literature review, the other tasks will focus on are curb negotiation and sit-to-stand. Additionally, gait will be used to evaluate the effectiveness of the body segment parameter regression equations developed in this dissertation.

Curb Negotiation

Slips, trips, and falls (STF) in a community-based environment pose a public safety concern. Falls on level or uneven surfaces is the 7^{th} leading cause of death in the US according to the National Safety Council (2011). Injuries from falls have been reported to include: sprains, fractures, cuts, and bruises. Injuries sustained from a STF may result in financial repercussions for the public or private entity where the injury occurs. For example, the average cost of a STF accident caused by an uneven surface or curb at shopping malls across the US is over $10,000 per case (Hantula, Bragger, & Rajala, 2001). These cases account for roughly 17% of all accidents in malls. This is on par with accidents from water and ice accidents in malls.

To avoid SLFs, the body must accommodate varying surface conditions during normal gait. As such, investigation into obstacle crossing has received much attention. The body must make several accommodations to successfully negotiate an obstacle. The supporting limb must support the body while the lead limb crosses the obstacle. While advancing, the lead limb must have sufficient toe clearance as to not catch the obstacle resulting in a trip or fall. After toe clearance, the lead limb must also clear the obstacle with enough heel clearance to prevent the posterior of the foot from stepping on or tripping over the obstacle. Once placed on the ground, the lead limb then must support the body as the trail limb comes forward. It too must cross the obstacle with enough toe and heel clearance to prevent a possible fall or perturbation or walking.

Although obstacle crossing has received much attention, curb negotiation has received less attention. Curb negotiation is similar to obstacle accommodation, however it differs in that the body must step or step down while maintaining forward motion and avoiding a trip or fall. Although similar, curb negotiation is distinctly different from obstacle negotiation or stair climbing. Despite these differences, research is lacking in this area.

89.8% of all trips and falls have been suggested to be related to two gait characteristics: step length and step time (Pavol, Owings, Foley, & Grabiner, 1999). The hip, knee, and ankle must flex during swing sufficiently to enable the toe to clear the ground. In healthy adults, during normal walking, toe clearance averages 1.3 -1.9 cm (Moosabhoy & Gard, 2006; Murray & Clarkson, 1966; Winter, 1992). This occurs midway (~50%) through the swing phase (Loverro et al., 2013; Moosabhoy & Gard, 2006; Winter, 1992). At the end of swing phase, the orientation of the foot changes and the heel comes closer to the ground just prior to heel strike. According to Winter (1992), toe clearance was most sensitive to the stance limb ab/adduction and swing leg knee angle followed by ankle and hip angle. Moosabhoy and Gard (2006) developed an analytical technique to identify changes in leg length and joint angles which would affect toe clearance. Their results differed from those found by Winter (1992). Their results suggest that while the hip and knee are important for toe clearance during mid swing, the ankle has the largest potential to increase or decrease toe clearance. The knee and hip were most likely to aid in toe clearance in early swing.

This suggests that these two measures should be included in evaluations of the potential risks for trips and falls. In a healthy population, it was found that when negotiating a curb (both descent and ascent) they were able to make adjustments in both step length and time in order to avoid placing their foot in a potentially hazardous position near the curb (Crosbie & Ko, 2000). However, elderly populations tend to place their foot closer to the step when negotiating a curb during both ascending and descending (Lythgo, Begg, & Best, 2007). In both of the previous studies, the participants knew of the curb and were able to make adjustments as needed to prevent a trip or fall.

van Dieen, Spanjaard, Konemann, Bron, and Pijnappels (2007) found that when a healthy individual expectedly descended from a curb, they would increase their step length. However, when they unexpectedly stepped down from a curb, their step length did not increase, and a fall occurred due to an inability to generate a rapid forward step (van Dieen et al., 2007).

A number of studies used center of pressure (COP) and center of mass (COM) values when describing negotiation of a curb or obstacle. During obstacle crossing at 15% of a person’s body height compared to level walking, no changes were seen in the loading response phase. During mid-stance, the COP velocity decreased when crossing the obstacle. During pre-swing, the COP velocity of the lead foot increased, while the trailing foot decreased in COP velocity (Y. Wang & Watanabe, 2008). Age differences (young vs. old) in crossing an obstacle and stepping onto a curb were shown by looking at anterior-posterior displacement (A-P), medial-lateral (M-L) displacement, COP average velocity, and stance time (Kim, 2009). This study showed that young adults' COP travels further in A-P and M-L directions during both tasks, average COP velocity is greater during both conditions, and stance time is less for young adults during both tasks.

The orientation of the COM relative to the COP in both the sagittal and frontal planes have been expressed as the inclination angle which is "the angle formed by the intersections of the line connecting the COP and COM with a vertical line through the COP" (p. 570) (Lee & Chou, 2006). Comparing the inclination angle between elderly adults with a history of falling to age matched controls; the fallers displayed a greater medial lateral (M-L) inclination angle than the stable adults, and a smaller anterior-posterior (A-P) inclination angle than stable adults during both obstructed and unobstructed gait (Lee & Chou, 2006). In healthy young adultsm a comparison of inclination angle between level ground walking, curb ascent and curb descent was made (Norrish, O'Reilly, Whitney, Campbell, & MacDonell, 2009). The subject's peak M-L inclination angles were larger in both curb ascent and curb descent than during level ground walking. The subject's average ML inclination angles were larger in the lead limb during curb ascent and in the trail limb during curb descent.

During negotiation of obstacles at varied heights, analysis of GRFs showed that vertical impulse increased as a function of obstacle height (Patla & Rietdyk, 1993). Patla and Rietdyk (1993) suggested that the anterior-posterior impulse during the braking phase and push-off phase, as well as the double-support phase increased in magnitude as obstacle height increased.

When describing differences during stepping down from an expected level change versus an unexpected level change, there was no double support phase observed when encountering an unexpected level change (van Dieen et al., 2007). This resulted in an increased vertical impulse, as well as an increased peak force of the lead limb when striking the landing, meaning that the subjects would have to increase their muscle activity in order to maintain postural stability (van Dieen et al., 2007).

