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Alterations in gait patterns are commonly observed in individuals with transtibial amputation (TTA) who use a prosthesis. Current commercially available ankle-foot prostheses (AFP) offer very little range of motion (ROM) at the ankle joint. Previous researchers have hypothesized that lack of ankle ROM significantly contributes to alterations in TTA gait patterns. However, different patterns have been observed among TTA using the same AFP. Therefore it is unclear how restricted ankle ROM in current commercially available ankle-foot prostheses (AFP) contributes to observed changes in gait. Alterations in gait patterns have been shown to increase the incidence of low back pain and other musculoskeletal injuries. TTA have a greater incidence of low back pain and osteoarthritis of the knee and hip. Therefore it is important for researchers to understand the influence of different prosthetic components on gait in order to optimize gait patterns and minimize complications due to alterations in gait.

The purpose of this study was to determine what compensatory alterations in gait patterns may occur as a result of imposed restricted ankle range of motion. Kinematic data was collected from 19 participants (9 men, 10 women) age 18-32 with no previous history of lower extremity injury or deformity in two conditions: level-ground walking with no restriction and level ground walking with the ankle restricted at 0 degrees plantarflexion by plaster casting. Results indicated that restricted ankle ROM contributes to decreased velocity and cadence and decrease in gait symmetry. A compensatory pattern was observed for pelvic obliquity, hip and knee flexion at toe-off and foot progression angle. Observed patterns did not resemble those observed in TTA. Results suggest that restricted ankle ROM contributes to some components of alterations in gait patterns observed in TTA. However a combination of other components, including loss of proprioception and power generation at the metatarsophalangeal (toe) joints may have a more significant contribution to TTA gait patterns than restricted ankle ROM alone.

The Effects of Restricted Ankle Range of Motion on Human Walking: An Application to Transtibial Amputee Gait

by

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I understand that my thesis will become part of the permanent collection of Oregon State University libraries. My signature below authorizes release of my thesis to any reader upon request.

Susan R. Silverman, Author

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CONTRIBUTION OF AUTHORS

Dr. JoonKoo Yun was involved with the conceptualization of the study and research design, data analysis, interpretation of results, and reviewing of the thesis.

Dr. Mike Pavol was involved with the creation of the biomechanical analysis model, data analysis and interpretation of results.

Effects of Restricted Ankle Range of Motion on Human Walking: An Application to Transtibial Amputee Gait Patterns

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Chapter 1: INTRODUCTION

Human walking is a cyclical and reciprocating pattern which allows for energy-efficient ambulation. Impairments of the human body may create compensatory motions that disrupt the gait pattern (Rose & Gamble, 1994). When gait patterns are altered, walking becomes less efficient and may lead to musculoskeletal impairments (Isakov, Keren, & Benjuya, 2000; Khodadadeh, Eisenstein, Summers, & Patrick, 1988).

The term transtibial amputee (TTA) describes an individual who has undergone amputation of the leg below the knee joint. Each year approximately 185,000 Americans undergo lower limb amputation (Ephraim, Wegener, MacKenzie, Dillingham, & Pezzin, 2005). Transtibial amputees (TTA), as can be expected due to limb loss, have shown alterations in gait pattern (Mattes, Martin, & Royer, 2000).

Alterations in gait patterns can be a significant problem in TTA, due to the associated risk of secondary musculoskeletal impairments, including low back pain (LBP) and knee and hip osteoarthritis, which has been observed to have a greater incidence in TTA than in the general population (Edhe et al., 2001; Ephraim et al., 2005; Gailey, Allen, Castles, Kucharik, & Roeder, 2008). TTA also have a significantly greater metabolic cost of walking (Mattes et al., 2000). The increased metabolic cost of walking for TTA has been attributed to differences in spatiotemporal parameters of gait (Highsmith, Schulz, Hart-Hughes, Latlief, & Philips, 2010).

Current commercially available ankle-foot prostheses (AFP) have a decreased ankle range of motion (ROM) compared to an anatomical ankle joint, which previous researchers have hypothesized disrupts typical gait patterns (Au, Herr, Weber, & Martinez-Villalpando, 2007). Winter and Sienko (1988) reported common compensatory gait patterns for individuals with TTA. This study noted, most predominately, hyperactive hip extensors during the toe-off phase. This was believed to be a compensation for the loss of power generation by the ankle plantarflexors. Concerning spatiotemporal parameters of gait, TTA have been shown to have a longer prosthetic limb swing time, longer step length and shorter stance time than for the intact limb (Mattes et al., 2000). Alterations in spatiotemporal parameters of gait have been observed, but the hypothesis that lack of ankle ROM plays a major role has not been confirmed.

The cyclical and reciprocating pattern of human gait also functions symmetrically (Rose & Gamble, 1994). Asymmetry in gait has been shown to be a relevant measure for investigating gait patterns and predicting future joint pain and degradation (Isakov et al., 2000). Disruptions in gait symmetry also increase the metabolic cost of walking by decreasing gait efficiency (Khodadadeh et al., 1988). Gait symmetry may decrease as a result of compensatory gait patterns. TTA have been observed to walk with at least one compensatory alteration in the gait pattern, leading to decreased gait symmetry (Mattes et al., 2000). Similarly to spatiotemporal differences in gait, lack of ankle ROM in the AFP has been hypothesized to be a major contributor to compensatory patterns and decreased gait symmetry (Au et al., 2007).

Alterations in spatiotemporal parameters, gait patterns and symmetry have been observed in the TTA population, it being hypothesized that lack of ankle ROM is a major contributor. However, individuals using the same prosthetic components can display different resulting gait patterns (Su, Gard, Lipschutz, & Kuiken, 2008). Previous research has reported that differences in gait patterns exist between vascular amputees and traumatic amputees. Hip-hike of the involved side has been associated with traumatic amputees, while a steppage gait (increased knee and hip flexion during swing), has been associated with vascular amputees (Mattes et al., 2000; Michaud, Gard, & Childress, 2000; Winter & Sienko, 1988). If the hypothesis is true that lack of ankle ROM is a primary contributor to alterations in gait patterns is true, then differences in gait patterns between individuals with traumatic and vascular amputation are not adequately explained. Dynamic Systems Theory stresses nonlinear connections in human development and the capacity of a system to reorganize in response to stimuli (Smith & Thelen, 1993). This would imply that there are possible influences of system components other than solely ankle ROM in determining the spatiotemporal parameters of gait and gait patterns.

The purpose of this study was to examine alterations in gait patterns as a result of imposed restricted ankle range of motion. Restricted ankle range of motion in individuals without TTA may create similar patterns to vascular or traumatic transtibial amputees, or cause participants to create other compensatory patterns. Differences in observed gait patterns from those seen in TTA may suggest that changes in other systems of the body have a greater effect on TTA gait than ankle ROM (Klute, Kallfelz, & Czerniecki, 2001).

Specific Research Questions

The following questions were investigated in this study:

Specific research question 1: how does restricted ankle ROM affect spatiotemporal parameters of gait?

Specific research question 2: how does restricted ankle ROM affect gait symmetry?

Specific research question 3: What pattern does restricted ankle ROM create during gait?

Assumptions and Limitations

The following assumptions were made for this study:

- Participants will walk in a consistent manner across all trials for each condition.
- The intact limb of TTA functions the same as both limbs of individuals without TTA
- Plaster casting will restrict ankle ROM similarly across all participants

The following limitations are noted for this study:

- ROM at the metatarsophalangeal joints (ball of the foot) was not restricted
    - Force generation at the toes was not controlled
- Participants retain proprioceptive sense in both limbs

The study was delimited as follows:

- 19 participants (9 men, 10 women)
- Age of participants 18-32 years

- Ankle ROM restricted on the dominant side

Operational Definitions

1. Gait symmetry- equality of joint angles and limb dependent spatiotemporal parameters of gait when compared bilaterally.
2. Compensatory pattern**-** Differences in joint angles observed on the same side of the body between the two conditions.

Chapter 2: REVIEW OF LITERATURE

The purpose of this literature review was to discuss the essential components of human walking, the alterations in gait patterns which occur due to limitations on the body and the effects of alterations in gait pattern on the human body. This review of literature is in support of a study which will examine the effects of restricted ankle range of motion on walking patterns, as it relates to walking patterns in transtibial amputees (TTA) who utilize a prosthetic limb for walking. The particular topics of interest included in this review are human walking and gait patterns, alterations in gait patterns due to constraints on the human body, lower limb prosthetic design and function, transtibial amputee gait and secondary conditions associated with altered gait patterns.

