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Biomechanical Assessment of Ertl and Burgess Transtibial Amputation Techniques

Abbie Elthea Ferris

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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

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

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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

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ABSTRACT

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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.

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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, Trivison, & 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, Guitierrez, & 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 limb 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, Yekutieli, & 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 descent, 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, Grootsholten, & Stam, 2004; Bussmann, Schrauwen, & Stam, 2008). Researchers have found during sit-to-stand, patients transfer weight towards the unaffected leg (Agrawal, Gailey, Gaunaard, Gailey, & O'Toole, 2011; Ozyurek, Demirbiken, & Angin, 2013). Agrawal, Gailey, O'Toole, Gaunaard, 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, Reflex, 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).

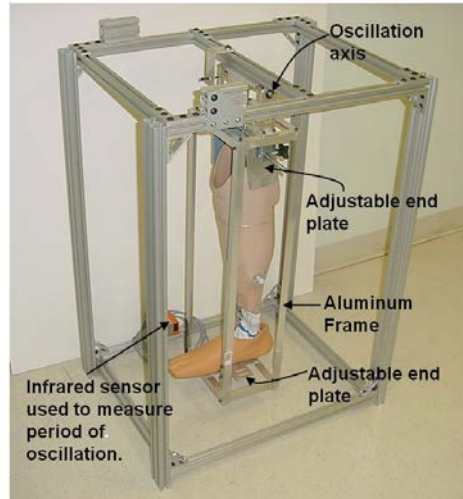


Figure 2.1. Oscillation rack configuration for measuring the period of oscillation of the prosthetic limb. Adapted from J. D. Smith et al. (2014)

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 ($-9.8 \text{ m}\cdot\text{s}^{-2}$), and d was the distance from the axis of rotation to the center of mass location.

The range of motion for this oscillation 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).

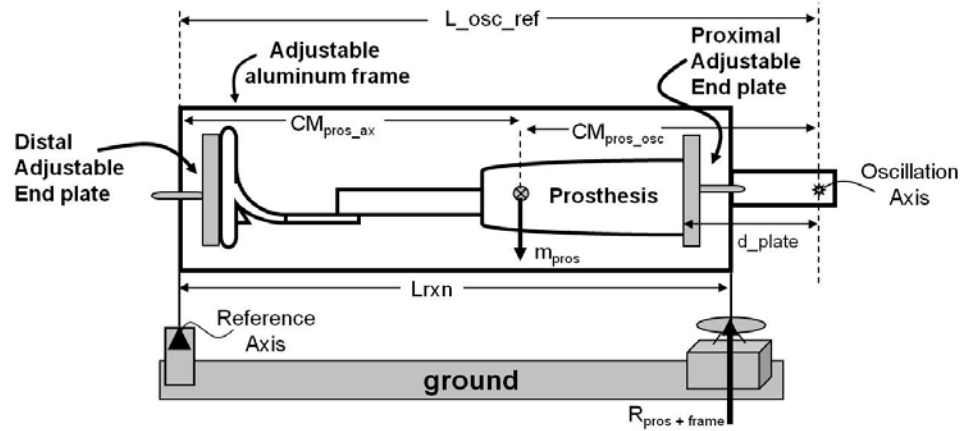


Figure 2.2. Experimental set for the reaction board test to calculate the prosthetic center of mass location. Adapted from J. D. Smith et al. (2014)

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:

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

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: $r = \left(\frac{C}{2 \cdot \pi}\right)$

Proximal radius: $R = \frac{0.255 \text{ m}}{2 \cdot \pi} = 0.041 \text{ m}$

Distal radius: $r = \frac{0.210 \text{ m}}{2 \cdot \pi} = 0.033 \text{ m}$

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

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

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

$d^* = 1100 \text{ kg} \cdot \text{m}^{-3}$

$$M_{residual} = d * Volume$$

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

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

Step 3: Calculate Adjusted Body Mass (ABM)