Patla and Shumway-Cook (1999) suggested variables which must be manipulated to prevent falls while negotiating an obstacle: toe clearance, hip and toe horizontal velocity, and COM position. When walking over obstacles of various heights, toe clearance increased for all heights (~10 cm) compared to level walking (3.5 cm). As height increased, toe velocity and hip velocity decreased (Patla & Rietdyk, 1993). This may help reduce fall risk by decreasing the possible impact velocity with the obstacle which would result in a larger perturbation to forward motion.

Loverro et al. (2013) evaluated the minimum foot clearance on several obstacles and the location of the lowest clearance point on the foot while negotiating a curb (17 cm). During curb ascent, they found the location of the minimum foot clearance happens at the edge of the curb for the lead limb (6.2 cm), and equal occurrences at the edge and surface of the curb for the trial limb (4.4 cm) (Loverro et al., 2013). On the descent, the minimum foot clearance occurred at the edge of the curb for the lead limb (2.1 cm). No minimum foot clearance was found for the trail limb due to the trail limb foot placement relative to the edge of the curb. Interestingly, the foot clearance heights were generally higher during curb negotiation than for stair ascent (3.9 cm) and descent (3.0 cm) of the lead limb. In general, minimum foot clearance was found toward the toe/forefoot region of the lead and trail limb, although this was highly variable (Loverro et al., 2013). During curb ascent, the lead limb has a longer step length (69.5 cm) than during descent (55.8 cm). These step lengths were longer than the step lengths for stair ascent (54.7 cm) and descent (48.8 cm). Schulz (2011) found toe clearance increased while negotiating obstacles by altering joint kinematics during swing. As a result, the author suggested fall risk decreased.

Relative to age-matched, able-bodied individuals, transtibial amputees have an increased risk of falling and fear of falling (W. Miller et al., 2001; Vanicek et al., 2009). As a result, 60% of these individuals report falls affect their daily activities, work, and confidence (Kulkarni et al., 1996).

McFadyen and Prince (2002) investigated stepping up onto a step in older and young adults. They found decreased toe clearance distances in the older adults which they contributed to limited frontal plane hip motion. Although Patla and Shumway-Cook (1999) suggested toe clearance was most important during obstacle crossing, Begg and Sparrow (2000) suggested heel clearance is most important during stepping up or down onto a step or curb. They found when compared to young adults, older adults had lower heel clearance when stepping onto a curb; conversely older adults had more vertical foot clearance than the young adults while stepping off a curb (Begg & Sparrow, 2000). Further, when stepping down, the older adults tended to place their trail foot further from the step edge than the young adults. Begg and Sparrow (2000) suggested this may negatively affect their safety.

In addition to evaluation of level walking, several physical environmental factors are encountered on a daily basis including stairs and curbs. Interaction with these environmental factors can increase the risk of falling in any population; however, the risk may increase in amputees. Interestingly, persons with transtibial amputation report curb negotiation is more demanding than negotiating stairs even though stairs and curbs are encountered with the same frequency (Larsson et al., 2009; Shumway-Cook et al., 2002). Surprisingly, the underlying biomechanical mechanisms making this task more challenging for transtibial amputees are unknown. By compiling a body of literature of tasks similar to curb negotiation, we can develop testable hypotheses.

Sit to Stand

Sit to Stand in Non-amputees

Many variables have been investigated in non-amputees during the sit-to-stand movement. Manipulations of foot placement, seat height, and trunk position have been evaluated using biomechanical analysis. Hughes, Weiner, Schenkman, Long, and Studenski (1994) identified three strategies for rising from a chair: momentum, stabilization, and combined. The momentum strategy relies on increased trunk flexion and arm swing to increase horizontal momentum assist with standing. Stabilization relies on the position of the base of support and center of mass, where generally there is little change in momentum with this strategy. The combined strategy relies on the momentum of the trunk and position of the base of support. In older adults, the momentum strategy resulted in a decreased time to stand and a higher success rate to stand (Hughes et al., 1994). As seat height decreases, the vertical GRF increases when the feet are placed such that the angle between the shank and foot was 90° or more posteriorly (increased knee flexion and dosiflexion) (Kawagoe, Tajima, & Chosa, 2000).

Foot Placement

Placement of the feet relative to the chair directly influences knee joint angle. When placed more posteriorly (closer to the chair), knee angle increases (more flexion) than when placed anteriorly (further forward), knee angle decreases (extension). Khemlani et al. (1999) found task time to rise was longer when the feet were placed more anteriorly than posteriorly. When the feet are placed in a more knee extended position, the individual swings their arms and leans their trunk forward to assist with rising from the chair (Fleckenstein, Kirby, & MacLeod, 1988). As a result, hip flexion increases along with the hip extension moment when the feet are placed further from the chair (Fleckenstein et al., 1988; Khemlani et al., 1999; Shepherd & Koh, 1996).

Placement of the feet also alters the vertical GRFs. When the feet are placed more anteriorly (less than 90° knee flexion), the vertical GRF is greater than when the knees were placed at 90° or greater (increased flexion) (Kawagoe et al., 2000; Stevens, Bojsen-Möller, & Soames, 1989). Additionally, the anterior component of the GRF was consistently lower in the anterior direction than the posterior foot placement regardless of chair height. In contrast, the anterior foot placement resulted in an increased anterior force. With an anterior foot placement, muscle activity duration increases to compensate for the increased distance covered by the center of mass (Kawagoe et al., 2000; Khemlani et al., 1999).