Human Walking

In the general population, bipedal walking is the primary means of locomotion. The series of motions involved in human walking are referred to as the "gait cycle" (Kuo, 2007; Rose & Gamble, 1994; Whittle, 2007). One gait cycle is defined as the time from heel strike of one leg to repeat heel strike of the same leg (Whittle, 2007). This is also referred to as a stride. The gait cycle is broken down into two components, the stance phase and the swing phase. These two phases can also be further broken down (Rose & Gamble, 1994). Within the stance phase there are two periods of double limb support when the stance leg is in a loading phase and accepting the weight transfer of the body while the opposite leg is unloading and preparing for the swing phase. Once the weight transfer has been completed there is a period of single limb support where the stance leg supports the entire weight of the body, the body's center of mass (COM) is accelerated forward and the opposite leg swings through, further assisting with propulsion of the COM forward. The swing phase can also be broken down into three parts: the initial swing which is meant to achieve foot clearance, mid-swing when the limb advances in front of the body, and terminal swing where the limb is decelerated and prepares for weight transfer and acceptance.

The "Six Determinants of Gait" theory (Saunders, Inman, & Eberhardt, 1953) describes human walking as a cyclical and reciprocating activity. The six determinants of gait theory has proposed that the body has an inherent system for ambulation which is meant to minimize the displacement of the body's COM, which allows for ambulation with minimal energy cost (Kuo, 2007; Whittle, 2007). As discussed by Whittle in "Gait Analysis" the six determinants of gait are: pelvic rotation, pelvic obliquity, knee flexion in stance, ankle rocker, foot rocker and lateral displacement of the body.

The motions of the pelvis comprise the first two determinants of gait. Pelvic rotation is the rotation of the pelvis about a superior-inferior center line. Pelvic rotation allows the individual to achieve adequate step length with less hip flexion and extension (Whittle, 2007). Pelvic obliquity is defined as the amount of lateral tilt of the pelvis occurring from the transverse plane (Michaud et al., 2000). During human walking the hip of the swing leg should be lower than the stance leg. If the pelvis were to keep level then the trunk and COM would follow a greater up and down movement during walking. In order for the pelvis to tilt during the gait cycle the knee and ankle must have sufficient motion to shorten the leg for ground clearance (Whittle, 2007).

The third, fourth and fifth determinants of gait are all concerned with adjusting the effective length of the stance leg by lengthening at the beginning and end of the stance phase and shortening at the mid-point in order to keep hip height relatively constant and smooth the transition of the body's COM from step to step (Whittle, 2007). The third determinant of gait, knee flexion in stance, is meant to shorten the effective leg length at mid-stance to reduce hip height. The fourth determinant, ankle rocker, lengthens the effective leg length at heel strike during the loading response while foot rocker, the fifth determinant of gait, lengthens the effective leg length again during heel-rise and toe-off (Whittle, 2007).

The sixth and final determinant of gait is minimal lateral displacement of the body. The walking base is kept narrow in order to minimize lateral displacement of the COM and preserve balance. This minimization of lateral motion reduces muscular energy associated with balance and conserves energy during walking (Whittle, 2007).

Using the six determinants of gait as a framework, the typical movement patterns of the lower body can be evaluated. Each joint and motion segment of the lower body must possess a certain range of motion (ROM) and level of strength in order to follow the six determinants of gait, which produces an energy efficient and smooth pattern of walking. Rose and Gamble (1994) utilized gait analysis to break down the functions of the lower extremities. At the pelvis, pelvic obliquity reaches its peak just after opposite toe-off with a small secondary rise during swing limb acceleration (Rose & Gamble, 1994). Pelvic obliquity functions as a shock absorber and limb length adjustment during walking with an observed inferior tilt of 5-8^{o} on the swing side (Kadaba, Ramakrishnan, & Wootten, 1990; Michaud et al., 2000). The motions of the hip are an abduction motion during the first portion of the swing phase, then adduction during the terminal swing phase with an average hip adduction of approximately 12 degrees (Kadaba et al., 1990). The hip achieves maximal flexion at terminal swing and slight extension of the stance side hip before heel strike of the swing side hip with an average hip flexion angle of approximately 43 degrees (Kadaba et al., 1990).

During human walking the knee functions as a shock absorber and as a means of adjusting leg length to prevent excessive vertical translation of the COM (Kuo, 2007; Whittle, 2007). Knee flexion occurs during the swing phase to assist in ground clearance of the swing side foot. Passive knee extension occurs with eccentric plantarflexion at heel strike to decelerate the limb and prepare for weight transfer (Rose & Gamble, 1994).

In human walking the ankle behaves like a spring with variable stiffness which provides energy for toe-off and proprioception to control foot orientation during the swing phase (Au et al., 2007). The ankle must plantarflex and dorsiflex twice during each gait cycle and the musculature about the ankle produces approximately five times more energy than is stored during the gait cycle (Versluys et al., 2009). The plantarflexion/dorsiflexion motion of the ankle creates ankle and foot rocker determinants in the six determinants of gait theory (Rose & Gamble, 1994; Whittle, 2007). The ankle is essentially neutral at heel strike followed by controlled plantarflexion to a flat foot position (Neptune, Kautz, & Zajac, 2001). There is then an eccentric plantarflexion contraction to the point of toe-off where the plantarflexors contract concentrically at terminal stance (Rose & Gamble, 1994). The plantarflexors play two major roles in walking according to Winter and Sienko (1988). These two roles are: controlling the forward rotation of the leg over the foot, and generation of mechanical power at toe-off. The plantarflexors generate 80-85% of the total mechanical power generated during the gait cycle (Neptune et al., 2001; Versluys et al., 2009; Winter & Sienko, 1988). A study of individuals after a lateral ligament sprain found that symmetry values in walking decreased in pathological ankles with less than eight degrees of dorsiflexion (Crosbie, Green & Refshauge, 1999). A physiologically normal dorsiflexion angle in adults is approximately 18 degrees with approximately 10 degrees of dorsiflexion required for symmetrical level walking (Crosbie, Green & Refshauge, 1999).

Tibial and foot rotations are also observed in connection with ankle motion during human walking. The tibia has a peak external rotation at toe-off while the foot is fixed in external rotation until heel-rise, remaining in external rotation in initial swing and then internally rotating slightly during terminal swing in preparation for heel strike (Rose & Gamble, 1994). The angle of the foot during the swing phase has been referred to as the “foot progression angle” which is determined as a transverse angle between the line of progression and the longitudinal axis of the foot. The typical foot progression angle is -3±5° (Grumillier, Martinet, Paysant, André, & Beyaert, 2008).

Alterations in Gait Patterns

Multiple constraints of systems and limitations present in the body may result in alterations in gait patterns (Kadaba et al., 1990; Sadeghi, Allard, Prince, & Labelle, 2000). Major alterations in gait patterns, which occur due to disease, trauma, degeneration, fatigue or pain, can lead to secondary conditions and further compensatory actions (Rose & Gamble, 1994). This is in agreement with Sanderson and Martin (1996) who stated that asymmetrical gait has been linked to increased prevalence of degenerative changes in the lumbar spine and knees. The gait cycle is meant to function as a cyclical, reciprocating, low-level loading action. If there is a disruption in the reciprocating action then walking becomes a series of static actions which create higher levels of impact and changes in muscle function (McGill, 2007). Alterations in gait patterns also lead to changes in gait symmetry. Gait in individuals without any existing pathologies has been shown to be symmetrical, with degrees of asymmetry occurring in pathological gait (Sadeghi et al., 2000). Gait changes may be categorized as either forced or compensatory changes (Rose & Gamble, 1994). A forced change is imposed upon the body due to injury, degeneration, etc. which cannot be altered by the individual and is the primary catalyst for the gait change. Compensatory gait changes result from loss of strength, flexibility or endurance, which may result from a forced change, and results in one or more alterations in the gait pattern (Rose & Gamble, 1994).

Using the “six determinants of gait” the importance of leg length adjustment during the gait cycle can be observed. These adjustments are created by the ankle and the knee (Whittle, 2007). Whittle documents some common gait changes which occur due to leg length discrepancies. These alterations in the gait pattern may be observed due to changes in the ankle or knee which prevent adjustment of functional leg length and have also been observed in the TTA population. Circumduction occurs when the swing leg is swung in an arcing motion away from the center line of the body, increasing the hip abduction angle during swing phase. Hip hiking is a reversal of the second determinant of gait, pelvic obliquity. When an individual hip-hikes the pelvis on the swing side is lifted for ground clearance. Steppage occurs when the knee and hip are flexed to a greater degree than in the typical gait pattern in order to achieve proper ground clearance.