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

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

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

$$Shank \text{ mass} = 0.035 * 62.29 \text{ kg} = 2.18 \text{ kg}$$

Foot % mass: 1.4% ABM

$$Foot \text{ mass} = 0.014 * 62.29 \text{ kg} = 0.87 \text{ kg}$$

Step 5: Calculate segment COM locations

Shank COM: 21% shank length (ℓ_{shank})

$$Shank \text{ length} = 0.21 * 0.36 \text{ m} = 0.08 \text{ m}$$

Foot COM: 40.14% foot length (ℓ_{foot})

$$Foot \text{ length} = 0.4014 * 0.26 \text{ m} = 0.10 \text{ m}$$

Step 6: Calculate segment radius of gyration (k)

Shank ROG = 19.4% segment length (ℓ_{shank})

$$Shank \text{ ROG} = 0.194 * 0.36 \text{ m} = 0.07 \text{ m}$$

Foot ROG = 24.5% segment length (ℓ_{foot})

$$Foot \text{ ROG} = 0.245 * 0.26 \text{ m} = 0.06 \text{ m}$$

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

$$MOI: I = m \cdot k^2$$

$$Shank \text{ MOI} = 2.18 \text{ kg} * (0.072 \text{ m})^2 = 0.011 \text{ kg} \cdot \text{m}^2$$

$$Foot \text{ MOI} = 0.87 \text{ kg} * (0.062 \text{ m})^2 = 0.003 \text{ kg} \cdot \text{m}^2$$

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 $1.5 \text{ m}\cdot\text{s}^{-1}$ 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 \pm 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 \pm SD		0.07 \pm 0.02	0.08 \pm 0.01	0.18 \pm 0.02	2.70 \pm 0.51	2.61 \pm 0.64	3.42 \pm 0.84	0.026 \pm 0.024	0.013 \pm 0.005	0.035 \pm 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 \pm SD		0.05 \pm 0.01	0.05 \pm 0.01	0.05 \pm 0.01	1.12 \pm 0.23	0.99 \pm 0.24	1.40 \pm 0.27	0.008 \pm 0.005	0.007 \pm 0.006	0.008 \pm 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·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 ($p < .05$)