Trunk Placement/Movement

When subjects were asked to exaggerate the forward trunk lean from the normal trunk motion, Doorenbosch, Harlaar, Roebroeck, and Lankhorst (1994) found the hip joint angle increased. Additionally, they found the hip and ankle moment increased significantly whereas the knee moment decreased. This suggested a transition from a knee strategy to a hip and ankle strategy as the trunk flexion increased. Muscle activity of the hamstrings and gastrocnemius, soleus, gluteus medius, muscles increased with increased trunk flexion (Doorenbosch et al., 1994; Roebroeck et al., 1994). Further, older adults and obese patients increase trunk flexion compared to younger and healthy adults when rising from a chair (Gross et al., 1998; Kerr, White, Barr, & Mollan, 1997; Papa & Cappozzo, 2000; Sibella, Galli, Romei, Montesano, & Crivellini, 2003). They were found to increase their horizontal momentum and lower the center of mass with the increased trunk lean prior to standing from the chair which was suggested to aid in standing (Gross et al., 1998; Papa & Cappozzo, 2000; Shepherd & Koh, 1996).

Sitting

Little information is available on the sitting phase. However, Kerr et al. (1997) found that during stand to sit movements, trunk lean was significantly less than that during standing regardless of age.

Sit to Stand in Amputees

It has been estimated that people with TTA sit-to-stand (STAND) roughly 50 times per day (Bussmann et al., 2004; Bussmann et al., 2008) which is similar to non-amputees(Dall & Kerr, 2010). Therefore, these tasks are important activities of daily living. Movement strategies of persons with transtibial amputation have been studied less by researchers than other populations. However, researchers have found during sit to stand, patients transfer weight towards the unaffected leg (Agrawal et al., 2011; Ozyurek et al., 2013). Agrawal et al. (2009) found patients with TTA produced 27% more vertical GRF with the intact limb during a sit to stand movement compared with the prosthetic side. Non-amputee controls, however, exhibited less than 10% asymmetry in vertical GRF during the same movement.

Patients who have undergone total knee replacement, hip replacement, or have hemiparesis, preferentially choose to place their feet more posteriorly, which reduces asymmetries in GRFs between limbs (Barclay-Goddard, Stevenson, Poluha, Moffatt, & Taback, 2004; Farquhar et al., 2009; Galli, Cimolin, Crivellini, & Campanini, 2008; Roy et al., 2006; Talis et al., 2008). This position results in the adoption of an ankle/hip strategy compared to a knee strategy. However, regardless of foot placement, patients selectively shift their weight to the unaffected limb, increasing the GRF of the uninvolved limb (Farquhar et al., 2009; Mizner & Snyder-Mackler, 2005; Roy et al., 2006). This movement strategy is similar to that adopted by transtibial amputees.

As common as the sit-to-stand task is in daily life, it is surprising that few studies have adequately investigated the task. The needs of the amputee population could benefit from this analysis. In addition to adding to the current body of literature, this study will also help guide surgical interventions for future amputees.

Lower Extremity Inertial Measurements

Estimating Body Segment Parameters

Inverse dynamics analysis requires knowledge of body segment parameters, motion capture, and GRFs. Today, we are able to reliably capture motion and GRFs in three planes through opto-electric infrared cameras and force plates (piezoelectric or strain gauge). However, inertial properties of the body are variable depending on the methodology used to obtain them. Inertial properties of the body include mass, center of mass location, and moment of inertia of each segment within the body. In conjunction with motion capture and GRFs, inertial properties make the calculation of joint moments and powers possible.

Researchers have used a variety of methods for estimating body segment parameters. The earliest investigators relied heavily on cadaveric data to estimate these body segment measures (Chandler, 1975; Clauser et al., 1969; Dempster, 1955; Hinrichs, 1985, 1990). These researchers developed regression equations to calculate the percent mass of each body segment, location of the center of mass as a percentage of segment length, and the location of the moment of inertia relative to the axis of rotation. Briefly, the body was dissected into segments. Each segment was weighed and expressed as a percentage of the overall body mass. The center of mass location of the segment was measured via reaction board testing. The segment is placed on a board which rests on two "knife edges" and one end rests on a scale. The center of mass location is calculated:

$$x = \left(\frac{R_2 - R_1}{W}\right) * d$$

Where x is the center of mass location, d is the distance between the two knife edges, R_1 is the reaction of the board without the segment, R_2 is the reaction of the board with the segment, W is the weight of the segment. The center of mass location is expressed as a percentage of the segment length. Moment of inertia was measured using an oscillation technique (Dempster, 1955; Mattes et al., 2000). The segment is suspended as a pendulum where the arc of the pendulum is known and the oscillation period (τ) is measured. Given τ, we can solve for the moment of inertia (I_{axis}):

$$\tau = \sqrt{\left(\frac{I_{axis}}{mgd}\right)}$$

Where m is the mass of the segment, g is the constant acceleration due to gravity (-9.8 $m \cdot s^{-2}$), and d is the distance from the axis of rotation to the center of mass location. However, because of the nature of measurements of cadaveric specimens, there are several limitations. These include an older population, pooling of body fluids, and loss of tissue, segmentation error, and measurement error.

More recent researchers have used dual x-ray absorptiometry (DEXA) scans (Durkin & Dowling, 2003; Durkin et al., 2002), gamma radiation scanning (de Leva, 1996; Zatsiorsky, 1983), magnetic resonance imaging (MRI) (Cheng et al., 2000; Martin et al., 1989; Mungiole & Martin, 1990; Pearsall et al., 1994), kinematic models (Drillis et al., 1964; Herbert Hatze, 1975), and geometric models based on geometric principles (Hanavan, 1964; H. Hatze, 1980; Jensen, 1978). Because many of these techniques have also produced regression equations that can be applied to similar populations, the popularity of these methods has been increasing.

With the estimation and definition of these inertial properties we assume all segments are treated as rigid bodies. This may not necessarily be true; there may be some tissue deformation during movement or loading. Further, identification of the appropriate landmarks which delineate each segment may be difficult depending on the over lying tissue (ex: obese patients). However, given these various techniques, there is a high correlation between the precision of the measure of the inertial properties (Challis, 1999).

The variability in the inertial properties has generally been accepted to have little influence on the joint moments during the stance phase of walking since due to the large GRF, center of pressure location, segment position, and moment arm lengths during stance are generally much larger than the inertial properties (Challis, 1996; Challis &

Kerwin, 1996). Therefore, the differences in inertial property selection are masked by larger outcome measures reported in the literature. As a result, it is important to investigate these differences during swing where they are more likely to cause differences in gait.