Lower Limb Prosthetic Design

Understanding the implications for gait alterations in individuals with TTA requires an understanding of the basic components of an ankle-foot prosthesis (AFP). There are multiple variations in design which have and are currently changing as technology improves and the amputee population changes.

Design and function of AFPs are based on the three “C”s, control, comfort and cosmetics (Versluys et al., 2009). Individuals who utilize a prosthetic device rate the ability to walk with comfort in a variety of conditions as the most important aspect of their fitting (Legro et al., 1999). AFPs are lightweight, passive structures, designed to have appropriate elasticity during the stance phase of walking (Eilenberg, Geyer, & Herr, 2010). AFPs have a base of support when the wearer stands or is in the stance phase, provide shock absorption at heel strike with a plantarflexion motion and model a passive metatarsophalangeal joint action during late stance phase (Edelstein, 1988).

There are two commons categories of AFPs, articulated and non-articulated assemblies (Edelstein, 1988). Articulated assemblies have a cleft corresponding to the anatomical ankle joint and may have a single or multi-axis joint with bumpers limiting the motion at these passive joints (Tang et al., 2008; Versluys et al., 2009). Hydraulic ankles also fall into this category, though no commercially available product exists at this time. Non-articulated assemblies have a continuous external surface from sole to shank (Edelstein, 1988). The most common AFP in this category is the solid ankle, cushion heel (SACH) foot, which has an immobile ankle joint, a soft heel which creates a plantarflexion motion at heel strike, and a flexible forefoot to allow for a toe extension motion in late stance (Edelstein, 1988; Tang et al., 2008). Also in this category is the stationary attachment, flexible endoskeleton (SAFE) foot which functions similarly to the SACH foot with more motion in the mid and forefoot (Edelstein, 1988). Energy store and release (ESR) feet are a more recent development in the field of AFPs (Edelstein, 1988; Versluys et al., 2009). ESR feet have been in recent development due to higher activity levels in traumatic amputees who were active prior to amputation (Versluys et al., 2009). While these feet may provide some energy store and release and the name suggests, the energy loss remains high and return of the stored energy at toe-off occurs later in the gait cycle than would occur in a natural human foot (Versluys et al., 2009). These feet also have a stationary attachment that does not allow for motion at the ankle.

For all AFP there is a pylon which connects the foot to the socket. Recent development in prosthetic design has utilized a vertical shock absorbing pylon (VSAP) (Klute et al., 2001). These devices attempt to reduce shock loads to increase comfort and attempt to reproduce the normal elastic energy-storing mechanisms of the gastrocnemius-soleus complex. The socket type also varies among TTA. The standard design is a patellar-tendon bearing prosthesis (Tang et al., 2008). TTA also have the option of a total surface bearing socket or Icelandic Roll-On Silicone Socket (ICEROSS) which disperse pressure over a greater surface area of soft tissue structures (Tang et al., 2008).

During walking the AFP functions in a passive manner (Au et al., 2007). None of the current commercially available feet are able to provide net positive work, and torque values at toe-off have been seen to be approximately 2/3 of the plantarflexion torque noted during intact walking (Au et al., 2007; Versluys et al., 2009). The metabolic cost of walking for individuals utilizing an AFP has been well documented to be greater than that of non-amputees at approximately 25-50% greater cost (Barth, Schumacher, & Thomas, 1992; Schmalz, Blumentritt, & Jarasch, 2002; Torburn, Powers, Guiterrez, & Perry, 1995; Waters & Mulroy, 1999). To date no significant differences in energy consumption have been seen in previous research at normal walking speeds with different types of AFPs (Barth et al., 1992; Schmalz et al., 2002; Torburn et al., 1995) allowing prescription of AFPs to remain largely subjective and dependent upon the practitioner and general comfort or wishes of the user (Hafner, 2005). Few discriminate effects in energy consumption and gait symmetry have been seen in different types of feet as well as sockets, socket/stump interfaces and limb mass (van der Linde et al., 2004). In general, SAFE and ESR feet are prescribed for more active individuals since they are more adaptable to altered terrain like grass and gravel, while SACH feet are prescribed for the elderly and slow walkers (Versluys et al., 2009).

Transtibial Amputee Gait and Associated Secondary Conditions

Walking with currently prescribed AFPs may cause discomfort and early fatigue in TTA (Versluys et al., 2009). The gait pattern of TTA has been shown to be slower, have shorter stride lengths, display asymmetrical patterns and greater energy expenditure (Han, Chung, & Shin, 2003; Highsmith et al., 2010; Michaud et al., 2000; Sanderson & Martin, 1997; Torburn et al., 1995; Vanicek, Strike, McNaughton, & Polman, 2009; Winter & Sienko, 1988). TTA have been reported to spend more time in stance on their intact limb and less on their prosthetic limb and to load their intact limb more than their prosthetic limb (Nolan et al., 2003). Gailey et al. (2008) reported that the vast majority of individuals with TTA who use an AFP walk with at least one gait deviation. Among these were (a) moving the intact limb towards the midline with increased external rotation of the lower limb (increase foot progression angle), (b) hip hiking and (c) shorter step length. Gailey et al. also reported that these findings may be the result of improper prosthetic fit or alignment, lack of proper gait training or development of poor habits and compensation for other physical limitations.

Winter and Sienko (1988) found that in amputees there was hyperactivity of the hip extensors in early and mid-stance and above normal energy generation of the contracting musculature, leading them to believe that the energy generation at the hip extensor group after heel-strike appears to be one of the major compensations for the lack of energy generated by the ankle plantarflexors. This is in agreement with Silverman et al. (2008) who found that the primary compensatory mechanism of TTA was greater positive leg hip joint power and work in early stance, particularly the gluteus maximus and biarticular hamstring muscles. Sadeghi et al. (2000) found that the hip flexors on the amputated side compensated for the lack of normal ankle function during push-off. Rose and Gamble (1994) studied gait patterns in TTA and found that the flexion and extension of the knee during the gait cycle differed in TTA compared to that of able-bodied individuals. Greater pelvic rotation angles have been observed (Su et al., 2008) and ground reactions forces (GRF) have been seen to be as much as 23% greater in the intact limb (Gailey et al., 2008).

Traumatic amputees are typically younger individuals who undergo amputation as a result of injury (Su et al., 2008). In studies of traumatic amputees, changes in pelvic obliquity has observed as a common gait change, with the greatest difference occurring during the middle of intact limb stance because the swing-side hip is raised above its stance-side counterpart in a hip-hike, due to the inability to dorsiflex the swing foot (Michaud et al., 2000; Su et al., 2008).

Vascular amputees are typically older and more sedentary than traumatic amputees (Su et al., 2008). These individuals typically undergo amputation due to complications from diabetes at an older age than traumatic amputees. In this group a steppage compensation or increased knee and hip flexion during stance is utilized to overcome the inability to dorsiflex the AFP during the swing phase. In one study this group displayed a 20% wider base of support during walking than able-bodied individuals, displayed a greater metabolic cost of walking and slower self-selected walking speeds (Su et al., 2008).

It is important to recognize alterations in gait patterns for individuals with transtibial amputations because musculoskeletal imbalances and/or pathologies often develop into secondary physical conditions or complications (Gailey et al., 2008). As stated previously, the gait pattern is meant to function symmetrically to produce an efficient means of ambulation (Hirowaka, 1989; McGill, 2007; Whittle, 2007). When a symmetrical gait pattern is not achieved a link has been shown to an increased prevalence of degenerative changes in the lumbar spine and knees (Sanderson & Martin, 1997). Asymmetrical loading also causes higher repetitive forces to be applied to the intact limb in the majority of TTA (Nolan et al., 2003).

Some of the most commonly observed secondary conditions observed in the TTA population are knee and hip osteoarthritis of the sound limb, as well as a risk of osteoarthritis in the involved limb due to decreased muscle mass, muscular activity and GRF (Gailey et al., 2008; Nolan et al., 2003). This is in agreement with Silverman et al. (2008) that TTA have a higher risk of developing musculoskeletal disorders in their intact leg compared to the involved limb due to increased asymmetry in the loading and stance time of their intact leg. Some studies, however, have shown that TTA who utilize a prosthesis which is property fitted and are trained correctly are not at risk for osteoarthritis (Gailey et al., 2008).