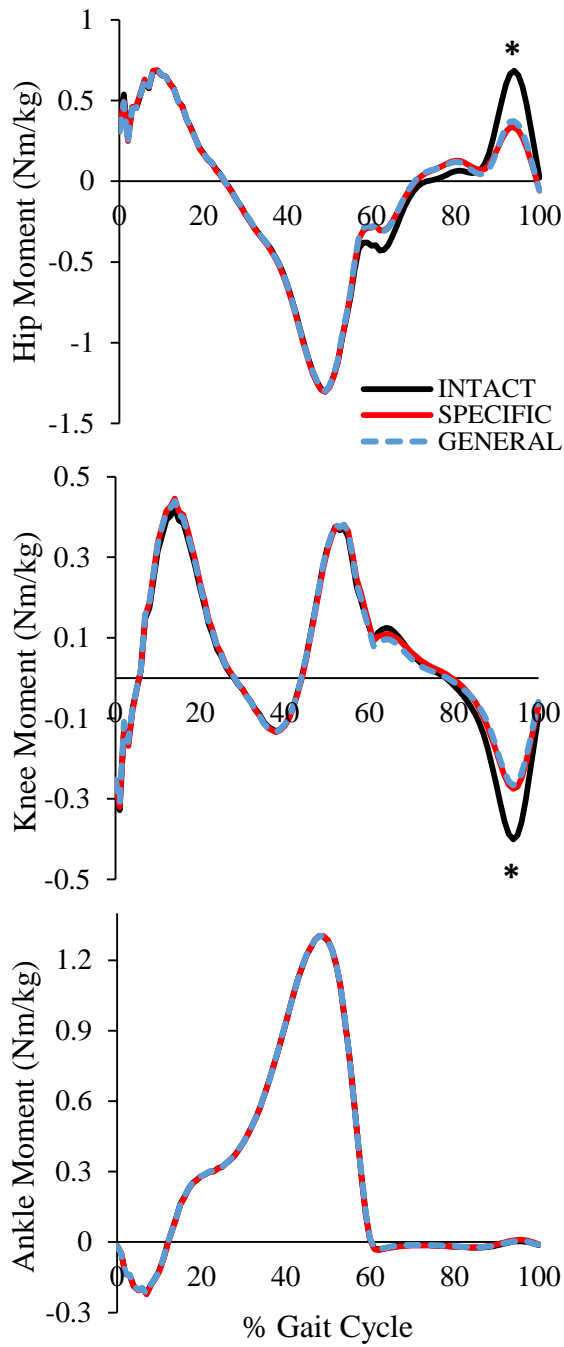


Figure 2.4. Mean hip, knee, and ankle joint moments of 9 TTAs using INTACT, SPECIFIC, and GENERAL inertial models. *Indicates INTACT measures were significantly different from the SPECIFIC and GENERAL models ($p < .05$). No Significant differences were found between SPECIFIC and GENERAL models.

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 negotiation 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, Zadavec, & 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).

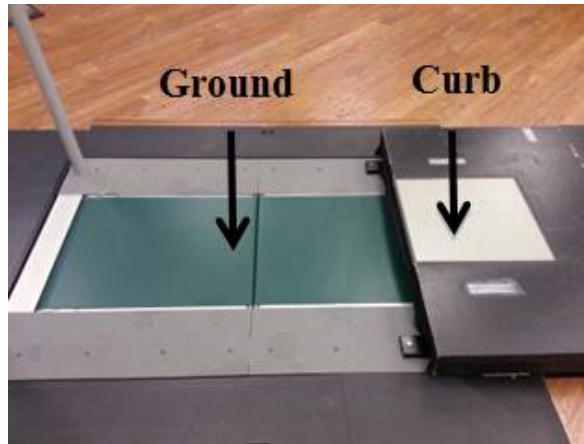


Figure 3.1. Illustration of the curb design. Two forces plates were embedded in the ground and a single force place mounted above the ground level created the curb.

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 $\pm 5\%$ 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 amputated 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 \pm SD.

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

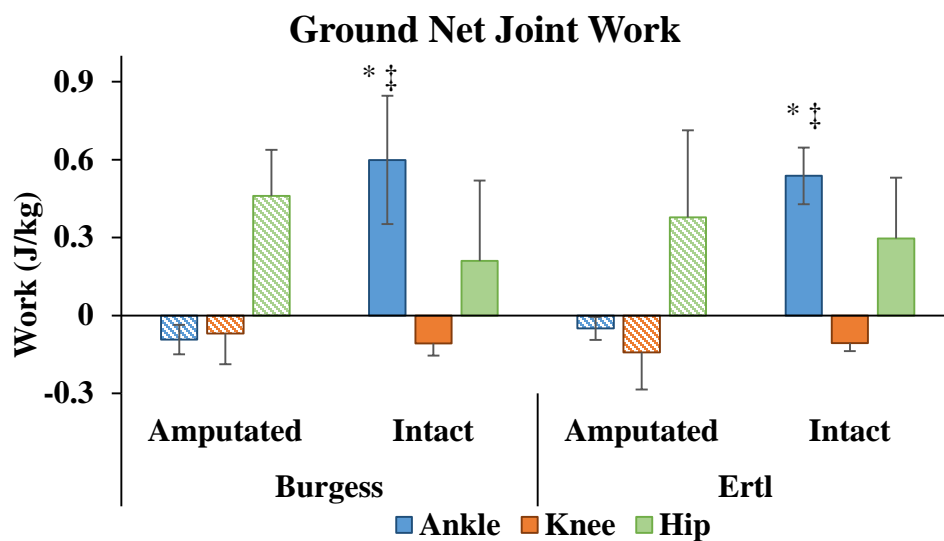


Figure 3.2. Net joint work for the ankle, knee, and hip for the GROUND step. *Significantly different from Ertl amputated limb; †Significantly different from Burgess amputated limb.