Estimating Inertial Properties of the Prosthesis

For amputee data, unfortunately, there are no accepted methods of measurement of these data for the amputated limb. Researchers often use estimates of the intact limb to model the inertial properties of the prosthetic limb even though the inertial properties of the prosthetic side are far less than those of the intact side.(Czerniecki et al., 1991; D. I. Miller, 1987). Smith, Ferris, Heise, Martin (2014) found that the mass is significantly lower in the amputated limb and mechanically, the amputated limb is significantly different from the intact limb during swing phase. Smith et al. (2014) developed an oscillation rack to measure directly the inertial properties of the prosthesis and combine these properties with estimates of the intact residual anatomy to model the amputated limb. These values are then used to calculate the appropriate joint moments and powers. Although this technique has limitations: it requires labs to build their own oscillation rack and perform these measurements per subject. To date, no regression equations have been developed to assist with these calculations.

After the loss of a lower limb, a prosthetic is fabricated using lightweight materials such as carbon fiber. As a result, the resulting prosthetic is lighter than the limb that it is replacing. This difference in mass results in asymmetrical inertial properties between the intact and amputated limb. The resulting prosthesis and residual limb roughly 35% less mass and the center of mass is located 35% closer to the knee joint than the intact shank and foot (Lin-Chan, Nielsen, Yack, Hsu, & Shurr, 2003a; Mattes et al., 2000). Further, as a result of the difference in the distribution and lower mass, the moment of inertia of the amputated limb is also significantly lower (~60%) compared to the intact limb. Some researchers have suggested these differences in inertial properties may address some of the asymmetries noted in gait of amputees (Mena, Mansour, & Simon, 1981; Tsai & Mansour, 1986).

For below knee amputees, using intact body segment parameters to estimate inertial properties of the prosthetic side may not be the most appropriate method. J. D. Smith et al. (2014) developed an oscillation rack to approximate the inertial properties of the amputated limb and found that the mass is significantly lower in the amputated limb. These values are then used to calculate the appropriate joint moments and powers. They found that these differences in inertia do not result in significant differences in kinetics during the stance phase of walking (J. D. Smith et al., 2014). This is likely due to the significantly larger GRFs overriding the differences in inertial properties during stance. However, during swing, significant differences were found at the hip and knee. Non-significant differences were noted in the ankle. However, these small differences were propagated up the kinematic chain and became significant at the knee and hip. As a result, it is important to investigate these differences during swing where they are more likely to cause differences in gait. To date, no regression equations have been developed to assist with these calculations.

Amputation Techniques

Transtibial amputation (TTA) is one of the most common amputations in the United States (D. G. Smith & Fergason, 1999). These amputations arise from a variety of needs ranging from dysvascularity to traumatic injury. The ultimate goal of the procedure is to maintain a healthy residual limb to enable walking independence. Due to the prevalence of transtibial amputations, multiple surgical techniques have been developed. The two more established techniques rely on a posterior flap and are commonly referred to as the Burgess and Ertl techniques. Both procedures have some benefit over the other. For example, a skilled surgeon is required for the Ertl procedure, where as a general orthopedist can effectively perform a Burgess procedure. Reasons for these differences are described in the following paragraphs describing each procedure.

Surgical Techniques

The traditional Burgess technique involves transecting the tibia and fibula at the same level. The anterior tibia is beveled to prevent sharp edges. The posterior musculature is salvaged and sutured over the ends of the tibia and fibula and sutured to the anterior tibia (Assal et al., 2005; D. G. Smith & Fergason, 1999). The process of attaching the musculature to the tibia is referred to as a myodesis. No attempt is made to seal the medullary canal.

The osteomyoplastic procedure (Ertl) is a longer, more involved, surgery This procedure has been modified over the years. The unmodified procedure involves transection of the tibia and fibula at the same level in the same fashion as the traditional technique (Ertl et al., 2013; Loon, 1962). However, a strip of periosteum seeded with bone chips is then folded over and sutured to the fibula. This procedure will seal the medullary canal. Additionally, over time the bone chips will grow and ossify within the periosteum and form a bone bridge between the tibia and fibula (Ertl et al., 2013; Loon, 1962).

The Ertl procedure has been modified using a portion of the removed fibula to create a bone bridge. The bone is fitted into a slot on the tibia and fibula to keep it in place. At the surgeon's discretion, sutures and screws are used to secure the bridge to the tibia and fibula.(Commuri et al., 2010; DeCoster & Homedan, 2006; Hussainy, Goesling, Datta, & Saleh, 2004)

In the traditional Burgess procedure, muscles of the calf are severed at their bellies and their normal length tension relationship is lost. However, during the Ertl procedure, the anterior and posterior muscles are sutured to each other over the osteoperiosteal bridge and to the tibia. The process of suturing these muscles together is referred to as a myoplasty. Theoretically this helps maintain the length tension relationship, aid in venous return, and minimize venous stasis. During any procedure, handling of the musculature is important (Rolf Dederich, 1963). The tension between opposing muscle groups must be taken into account to prevent the residual limb from ab/aducting or flex/extending. Rolf Dederich (1963) reported success (less pain, healthy residual limbs) when performing amputations in the manner described above. The Ertl procedure is completed by suturing the anterior and posterior musculature to each other and the skin is then sutured closed (Rolf Dederich, 1963).

The Ertl technique is suggested to be superior to the Burgess technique for several reasons. First, the fibula is more stable in the residual limb which may help increase lateral stability and reduce pain. Additionally, the connection of the tibia and fibula is suggested to improve distal limb loading (Ertl et al., 2013).