The most common musculoskeletal secondary condition in TTA is low back pain (LBP) (Edhe et al., 2001; E. Smith, Comiskey, & Ryall, 2007). The prevalence of LBP in the general population has been documented at 12-45% (Edhe et al., 2001). Among individuals with transtibial amputations, LBP has been documented at 52-95% (Edhe et al., 2001; E. Smith et al., 2007). Some studies believe this may be the result of decreased hip extension, anterior pelvic rotation, lateral pelvic tilt, leg length discrepancy, back extensor strength and endurance and iliopsoas flexibility as compared to able bodied individuals (Gailey et al., 2008; Kulkarni, Gaine, Buckley, Rankine, & Adams, 2005; McGreggor & Hukins, 2009; E. Smith et al., 2007). Leg length discrepancy during swing due to the inability to dorsiflex the ankle, has been described in TTA and may be related to LBP due to the gait alterations and asymmetry. McGill (2007), however documents that leg length discrepancy in able-bodied individuals only shows a correlation with LBP at a discrepancy of greater than 5cm.

According to Kulkarni, Gaine and Buckley (2005), TTA with and without LBP had no significant differences in lower extremity joint ranges of motion, leg length or BMI between pain and pain-free amputees. There was also no significant difference in disc pathology between the two groups. This study did find, however, that there were significant differences in postural musculature with a final conclusion of postural muscles functioning asymmetrically during gait, increasing susceptibility to LBP in transtibial amputees. Similarly Su et al.,(2008) found that the hip hike motion commonly observed in traumatic TTA, decreases gait efficiency and may lead to hip and back pathology.

In an intervention study 9 traumatic transfemoral amputee participants with LBP underwent a 10 month intervention to improve gait symmetry and work through psychological issues associated with life as an amputee. After the intervention and at follow-up none of the participants reported a recurrence in LBP (Sjödahl, Jarnlo, & Persson, 2001). Similarly, Smith, Comiskey and Ryall (2008), found that in 52.9% of amputees with LBP, postural and gait abnormalities appeared to be the cause of pain.

LBP in both TTA and able-bodied individuals has been shown to decrease levels of physical activity (PA) and lead to secondary (or in the case of amputees, tertiary) conditions associated with inactivity and sedentary lifestyles (Simonsick et al., 1993). Among persons with chronic pain, it has been noted that it is often not the underlying condition, such as a limb amputation, that primarily impairs the individual, but the chronic pain itself (Ephraim et al., 2005). On the SF-36 general health survey, individuals with amputations had lower average scores than the general population (Legro et al., 1999). In the case of TTA, maintaining PA levels can help prevent conditions associated with inactivity, however, the high incidence of LBP may be a significant limiting factor for this population for participating in regular PA.

Residual limb pain and stump-socket interfaces have been studied to examine their effects on TTA gait patterns with no significant correlations shown (Versluys et al., 2009). Prosthetic limb inertial properties and weight have also been examined to study their effects on TTA gait with no significant differences noted (Mattes et al., 2000). Alterations in ankle angle in TTA has shown a significant different in metabolic cost for walking and alterations in gait patterns (Schmalz et al., 2002). One gait analysis study in able-bodied individuals with restrictions in ankle ROM also showed significant alterations in gait patterns including decreased step length and gait asymmetry in phase lengths (Crosbie, Green & Refshauge, 1999). Participants in this study, however, were recovering from lateral ankle sprains and still experiencing pain as a restriction in their gait pattern. Further research concerning ankle range of motion and gait patterns in able-bodied individuals may allow for a better understanding of prosthetic ankle function in TTA. Understanding the gait patterns that occur in TTA may improve prosthetic ankle prescription and assist clinicians in preventing secondary conditions.

Ankle Range of Motion and Gait Patterns

As stated previously, TTA who utilize a prosthesis for walking display alterations in gait patterns and symmetry, and an increased metabolic cost of walking which may lead to LBP and other secondary conditions. Previous research has hypothesized that lack of ankle range of motion is a major contributor to alterations in gait patterns (Au et al., 2007; Crosbie, Green & Refshauge, 1999; Hansen, Childress, Miff, Gard, & Mesplay, 2004). These hypotheses, however, do not adequately explain the differences in patterns observed in vascular and traumatic amputees.

Further research is needed to examine alterations in gait patterns and gait symmetry that occur when there is a loss of ankle ROM in individuals without TTA. Ankle ROM has been shown to be a major contributor to creating alterations in gait patterns, however other subsystems may play a greater role in individuals with TTA than ankle ROM to create more optimal gait patterns.

Chapter 3: METHODS

Approval for the study protocol (Appendix A) was obtained from the Oregon State University Institutional Review Board.

Participants

19 participants (9 men, 10 women) completed the study. Participant demographic information is presented in Table 3.1. Participants had no previous history of serious lower extremity injury, malformation or current lower extremity pain (screening- Appendix B). Participants were recruited from Oregon State University and the surrounding community using fliers and e-mail request for study volunteers. Informed consent was obtained from all participants prior to participation in the study.

Table 3.1: Participant Demographic Information

	Mean ± SD
Age (years)	23.7 ± 4.5
Height (m)	1.75 ± 0.08
Mass (Kg)	75.8 ± 14.04
Right Dominant	15
Left Dominant	4

Experimental Design

All participants completed 10 walking trials in each of two conditions during one testing period. Condition one was the control condition: walking at a self-selected comfortable walking speed wearing athletic shoes. The other condition was the experimental condition: walking at a self-selected comfortable walking speed wearing athletic shoes with the dominant ankle restricted at approximately 0 degrees plantarflexion using plaster casting material. The dominant limb was determined using step up, ball kick, and balance recovery tests (Appendix B). The order of completion of both conditions was randomized for each participant, with practice trials performed before each condition.

The experimental condition restricted ankle ROM to approximately 0 degrees of plantarflexion by application of a plaster splint using plaster casting (Johnson & Johnson, New Brunswick, NJ), synthetic cast padding (Johnson & Johnson, New Brunswick, NJ) and under-cast stockinette (3M, St. Paul, MN). The under-cast stockinette was applied directly to the skin followed by a layer of cast padding. The ankle was positioned at approximately 0 degrees plantarflexion using a goniometer. The roll of plaster casting was then applied from 1cm below the first metatarsal head to the flare of the gastrocnemius muscle, with researchers assisting in holding the ankle in the correct position. After the splint had dried for approximately 3-5min the participant's shoe, with insole removed, was placed over the splint for the experimental condition. Foam inserts were placed into the non-dominant shoe to aid with the shoe/cast thickness discrepancy. A five minute accommodation period was given before collecting data in the experimental condition. During this time, participants were instructed to walk as normally as possible.

Figure 3.1: Plaster Splint Application

Gait Analysis Instrument

A nine-camera motion capture system (Vicon, Los Angeles, CA) was utilized to record the position of passive-reflective markers during each trial. Reflective markers were affixed to the skin, tight-fitting clothing or footwear with at least two markers per body segment (Table 3.2). Segments included the trunk, pelvis, thigh, leg and foot. Before each condition, a static trial of quiet standing in a known position was recorded.   During each trial, participants were instructed to walk as is they were walking to class or the store across a 4 meter pathway. Ten trials were recorded in each condition.

Table 3.2: Motion Capture Marker Placements

Segment	**Marker Placement**
Trunk	Sternum, C7, T10
Pelvis	Anterior superior iliac spine, posterior superior iliac spine
Thigh	Mid thigh
Leg	Lateral epicondyle, mid-leg
Foot	Lateral malleolus, heel, head of fifth metatarsal

Trials were recorded at 60Hz. Data were filtered using a no-lag Butterworth filter, low-pass fourth-order, with a cutoff frequency of 10Hz, as determined by residual analysis. For each trial of interest, BodyBuilder software (Vicon, Los Angeles, CA) was used to reconstruct the 3-dimensional paths of the reflective markers from what was observed by the nine cameras. Positions of the heels, toes and joint centers and body segment orientations were computed in 3-D using transformations derived from the trial of quiet standing and from measured body dimensions. A custom MATLAB program (Mathworks, Natick, MA) was used to extract the variables of interest.