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).

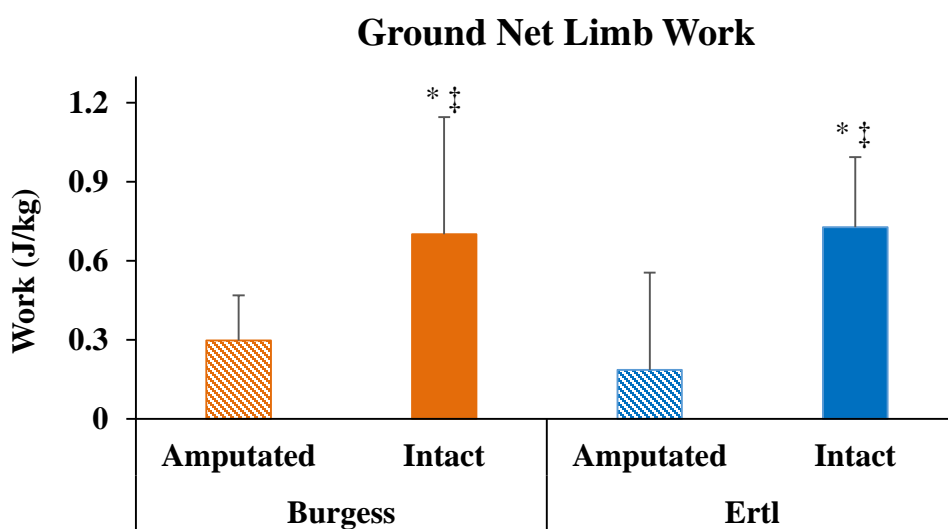


Figure 3.3. Net limb work for the entire limb for the GROUND step. *Significantly different from Ertl amputated limb; †Significantly different from Burgess amputated limb

CURB Step Results

A significant model effect was found for the CURB step (Wilk's Lambda: $p = .0550$, $F(33,30.166) = 1.79$, Hotelling-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 ($\text{J} \cdot \text{kg}^{-1}$) on the CURB step. Data shown mean \pm SD.

<u>Group</u>	<u>Amputated</u>			<u>Intact</u>		
	<u>Ankle</u>	<u>Knee</u>	<u>Hip</u>	<u>Ankle</u>	<u>Knee</u>	<u>Hip</u>
<u>Positive Work</u>						
Ertl	0.31 ± 0.21	0.28 ± 0.14	1.23 ± 0.27	$0.81 \pm 0.55^{* \ddagger}$	$0.54 \pm 0.29^{* \ddagger}$	1.01 ± 0.10
Burgess	0.22 ± 0.13	0.13 ± 0.07	1.07 ± 0.41	$0.77 \pm 0.19^{* \ddagger}$	$0.47 \pm 0.20^{\ddagger}$	0.88 ± 0.26
<u>Negative Work</u>						
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.

\ddagger 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.