Although a suggested benefit of an Ertl procedure is end loading of the limb, many prosthetists still create a socket using circumferential loading, patella tendon bearing sockets, or total surface bearing which are appropriate for traditional amputations. Unfortunately, little research is available to suggest there is an increase in end bearing ability, thus spurring a change in socket design. To date, only one study has investigated the pressures within the socket of an Ertl amputee wearing a total surface bearing prosthesis (Commuri et al., 2010). This case study evaluated many walking tasks: normal walking, brisk pace, backward walking, sideways gait, and stair climbing. Pressures within the socket were evaluated for each task. The investigators reported forces in proximal and distal sensors. The distal sensors indicated weight was borne on the distal end of the stump during all activities. This important finding demonstrates the ability of the amputee to bear weight painlessly. However, it is important to note these results were not compared to a traditional amputation and thus conclusions are hard to draw. Further, the socket was not altered in any way to fully evaluate distal loading of the limb. However, if an Ertl limb is able to tolerate increased loads, this may help to contribute to a decrease in inter-limb asymmetries.

Despite the fact that there is an abundance of speculations as to why the Ertl may be a superior surgery functionally, little scientific research is available to support these speculations. Further, if there are no functional benefits of this complicated surgery, are there physiological changes which occur which may be beneficial? If not, the added difficulty and time may not outweigh the benefits.

Following amputation, several complications and/or adaptations due to the amputation may arise. One such complication is the continued development of bone at the distal ends of the severed bones. Small growths are referred to as bone spurs; a large growth of bone into areas which normally do no ossify is termed bone overgrowth. Both processes may result in significant amounts of pain and possibly revision surgeries. In addition to bony changes, the neurological system must adapt to the removal of a limb. This is commonly manifested as phantom pain. The individual feels excruciating pain in the limb that has been removed.

Bone

To understand how bone responds to the trauma of amputation, a review of the physiology of bone is required. Although not directly related to the study, it is important to understand the physiological changes that occur following amputation. Bone marrow is located in the medullary cavity of long bones (Gurkan & Akkus, 2008). Contained within this space are where red blood cells are formed, fat is stored, and the central artery and vein pass. Blood flow within the medullary canal flows out to the bone from the central artery and then diffuses back into the central vein. Housed within the marrow are the mesenchymal cells and hematopoietic stem cells which give rise to osteoblasts and osteoclasts responsible for bone generation and resorption. It is suggested that these cells maintain bone homeostasis with mechanical loading of the bone.

Adult bone marrow contains mesenchymal cells which are able to differentiate into different types of cells within the body (muscle, bone, tendon, etc.) for regenerative purposes (Gurkan & Akkus, 2008). These cells are responsive to mechanical signals such as hydrostatic pressure, fluid flow shear stress, and rheological properties of their environment. The marrow is susceptible to intermedullary pressure changes due to mechanical loading of the bone due to activity or inactivity (Gurkan & Akkus, 2008).

In the intact limb, the intramedullary pressure is about one fourth of the rest of the system, roughly 30mm Hg (Gurkan & Akkus, 2008). Intramedullary blood flow is directly related to the intramedullary pressure and may be the result of total blood flow into and out of the bone. Although mean arterial pressure may change, the pressure within the medullary canal will remain constant. This suggests a regulatory mechanism within the system. However, if venous congestion or the arterial blood supply increases, the intramedullary pressure will also increase. Additionally, with an increase in venous congestion, the blood supply to the bone decreases. This has been suggested to lead to necrosis (Gurkan & Akkus, 2008). Often following amputation, there is an increase in venous congestion at the distal end of the residual limb.

To further investigate the influence of venous occlusion on bone growth, Welch, Johnston, Waldron, and Poteet (1993) studied venous occlusion in goats over a 30 day period. The femoral vein was occluded in one group of goats whereas the control group underwent a sham surgery. They found compared to a control group, cortical and cancellous bone growth increased significantly in the venous occlusion group. The increase in bone growth (89%) was accompanied by an increase in osteoclast proliferation on the surface of the bone. The authors suggested the increase in periosteal pressure stimulated osteoclasts (Welch et al., 1993). This is important for amputees since this increase in bone growth could result in the development of bone spurs on the distal end of the residual limb.

The viscous nature of marrow has been described as Newtonian in nature (constant viscosity which is independent of shear rate) (Gurkan & Akkus, 2008). It has been suggested, though animal models and bed rest model, that a decrease in pressure results in a loss of bone whereas an increase in pressure resulting in increased bone growth. Additionally, intermedullary hypertension due to venous occlusion increases the pressure and has shown to have a bone forming effect.

Shear stress through fluid flow is also suggested to influence the activity of osteoclasts and osteoblasts. Most often bone generation is preceded by done degradation. The stimulus for bone generation comes from the osteoactive agents (such as prostaglandin E2 and I2). However, the flow within the bone is from the center of the bone out. Therefore the signals to produce bone are able to travel from the inside out to signal bone growth.

There are two competing theories on the mechanisms contributing to bone formation through intermediary pressure mechanisms. First, the pressure may stimulate osteoblast formation thereby increasing bone. The second is that there is an increase in nitric oxide which catalyzes a stimulus to osteoprogenator cells within the marrow which produce osteoblasts (Gurkan & Akkus, 2008).

Turner and Robling (2004) further investigated two pathways of intramedullary stimulation for the stimulation of bone growth. As fluid passes through the canaliculi, a mechanoreceptor or a Ca^{2+} modulated release may stimulate the release of ATP which then triggers the release of PGE2 to stimulate the proliferation of osteoclasts. However, it is unknown if the fluid force alone is sufficient to stimulate these changes. (Turner & Robling, 2004) Additionally, the release of prostaglandins and nitric oxide after mechanical loading leads to the stimulation of bone growth. Nitric oxide specifically inhibits osteoblasts, preventing bone breakdown. Therefore, venous stasis may trigger these pathways thus increasing bone growth/density.

Rolf Dederich (1963) referred to an 1899 study where the marrow of the bone and periosteum was removed from the end of the bone following an amputation surgery which resulted in necrosis. Because of this anecdotal evidence, he supported the myoplasty procedure where the periosteum is used to seal the medullary canal. This procedure is suggested to restore a positive pressure within the bone and aid in venous return.