Data Reduction

Variables of interest included spatiotemporal parameters of gait and joint angles. The spatiotemporal parameters were a) velocity, b) cadence, c) step length, d) step width, e) step and swing phase durations. The joint angles were a) trunk lateral bend, b) pelvic rotation, c) pelvic obliquity, d) hip flexion, e) knee flexion, and f) foot progression angle.

Step length was defined as the distance from ipsilateral to contralateral heel strike (Skinner & Effeney, 1985). Cadence was measured in steps per second. Stance and swing phase durations were evaluated as total time (seconds) in each phase with stance phase consisting of the time from heel strike to toe off and swing phase consisting of the time from toe off to heel strike.

Trunk angle was computed according to a Cardan rotation sequence of forward flexion, lateral flexion, rotation. Pelvis angle was computed according to a Cardan sequence of anterior/posterior rotation, superior/inferior rotation, transverse rotation. Pelvic obliquity was defined as the amount of lateral tilt of the pelvis occurring from the transverse plane (superior/inferior rotation) (Michaud et al., 2000). A positive pelvic obliquity angle indicates an upward (superior) rotation of the pelvis on the swing-limb side. Hip and knee angles were computed according to a Cardan rotation sequence of flexion, abduction, external rotation or the distal segment. Foot angles were computed according to a Cardan rotation sequence of plantarflexion, abduction, eversion. Foot progression angle was measured as abduction during swing.

The following variables were considered limb-dependent gait measures: a) step length, b) step width, c) phase lengths, d) joint angles. Limb-dependent gait measures were evaluated for symmetry using a degree of asymmetry (DoA) value. DoA was evaluated in each condition using the absolute value of the difference between limbs: DoA= |(Dominant)-(Non-dominant)|.

A DoA value of zero indicated perfect symmetry of the limbs for the specified condition. Larger values indicated greater asymmetry of the limbs for the specified condition.

Values were averaged across steps and trials using Microsoft Excel 7 (Microsoft, Redmond, WA).

Statistical Analysis

The following statistical analyses were used to examine variables of interest: Specific research question 1: How does restricted ankle range of motion affect spatiotemporal parameters of gait?

To answer the first research questions two separate analyses were conducted.

First, the effects of restricted ankle range of motion on spatiotemporal parameters of gait that were not limb-dependent were examined by 1-way repeated measures multivariate analysis of variance (MANOVA). Dependent variables were velocity and cadence, the independent variable was condition.

Second, a 2 x 2 (limb by condition) repeated measures MANOVA was used to examine the effects of restricted ankle range of motion on limb-dependent spatiotemporal parameters of gait. Dependent variables were a) step length, b) step width, c) swing phase length and d) stance phase length. Independent variables were condition and limb.

Specific research question 2: How does restricted ankle range of motion affect gait symmetry?

Gait symmetry was evaluated changes in the symmetry of limb-dependent gait measures in each condition using calculated degree of asymmetry (DoA) values. In order to examine gait symmetry as an overall concept, limb-dependent gait measures included both spatiotemporal parameters and joint angles.

To answer the second research question, DoA values were examined between conditions using 1-way repeated measures MANOVA. Dependent variables were DoA values for: a) step length, b) step width, c) stance phase length, d) swing phase length) e) peak trunk lateral bend in swing, f) peak pelvic obliquity in swing, g) range of pelvic rotation in swing, h) peak hip abduction in swing, i) peak hip flexion in swing, j) hip flexion at toe-off, k) peak knee flexion in swing, l) knee flexion at toe-off, m) peak knee flexion during weight acceptance, n) peak foot progression angle in swing. Independent variable was condition.

Specific research questions 3: what compensatory patterns does restricted ankle ROM create during gait?

Compensatory gait patterns were evaluated using limb-dependent joint angles. To answer the third research question, joint angles were examined using 2 x 2 (limb by condition) repeated measures MANOVA. Dependent variables were a) peak trunk lateral bend in swing, b) peak pelvic obliquity in swing, c) range of pelvic rotation in swing, d) peak hip abduction in swing, e) peak hip flexion in swing, f) hip flexion at toe-off, g) peak knee flexion in swing, h) knee flexion at toe-off, i) peak knee flexion during weight acceptance, j) peak foot progression angle in swing. Independent variables were condition and limb.

Data were analyzed using SPSS 13.0 software. Alpha level was set at 0.05. Interaction effects of 2-way analyses were examined to determine changes in dependent variables that were dependent upon limb.

Chapter 4: RESULTS

Spatiotemporal Parameters of Gait

Gait analysis for velocity and cadence indicated that the mean velocity and cadence in the experimental condition decreased compared to the control condition (Table 4.1).

Table 4.1: Velocity and Cadence Descriptive Statistics

	Control		Experimental	
	Mean	SD	Mean	SD
Velocity (m/s)**	1.36	0.24	1.24	0.27
Cadence (steps/s)**	1.82	0.14	1.77	0.17

** significant at .001 level.

Results of the one-way repeated measures MANOVA indicated a significant overall difference in velocity and cadence (λ=0.004, $p<0.01$, η^2=0.503). The follow up, follow-up analysis indicated a significant difference in both velocity and cadence between the control and experimental conditions, $F(1,19) = 17.02$, $p<0.01$, $\eta^2 = 0.49$ and $F(1,19) = 12.25$, $p<0.01$, $\eta^2 = 0.41$, respectively.

Limb-dependent spatiotemporal values were evaluated using a 2 x 2 (limb by condition) repeated measures MANOVA. Results revealed a significant overall interaction for limb-dependent spatiotemporal parameters (λ=0.046, $p<0.01$, η^2=0.95). Follow-up results indicated a significant difference in swing and stance phase lengths $F(1,19)$ =232.66, $p<0.01$, $\eta^2 = 0.93$ and $F(1,19) = 135.80$, $p<0.01$, $\eta^2 = 0.88$, respectively. Means and standard deviations for limb-dependent spatiotemporal parameters are presented in Table 4.2. See Figures 4.1 and 4.2 for swing and stance phase length interactions.

Table 4.2: Spatiotemporal Parameters Descriptive Statistics

	Control		Experimental	
	Mean	SD	Mean	SD
Step Length-Dominant (cm)	74.05	8.28	70.62	8.81
Step Length- Non-Dominant (cm)	74.41	9.09	69.21	11.90
Step Width- Dominant (cm)	10.02	2.93	10.91	3.64
Step Width- Non-Dominant (cm)	8.44	3.51	9.31	3.57
Stance Phase Length- Dominant (sec)**	0.70	0.07	0.71	0.09
Stance Phase Length- Non-Dominant (sec)**	0.69	0.08	0.74	0.09
Swing Phase Length- Dominant (sec)**	0.41	0.02	0.44	0.03
Swing Phase Length- Non-Dominant (sec)**	0.41	0.02	0.40	0.03