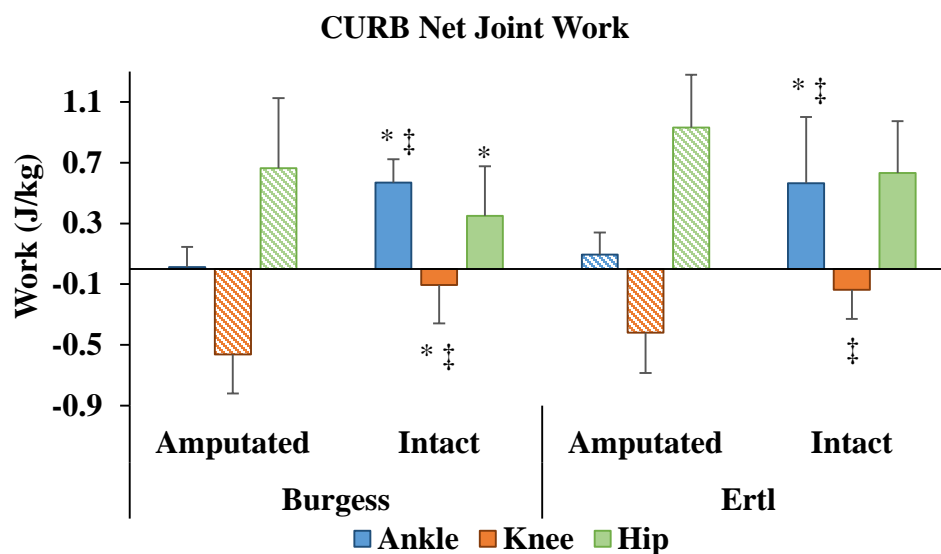


Figure 3.4. Net joint work for the ankle, knee, and hip for the CURB step. *Significantly different from Ertl amputated limb; ‡Significantly different from Burgess amputated limb.

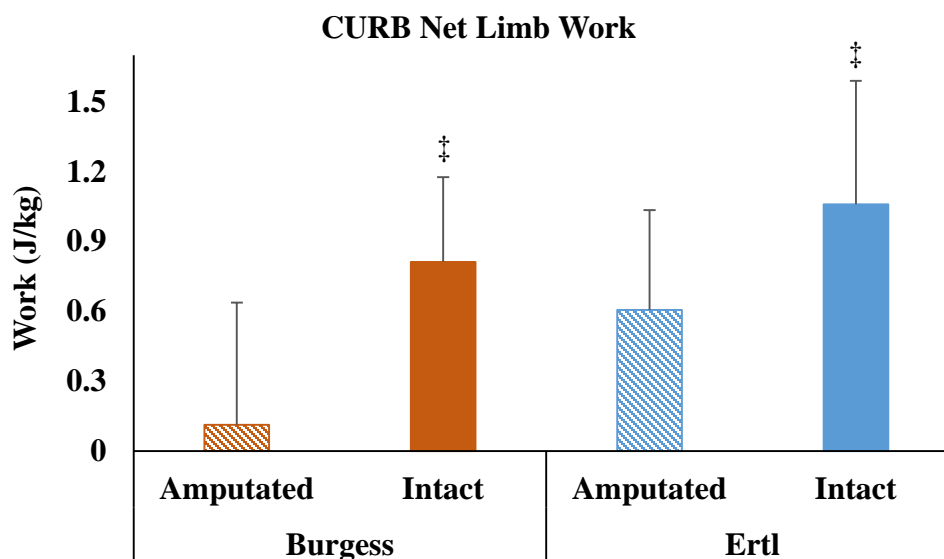


Figure 3.5. Net joint work for entire limb for the curb step. ‡Significantly different from Burgess amputated limb.

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.

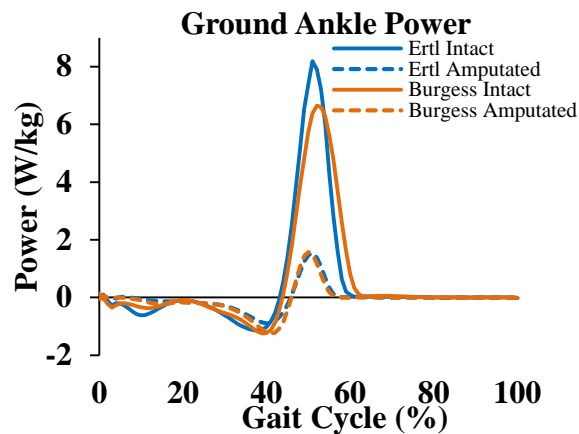


Figure 3.6. Ankle power for the GROUND step. Data are shown from heel strike to heel strike.

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.