Venous stasis has been shown to stimulate periosteal bone growth in dogs, goats, and rats; however, the mechanisms are unclear. L. Wang, Fritton, Weinbaum, and Cowin (2003) examined the ability of fluid shear stress within the bone structure (cannaliculi) to trigger bone growth. They developed an osteon model to quantify the fluid shear stress within the lacunar-canalicular pores. They described two forces which drive the fluid: cyclic mechanical loading and pulsatile extravascular pressure in the osteonal canal. This model would act similarly to the distal end of a long bone (ex: tibia) where the medullary canal was not sealed and left open.

In their model, fluid pore pressure and shear stresses were derived using a poroelasticity theory. (L. Wang et al., 2003) Their results suggest mechanical loading is the largest contributor to shear stress that acts on the bone cells. The pulsatile force of blood circulation induces a significantly smaller force. They concluded this small force from blood stasis is unlikely to contribute to bone growth. However, the increase in pressure may ultimately increase pressure under the periosteum thereby stimulating bone growth. The authors also suggest that rather than a mechanical pathway to stimulate bone growth due to venous stasis, there is a lack of clearance of waste materials (carbon dioxide tension, pH, and oxygen tension) which may contribute to the periosteal bone formation. Wang et al. (2003) suggested over time the buildup of these nutrients may contribute to bone necrosis. Necrosing bone and tissue would result in revision surgeries and decreased mobility in persons with an amputation.

Regardless of the cause, it common to see various bony changes in individuals with amputation ranging from bone spurs to bone overgrowth. Bone overgrowth is commonly observed in pediatric amputees (Dudek et al., 2003). Bone overgrowth is characterized by a proliferation of bone in the tissues adjacent to the distal portion of the amputated bone. Dudek et al. (2003) differentiated between bone over growth and heterotopic ossification (HO). HO is a bone outgrowth of trabecular bone from the bone itself. However, it can also refer to bone growth in tissues that are not normally ossified. It is most commonly seen in the spine or following a trauma. In pediatric amputee literature, this process is referred to as bony over growth or osseous overgrowth. The authors chose to use the term bone overgrowth to describe the condition as it encompasses over growth of bone contiguous with the bone and ossification of tissues that do not normally ossify. Often a bone spur is the result of the periosteum being stripped; however, the ideology of bone overgrowth is unknown. Yet, bone overgrowth is more frequently observed in traumatic amputations (43%) and pediatric populations. (Firth, Masquijo, & Kontio, 2011)

Bone overgrowth commonly results in pain, poor socket fit, skin perforation, and ulcers. This process is different from the development of a bone spur. Dudek et al. (2003) presented two adult cases reports. These were the only adult subjects the author was familiar who had demonstrated the development of bone overgrowth. One subject had a large spike at the end of their residual bone which caused significant pain. This was revised surgically and resulted in significantly less pain. The bone overgrowth did not return. The second patient also had significant bone overgrowth but the pain was managed through socket fit. Both patients had undergone a traditional amputation.

As Dudek et al. (2003) suggested, bone overgrowth is not prevalent in adults, and is likely associated with trauma. However, after that article was written, the incursions in Iraq began. As a result, an increase in the prevalence of bone overgrowth in military personnel during operations Iraqi Freedom (OIF) and Enduring Freedom (OEF) was evaluated (Potter et al., 2007).

Potter et al. (2007) evaluated the military patient records for bone overgrowth following an amputation, prevalence of bone overgrowth (mild, moderate, severe), location of injury (all were traumatic), and treatment. Of the 213 residual limb radiographs, 134 (63%) showed evidence of bone overgrowth. Seventy-two of these limbs were moderate to severe. Twenty-five limbs with bone overgrowth required excisional surgery. Of these patients, none suffered a spinal cord injury and only two received a traumatic brain injury. Six of the patient’s intraoperative cultures returned positive for continuing infection (previously treated before surgery). These were treated with specific antibiotics following surgery.

Potter et al. (2007) go on to describe many attributes to wound care and follow-up; however, the main interest of this entry is to focus on the bone overgrowth, only the sections referring to this are entered. Following surgery, it is common to place negative pressure dressings or compression bandages on the residual limb. However, the authors suggest the shear stress from the negative pressure dressing at the cellular level may stimulate endothelial cell growth pathways. This may also be a contributing factor to increased bone growth (Potter et al., 2007). Due to the rapid rise in the number of adults presenting with bone overgrowth (specifically those returning with blast injuries), the authors also support the hypothesis that traumatic events may stimulate osteoblasts.

This newer article has expanded on the article of Dudek et al. (2003). The five-year difference in the two articles (few cases vs. over 100 cases of bone overgrowth in adults) is likely due to the fact that there was a larger pool of traumatic amputees to evaluate due to OEF and OIF. Additionally military personnel who have access to such records wrote the current article. No evidence of an osteomyoplastic surgical technique in these individuals was seen. However, the authors did mention the use of vacuum assisted drains post operatively. These may contribute to or exacerbate the shear stress in the extracellular fluid and activate the mechanoreceptors of osteoblasts thereby increasing bone tissue in the surrounding musculature. It is important that a relatively small portion of the current population required surgical intervention for relief of pain from the bone overgrowth. The other patients were able to be handled with prosthetic intervention/changes.

Bone overgrowth is present in both surgical techniques, although it is less prevalent in those who have undergone an Ertl procedure. Additionally, the physiological benefit to the bone tissue when the medullary canal is sealed appears to be substantial. Further research is warranted to investigate the long-term outcomes of Ertl and traditional surgeries with regard to bone health and development of bone spurs and bone overgrowth prevalence.

Venous return. Although the focus of this discussion is on bone development and response to amputation, it has also touched on blood flow within the system. Thus it is important to include a brief discussion of the vascular changes associated with both amputation techniques.

Rolf Dederich (1963) provided a compelling arteriogram demonstrating an increase in the vasculature in an above knee amputee following a revision surgery using the Ertl procedure. Prior to and following the surgery, an arteriogram was performed. After just a four-week period, a considerable improvement in vascularity was apparent.