**significant at 0.01 level

Figure 4.1: Swing Phase Length Interaction

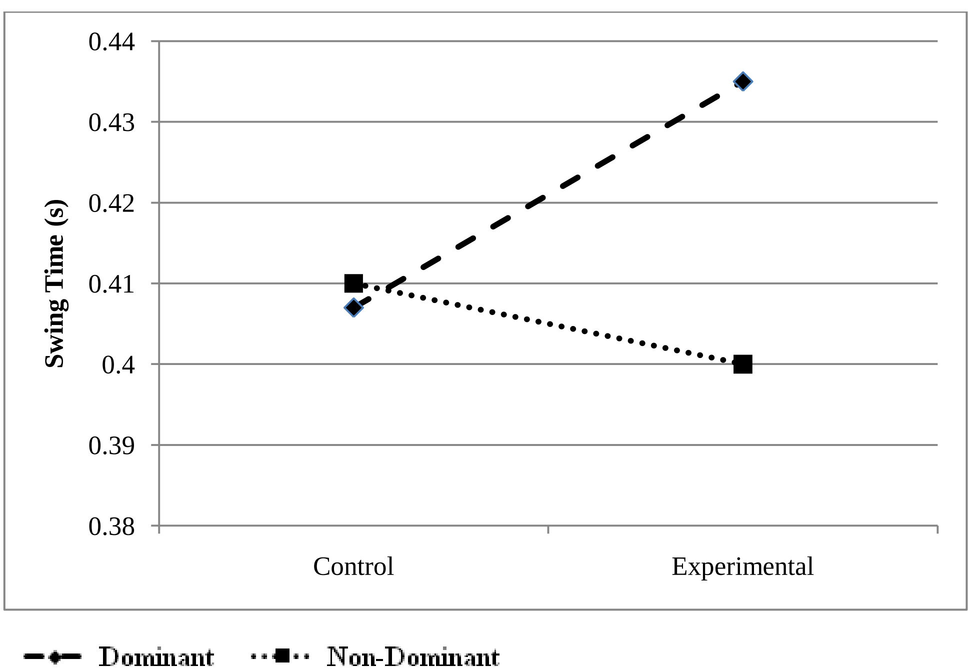
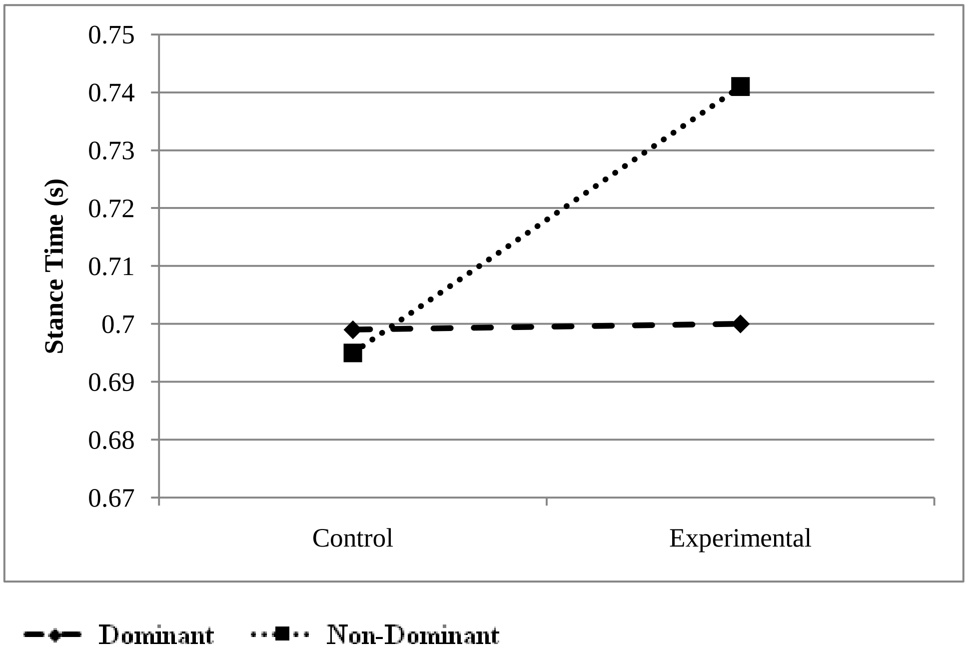


Figure 4.2: Stance Phase Length Interaction



Gait Symmetry

Mean DoA values for limb-dependent measures in both conditions are presented in Table 4.3. Results of the one-way repeated measures MANOVA indicated an overall significant difference in gait symmetry (λ=0.046, $p<0.01$, η^2=0.95) as measured by DoA.

Follow up univariate analysis showed a significant difference in symmetry between conditions for stance phase length $F(1,19) = 40.87$, $p<0.01$, $\eta^2 = 0.69$, swing phase length $F(1,19) = 77.38$, $p<0.01$, $\eta^2 = 0.81$, hip flexion at toe-off $F(1,19) = 12.78$, $p<0.01$, $\eta^2 = 0.42$ and knee flexion at toe-off $F(1,19) = 20.87$, $p<0.01$, $\eta^2 = 0.54$.

Table 4.3: Limb Dependent Gait Measures DoA Descriptive Statistics

	Control		Experimental	
	Mean	SD	Mean	SD
Step Length (cm)	1.82	1.40	3.14	3.76
Step Width (cm)	2.65	1.45	2.87	1.74
Swing Phase Length (sec)**	0.01	0.00	0.03	0.01
Stance Phase Length (sec)**	0.01	0.01	0.04	0.01
Trunk Lateral Bend (deg)	7.8	1.5	2.37	1.46
Pelvic Obliquity (deg)	1.47	1.51	2.12	1.57
Pelvic Rotation (deg)	1.01	0.63	1.41	1.02
Hip Abduction (deg)	2.46	1.68	2.87	2.03
Peak Hip Flexion- Swing (deg)	1.73	1.04	1.34	0.65
Hip Flexion at toe-off (deg)**	1.38	1.82	2.82	1.9
Peak Knee Flexion- Swing (deg)	2.27	1.07	4.07	3.7
Knee Flexion at toe-off (deg)**	2.25	1.38	5.13	3.17
Knee Flexion Weight Acceptance (deg)	2.52	1.95	3.15	2.15
Foot Progression Angle (deg)	5.29	5.48	5.43	4.14

**significant at 0.01 level

Compensatory Patterns

Mean and standard deviation values for limb-dependent gait measures in each condition are presented in Table 4.4. A 2 x 2 (limb by condition) repeated measures MANOVA was used to examine limb-dependent gait measures between conditions.

Results indicated a significant limb by condition interaction (λ=0.046, $p<0.01$, η^2=0.95). The follow up, univariate analysis reveal a significant limb by condition interaction for pelvic obliquity $F(1,19) = 12.38$, $p<0.01$, $\eta^2 = 0.41$, hip flexion at toe-off $F(1,19) = 8.57$, $p<0.01$, $\eta^2 = 0.32$, knee flexion at toe-off $F(1,19) = 52.33$, $p<0.01$, $\eta^2 = 0.74$ and foot progression angle $F(1,19) = 7.29$, $p<0.01$, $\eta^2 = 0.29$. See Figures 4.3 through 4.7 for limb by condition interactions.

Table 4.4: Limb-Dependent Gait Measures Descriptive Statistics

	Control		Experimental	
Limb-Dependent Gait Measures (deg)	Mean	SD	Mean	SD
Trunk Lateral Bend (swing)- Dominant	1.48	1.47	1.60	1.61
Trunk Lateral Bend (swing)- Non-Dominant	2.40	1.44	2.80	1.59
Pelvic Obliquity (swing)- Dominant**	1.69	1.93	1.15	1.02
Pelvic Obliquity (swing)- Non-Dominant**	1.14	1.02	1.87	1.34
Pelvic Rotation (range, swing)-Dominant	9.89	2.79	9.36	3.03
Pelvic Rotation (range, swing)- Non-Dominant	10.33	3.42	10.21	3.68
Hip Abduction (peak, swing)- Dominant	6.73	2.37	6.18	2.69
Hip Abduction (peak, swing)- Non-Dominant	7.63	2.82	6.52	3.26
Hip Flexion (peak, swing)-Dominant	35.91	8.12	36.12	8.00
Hip Flexion (peak, swing)- Non-Dominant	35.86	7.22	36.27	7.80
Hip Flexion (toe-off)-Dominant**	0.30	8.05	-0.01	8.85
Hip Flexion (toe-off)- Non-Dominant**	0.69	7.28	2.02	8.06
Knee Flexion (peak, swing)-Dominant	62.96	4.22	61.05	6.28
Knee Flexion (peak, swing)- Non-Dominant	61.05	6.28	62.93	4.86
Knee Flexion (toe-off)-Dominant**	41.29	5.15	36.62	5.04
Knee Flexion (toe-off)- Non-Dominant**	41.07	5.78	41.54	6.18
Knee Flexion (peak, weight acceptance)- Dominant	14.31	5.40	15.46	5.06
Knee Flexion (peak, weight acceptance)- Non-Dominant	14.61	5.89	14.79	5.50
Foot Progression (peak, swing)- Dominant**	13.61	6.75	12.99	7.47
Foot Progression (peak, swing)- Non-Dominant**	9.66	4.36	11.15	5.23