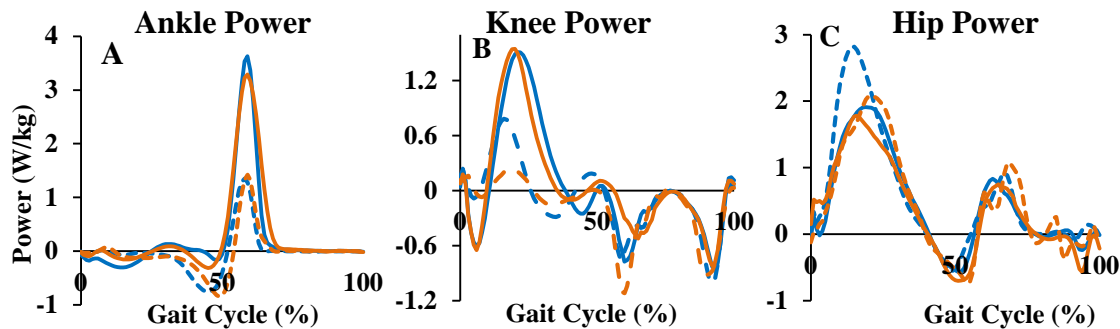


Figure 3.7. Joint powers for the ankle (A), knee (B) and hip (C) for the curb step. (—) Ertl intact, (—) Burgess intact, (---) Ertl amputated, (---) Burgess amputated.

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, Demirboken, & 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, previous 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 kinematically. 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 $\sim 0^\circ$ flexion (full extension). The maximum knee flexion angle of the Ertl amputated limb was significantly larger ($\sim 10^\circ$) 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 ($\sim 8^\circ$) 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 ($\sim 80^\circ$ Burgess vs $\sim 88^\circ$ 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.

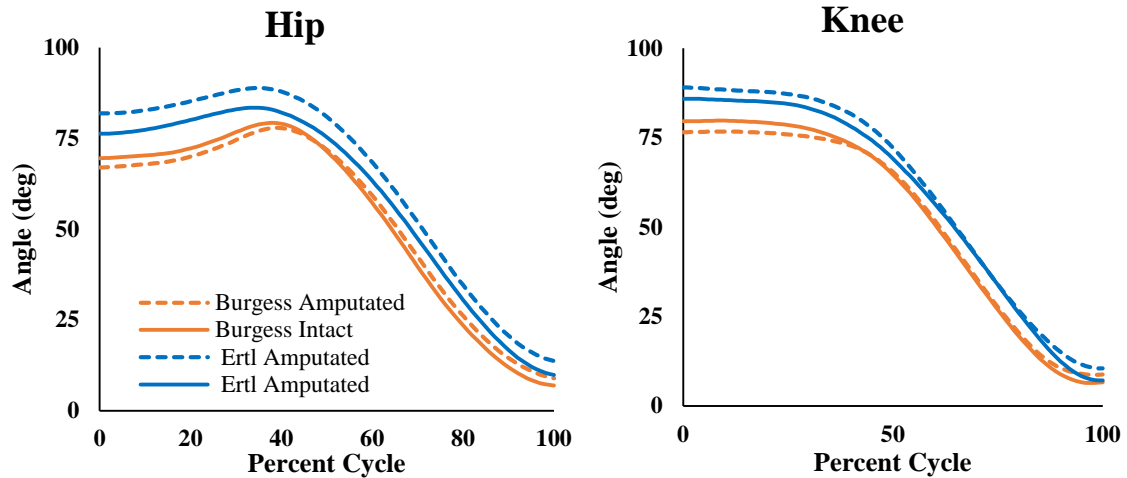


Figure 4.1. Mean sagittal hip and knee joint angles during the STAND task. 0 degrees indicates full extension at the hip and knee.

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 \pm SD.

Group	Amputated GRF (%BW)	Intact GRF (%BW)	SI
Ertl	63.3 \pm 8.4	80.3 \pm 11.6* [‡]	88.34 \pm 11.92
Burgess	52.5 \pm 8.0*	71.1 \pm 11.0 [‡]	85.15 \pm 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 ($Nm \cdot s \cdot BW^{-1}$) of the ankle, knee, and hip during STAND. Data are shown mean \pm SD.