This was further demonstrated in a rabbit model (Hansen-Leth, 1979). Two groups of rabbits had amputations of the proximal crus. One group received a myoplasty and the second did not. After only six weeks, the animals were sacrificed and an arteriogram was performed. The animals who did not receive the myoplasty showed smaller and fewer number of vessels. The myoplasty group showed a large increase in the size and number of vessels compared to the non-myoplasty group. Following up on this study, 31 patients were monitored for blood flow following amputation (Pilegard, Rorbaek Madsen, Hansen-Leth, & Terp, 1985). In fifteen patients, a myoplastic amputation was performed and six had the medullary canal plugged with a bone chip. Blood flow increased by 2/3 in the six patients had their canal closed. While this does not directly relate to bone development, it does suggest there are additional physiological benefits associated with sealing the medullary canal and returning the pressure to normal.

Through this review of the literature, we can see a need for evaluating amputation techniques on functional outcomes of persons with TTA. The analysis of these tasks can shed light on how to improve the quality of life and independence of persons with TTA. Additionally, to improve the quality of the data obtained from inverse dynamics analysis, we must also develop regression equations to predict these values. These equations will enable researchers to quickly calculate the body segment parameters with tools available within their laboratory.

APPENDIX B

EFFECT SIZES

Effect size calculations for each study. Data are shown for significant variables only. Bolded effect sizes denote significant contrasts.

STAND		Burgess Amputated		Burgess Intact			Ertl Amputated		Ertl Intact		Effect Size					
	n	Mean	SD	Mean	SD	N	Mean	SD	Mean	SD	BA vs BI	BA vs EA	EA vs EI	EA vs BI	BA vs EI	EI vs BI
Max Knee angle	7	78.10	8.77	80.87	13.00	11	89.50	7.57	86.87	5.00	0.25	**1.39**	0.41	**0.81**	**1.23**	0.61
GRF BW	7	52.53	8.00	71.06	11.03	11	63.25	8.41	80.29	11.60	**1.92**	**1.31**	**1.68**	0.80	**2.79**	0.82
Knee power	7	0.71	0.34	1.99	0.67	11	1.02	0.53	2.92	1.25	**2.40**	0.70	**1.99**	**1.59**	**2.42**	**0.93**
Hip power	7	1.25	0.73	1.86	0.72	11	2.06	0.85	1.95	1.15	0.83	1.02	0.10	0.25	0.73	0.10
AI Knee	7	0.01	0.01	0.02	0.01	11	0.01	0.00	0.02	0.01	**0.91**	0.56	**2.82**	**1.63**	**1.60**	0.51
AI Hip	7	0.02	0.00	0.03	0.01	11	0.02	0.01	0.02	0.01	1.62	0.18	0.60	0.80	0.66	**1.65**
Total knee work	7	0.25	0.25	0.51	0.21	11	0.19	0.14	0.77	0.25	**1.11**	0.32	**2.89**	**1.80**	**2.09**	**1.16**

Curb		Burgess Amputated		Burgess Intact			Ertl Amputated		Ertl Intact		Effect Size					
	n	Mean	SD	Mean	SD	N	Mean	SD	Mean	SD	BA vs BI	BA vs EA	EA vs EI	EA vs BI	BA vs EI	EI vs BI
Ankle Pos Work	7	0.22	0.13	0.77	0.19	5	0.31	0.21	0.81	0.55	**3.33**	0.50	**1.22**	**2.32**	**1.48**	0.10
Knee Pos Work	7	0.13	0.07	0.47	0.20	5	0.28	0.14	0.54	0.29	**2.29**	1.47	**1.13**	1.08	**1.96**	0.29
Hip Pos Work	7	1.07	0.41	0.88	0.26	5	1.23	0.27	1.01	0.10	0.54	0.48	1.10	1.33	0.19	0.65
Ankle Net Work	7	0.01	0.13	0.57	0.15	5	0.09	0.14	0.57	0.44	**3.86**	0.59	**1.45**	**3.17**	**1.71**	0.01
Knee Net Work	7	-0.56	0.26	-0.11	0.25	5	-0.42	0.27	-0.14	0.19	**1.80**	0.55	1.21	**1.20**	**1.89**	0.14
Hip Net Work	7	0.66	0.46	0.35	0.33	5	0.93	0.35	0.63	0.34	0.79	0.65	0.87	**1.72**	0.08	0.85
Net Limb		0.11	0.53	0.81	0.36	5	0.61	0.43	1.06	0.53	**1.55**	1.03	0.94	**0.52**	**1.79**	0.55

Ground		Burgess Amputated		Burgess Intact			Ertl Amputated		Ertl Intact		Effect Size					
	n	Mean	SD	Mean	SD	N	Mean	SD	Mean	SD	BA vs BI	BA vs EA	EA vs EI	EA vs BI	BA vs EI	EI vs BI
Ankle Neg Work	7	-0.21	0.08	-0.26	0.11	5	-0.19	0.08	-0.31	0.12	**0.53**	0.35	**1.28**	**0.81**	**1.01**	0.45
Ankle Pos Work	7	0.12	0.05	0.86	0.28	5	0.14	0.05	0.85	0.18	**3.66**	0.31	**5.50**	**3.59**	**5.60**	0.05
Ankle Net Work	7	-0.09	0.06	0.60	0.25	5	-0.05	0.04	0.54	0.11	**3.86**	0.84	**7.06**	**3.66**	**7.25**	0.32
Net Limb Work	7	0.11	0.53	0.81	0.36	5	0.61	0.43	1.06	0.53	**1.55**	1.03	**0.94**	**0.52**	**1.79**	0.55

Study I		Specific		Intact		General		Effect Size		
	n	Mean	SD	Mean	SD	Mean	SD	S v I	S v G	G v I
Knee TS	9	-0.28	0.07	-0.40	0.09	-0.27	0.07	**1.52**	0.14	**1.67**
Hip TS	9	0.34	0.10	0.69	0.15	0.38	0.22	**2.70**	0.24	**1.61**
Shank COM	9	0.07	0.02	0.18	0.02	0.09	0.01	**5.99**	1.04	**6.92**
Shank Mass	9	2.70	0.50	3.42	0.84	2.61	0.64	**1.04**	0.16	**1.09**
Shank MOI	9	0.03	0.02	0.03	0.01	0.01	0.00	0.48	0.75	**2.22**
Foot Mass	9	1.12	0.23	1.39	0.30	0.99	0.24	**1.02**	0.58	**1.48**