**significant at 0.01 level

Figure 4.3 Pelvic Obliquity Interaction

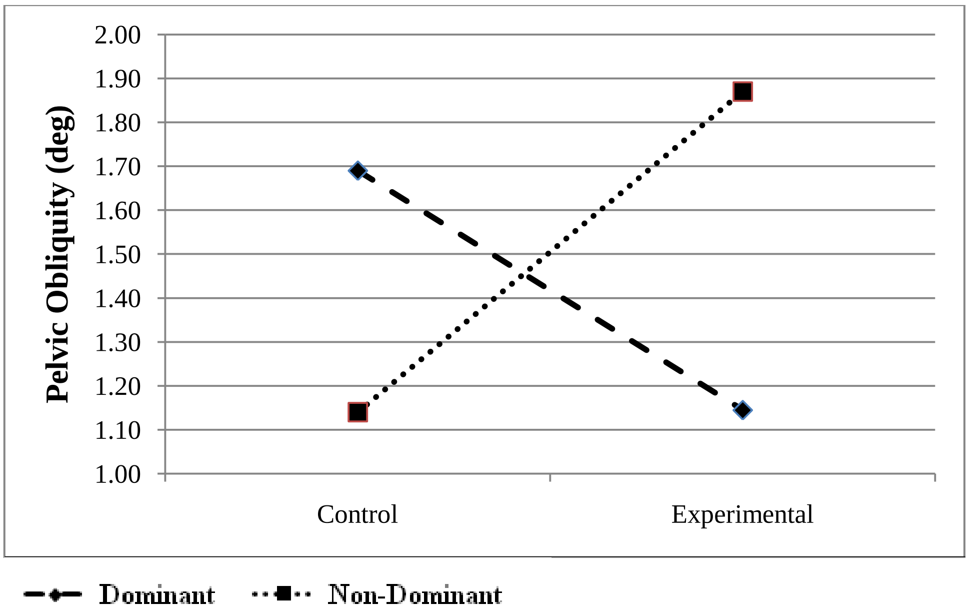


Figure 4.4: Hip Flexion at Toe-off Interaction

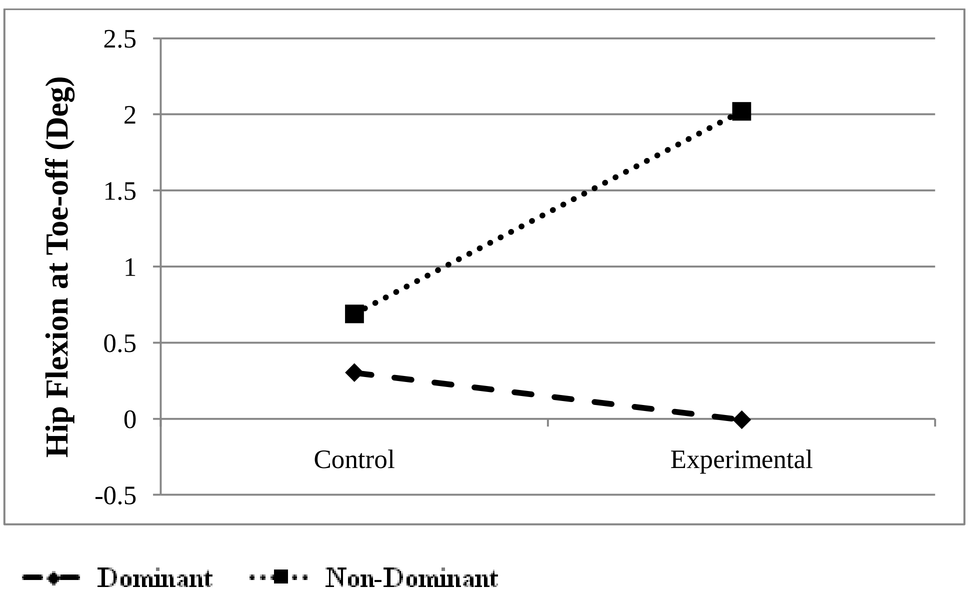


Figure 4.5: Knee Flexion at Toe-off Interaction

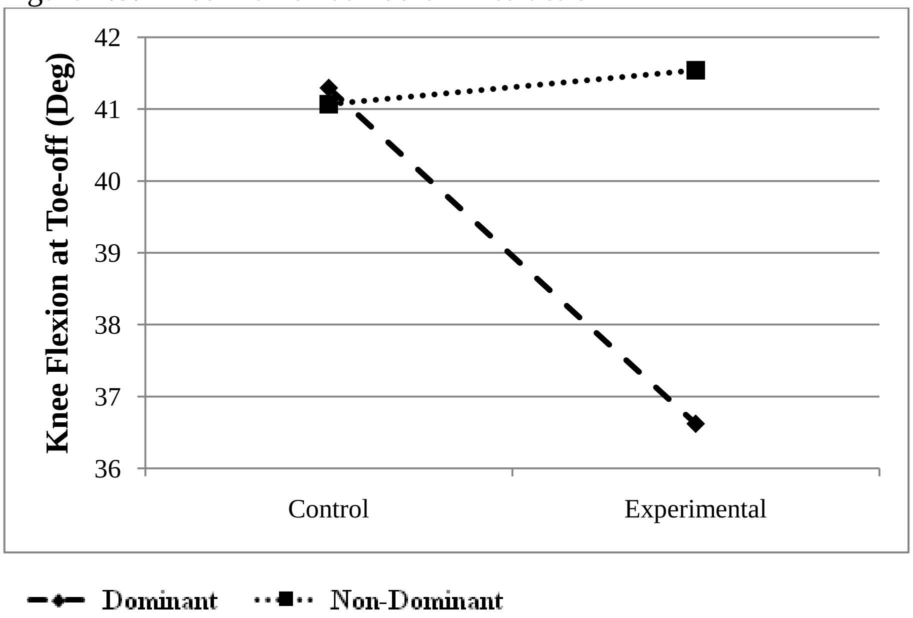
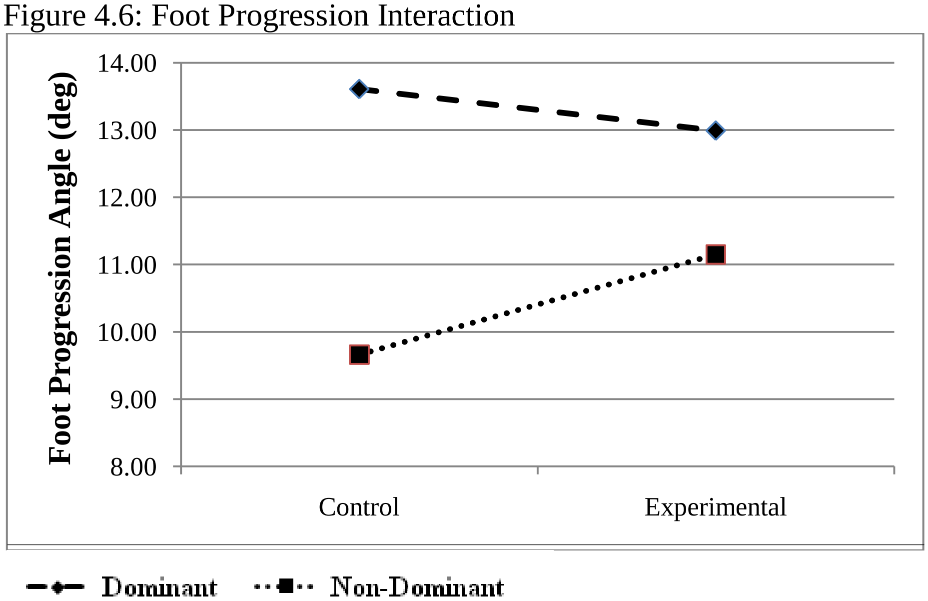


Figure 4.6: Foot Progression Interaction



Chapter 5: DISCUSSION

The purpose of this study was to examine the effects of restricted ankle ROM on three components of gait: spatiotemporal parameters, gait symmetry and compensatory patterns.

The first research question sought to examine how restricted ankle ROM affects spatiotemporal parameters of gait. Spatiotemporal parameters of gait can give an indication of overall gait function and efficiency (Highsmith et al., 2010). For this question, two significant findings were discovered. First, restricted ankle ROM contributes to decreased velocity and cadence. Second, restricted ankle ROM contributes to alterations in swing and stance phase lengths. As a whole, temporal parameters (velocity, cadence and phase lengths) were affected by restricted ankle ROM, but spatial parameters (step length and step width) were not significantly affected.

A decrease in velocity and cadence was observed during the experimental condition. This may be due to a decrease in gait efficiency. Previous studies have found that TTA have a self-selected walking speed that is similar in metabolic cost to individuals without amputation (Genin, Bastien, Franck, Detrembleur, & Willems, 2008). Walking with restricted ankle ROM may also decrease gait efficiency and influence participants to walk slightly slower in order to maintain a similar metabolic cost for walking.