Group	Amputated			Intact		
	Ankle	Knee	Hip	Ankle	Knee	Hip
Ertl	0.011 \pm 0.004	0.009 \pm 0.005	0.023 \pm 0.009	0.008 \pm 0.004	0.024 \pm 0.006* [‡]	0.019 \pm 0.007
Burgess	0.008 \pm 0.004	0.013 \pm 0.008	0.022 \pm 0.003	0.012 \pm 0.004	0.020 \pm 0.008* [‡]	0.029 \pm 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.

<u>Group</u>	<u>Amputated</u>		<u>Intact</u>	
	Knee	Hip	Knee	Hip
Ertl	0.20 ± 0.14	0.65 ± 0.28	0.76 ± 0.25 ^{*‡}	0.51 ± 0.33
Burgess	0.26 ± 0.24	0.42 ± 0.21	0.51 ± 0.21 ^{*‡†}	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 \pm SD.

Group	Amputated		Intact	
	Knee	Hip	Knee	Hip
Ertl	1.02 \pm 0.53	2.06 \pm 0.85	2.92 \pm 1.25 ^{*‡}	1.95 \pm 1.15
Burgess	0.71 \pm 0.34	1.25 \pm 0.73	1.99 \pm 0.67 ^{*‡†}	1.86 \pm 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 ($p < .05$)

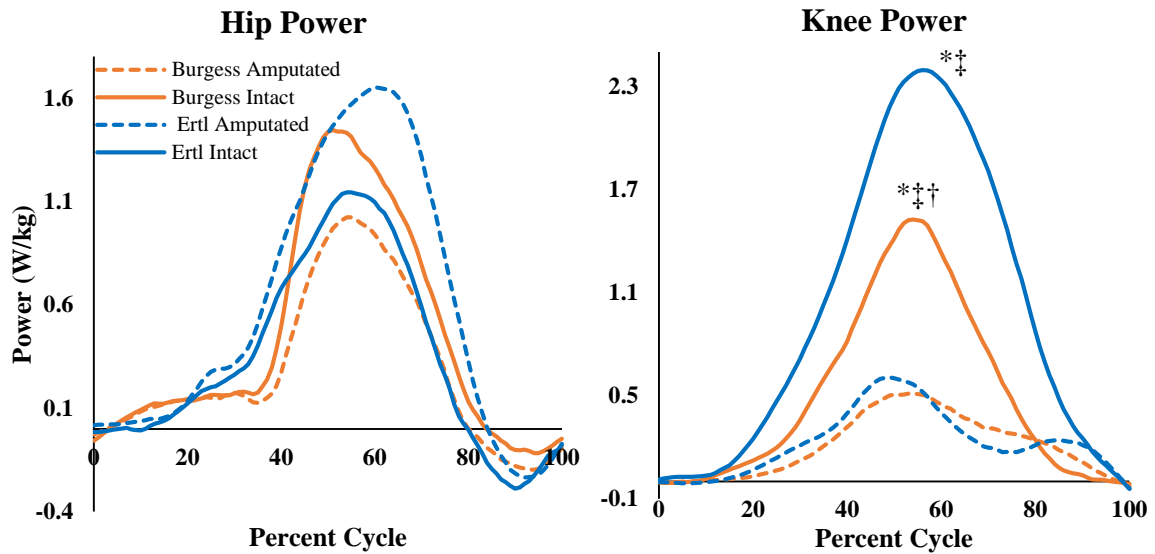


Figure 4.2. Sagittal plane joint powers for the hip and knee joints during the STAND task. At 0 % of the cycle the person is sitting and at 100% of the cycle, the person is fully standing. *Significantly different from Ertl amputated limb; †Significantly different from Burgess amputated limb; ‡Significantly 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 ~ 0.1 s. 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; Roebroek 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.

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enable researchers to quickly calculate the body segment parameters with tools available within their laboratory.

APPENDIX B

EFFECT SIZES

APPENDIX C
INSTITUTIONAL REVIEW BOARD DOCUMENTS