	Burgess		Ertl		Effect Size
STS Time	13.27	2.83	9.33	2.66	**1.44**

APPENDIX C

INSTITUTIONAL REVIEW BOARD DOCUMENTS

Informed Consent Document

UNIVERSITY of
NORTHERN COLORADO

CONSENT FORM FOR HUMAN PARTICIPANTS IN RESEARCH
UNIVERSITY OF NORTHERN COLORADO

Project Title: **Functional and rehabilitative outcomes after transtibial osteomyoplastic amputation**

Researcher: Jeremy D. Smith Ph.D., School of Sport and Exercise Science
Phone: 970-351-1761 E-mail: Jeremy.Smith@unco.edu

Purpose and Description:

The researchers are performing this study to identify functional outcomes related to transtibial amputations of the lower extremity. You are being asked to participate in this research study because you have had an osteomyoplastic (i.e., Ertl) amputation or a traditional posterior flap amputation. This study will use gait analyses and other functional tests to determine the effectiveness of Ertl and traditional amputation techniques in restoring limb function following surgery. If you join the study, you will be asked to participate in a single 3 hour testing session to measure your muscle strength, gait ability, postural steadiness, overall physical function, and questionnaires. Up to 20 people will participate in the study.

Functional Performance

You will be asked to perform walking, stair climbing, balance, and knee flexibility tests. Electrodes, which measure the electrical activity (EMG) of your muscles, will be attached to the surface of your skin over various leg and gluteal muscles. It may be necessary to shave your hair, lightly abrade and clean your skin with alcohol in the small areas where these electrodes will be attached to improve the quality of the signal.

Muscle Performance

We will measure your muscle performance as you maximally contract the thigh muscles of each leg. This testing will occur with you seated and/or lying down while pushing against a lever with your leg. We will ask you to maximally contract your muscles up to three times. If at any time, discomfort from any of the testing becomes more than you care to tolerate, just let us know and we will stop the testing.

Questionnaires

During the muscle performance assessments you will be asked to remove your prosthesis and liner so that we can take measurements of your prosthesis. During this time, you will be asked to complete a set of questionnaires which ask about your general health and specific questions regarding your amputation.

Motion Analysis

For this testing, a set of reflective markers will be placed on your trunk and legs to measure total body movement. We will then use a set of cameras to record your movements. Once the data are collected, there will be no personally identifiable record of you in the dataset. The tasks you will perform are: Walking 10 meters several times, standing on one leg, and getting up out of a chair. We will measure your postural steadiness by having you stand still on a force plate for approximately 30 seconds under four different conditions: 1) standing with your eyes open, 2)standing with your eyes closed, 3) standing on a piece of foam with your eyes open, 4) standing on a piece of foam with your eyes closed.

What are the possible discomforts or risks?

Though the testing procedures to which you will be exposed are safe, some participants do report some muscle soreness after muscle strength testing for approximately 2 days after testing. This soreness is similar to the muscle soreness that you may feel if you lift weights or vigorously exercise after a long layoff. Although the force levels to be used in this study pose very little risk for injury, possible injuries include musculoskeletal injury or falls. The risks of a fall are no higher than normal daily life. You may develop a skin irritation caused by the adhesive used to attach the reflective markers to your skin or due to the skin preparations that are necessary for EMG electrode placement. The study may include risks that are unknown at this time.

What are the possible benefits of the study?

This study will provide you with feedback regarding your physical performance because you will have measures of your strength and function. This information is in addition to what you would normally be provided by your doctor or physical therapist. This study is designed for the researchers to learn more about the effects of specific amputation techniques on functional outcomes following surgery. This study is not designed to treat any illness or to improve your health. Also, there may be risks, as discussed in the section describing the discomforts or risks.

Will I be paid for being in the study?

You will be given a $50 gift certificate for your participation in this study.

We will take every precaution in order to protect your confidentiality. We will assign a subject number to you. Only the principal investigator and other researchers involved in the project will know the name connected with a subject number and when we report data, your name will not be used. Data collected and analyzed for this study will be kept in a locked cabinet in the Biomechanics Lab, which is only accessible to the research team.

Participation is voluntary. You may decide not to participate in this study and if you begin participation you may still decide to stop and withdraw at any time. Your decision will be respected and will not result in loss of benefits to which you are otherwise entitled. Having read the above and having had an opportunity to ask any questions, please sign below if you would like to participate in this research. A copy of this form will be given to you to retain for future reference. If you have any concerns about your selection or treatment as a research participant, please contact the Office of Sponsored Programs, Kepner Hall, University of Northern Colorado Greeley, CO 80639; 970-351-2161.

____

Subject's Signature   Date

____

Researcher's Signature   Date

______ Please initial here if you agree to allow researchers to use video recordings of your motion in research presentations and educational presentations.

Approval Letter

UNIVERSITY *of* NORTHERN COLORADO

Institutional Review Board

DATE:	August 10, 2015
TO:	Jeremy Smith, PhD
FROM:	University of Northern Colorado (UNCO) IRB
PROJECT TITLE:	[372395-11] Functional and rehabilitative outcomes after transtibial osteomyoplastic amputation
SUBMISSION TYPE:	Continuing Review/Progress Report
ACTION:	APPROVED
APPROVAL DATE:	August 6, 2015
EXPIRATION DATE:	August 6, 2016
REVIEW TYPE:	Expedited Review

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

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

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

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

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

All NON-COMPLIANCE issues or COMPLAINTS regarding this project must be reported promptly to this office.

Based on the risks, this project requires continuing review by this committee on an annual basis. Please use the appropriate forms for this procedure. Your documentation for continuing review must be received with sufficient time for review and continued approval before the expiration date of August 6, 2016.

Please note that all research records must be retained for a minimum of three years after the completion of the project.

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