A significant difference in step and swing phase lengths was observed during the experimental condition. When ankle ROM was restricted, stance time on the unrestricted limb increased and swing time decreased. Stance time on the restricted limb was similar and swing time increased (see figures 4.1 and 4.2). Previous researchers have hypothesized that TTA spend more time in stance on the intact limb due to pain or a decrease in balance (Nolan et al., 2003). Our study shows that lack of ankle ROM also significantly contributes to these differences in stance and swing phase lengths.

While restricted ankle ROM affected temporal parameters of gait similarly to TTA gait, spatial parameters (step length and step width) were not significantly affected. Participants in our study showed no significant difference in step length or step width. Previous studies in TTA gait patterns and in individuals with restricted ankle ROM due to injury or joint fusion, have shown a decreased contralateral step length (Crosbie, Green & Refshauge, 1999; Goodman et al., 2004; Isakov et al., 2000). Greater step width has also been associated with TTA gait in order to increase the base of support for improved balance (Vanicek et al., 2009).

Our findings indicate that restricted ankle ROM does not contribute to increased step width. Increased step width in TTA may more likely be due to an increase in base of support to improve balance than due to lack of ankle ROM. Our findings also indicate that restricted ankle ROM does not contribute to decreased contralateral step length. This may be due to greater ROM at the metatarsophalangeal (MTP) joints, or ability to maintain proprioceptive sense in terminal stance. If an individual with restricted ankle ROM was able to fully extend the toes while maintaining balance in terminal stance, they may be able increase step length by compensating for loss of ankle ROM with increased toe extension. These results do not support the hypothesis that ankle ROM independently contributes significantly to changes in spatial parameters of gait in TTA.

The second research question sought to examine how restricted ankle range of motion affects gait symmetry. Gait symmetry has been shown to be a relevant measure for investigating gait characteristics and implications for future joint pain and degradation (Isakov et al., 2000).

Our results revealed that restricted ankle ROM contributes to an overall decrease in gait symmetry. Limb-dependent gait measures, including spatiotemporal parameters, were significantly less symmetrical in the experimental condition. Results also showed symmetry of the hip and knee joints at toe-off was significantly decreased. These findings indicate that restricted ankle ROM contributes to an overall decrease in gait symmetry.

During walking, the ankle reaches an end-point of its ROM during terminal stance and toe-off. It would logically follow that ankle ROM is a factor that influences gait symmetry at toe-off. When ankle ROM is restricted, changes in gait may occur in the ipsilateral or contralateral limb. If changes do not occur equally in both limbs, the gait pattern becomes less symmetrical. We observed a significant decrease in gait symmetry specifically in two limb-dependent gait measures: hip flexion at toe-off and knee-flexion at toe-off. These results indicate that inability to plantarflex the ankle at toe-off contributes to a decrease in gait symmetry, specifically at the hip and knee.

The third research question sought to examine compensatory gait patterns resulting from restricted ankle ROM. Compensatory gait patterns were evaluated by measurement of limb-dependent gait measures. Changes in limb-dependent gait measures may reveal some of the underlying factors that influence the observed changes in temporal parameters and gait symmetry. Statistical results of our study revealed significant changes in the gait pattern at the a) pelvis, b) hip, c) knee and d) foot.

In addition to statistical analysis, ensemble averages were created for each participant. Participant ensemble average patterns were examined individually to determine if a difference in patterns was apparent between participants that may have been lost within our statistical analysis (see Figures 5.1-5.4). A difference in pelvic obliquity pattern on the non-dominant limb was observed in both conditions for one participant. No other major differences were noted.

Figure 5.1: Pelvic Obliquity Individual Ensemble Averages

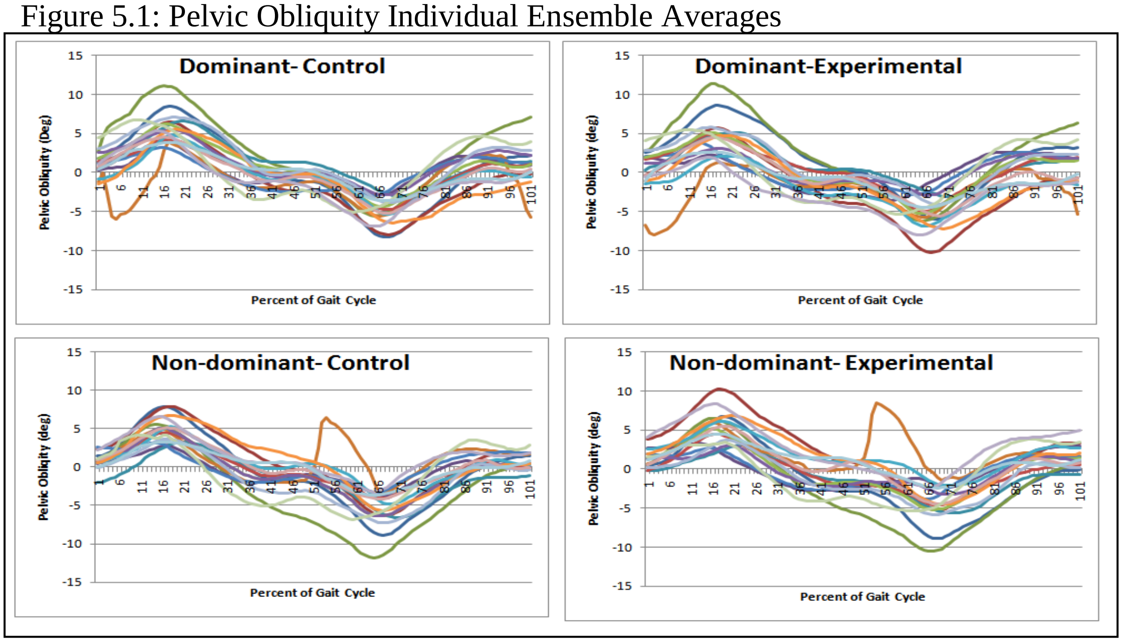


Figure 5.2: Hip Flexion Individual Ensemble Average

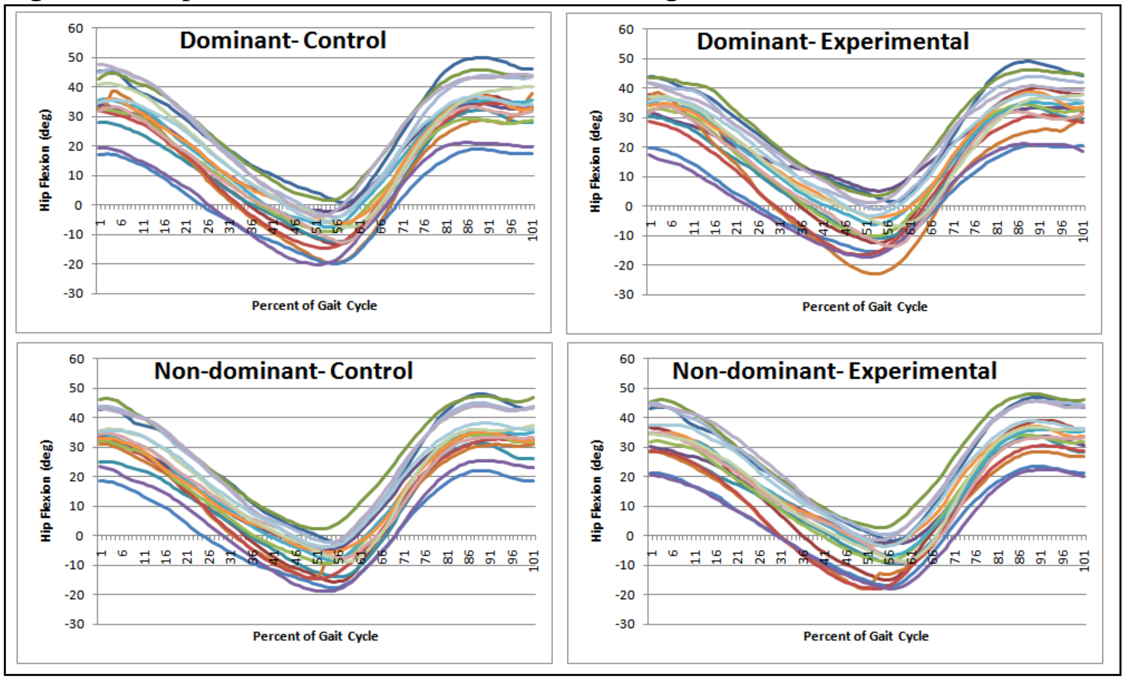


Figure 5.3: Knee Flexion Individual Ensemble Averages

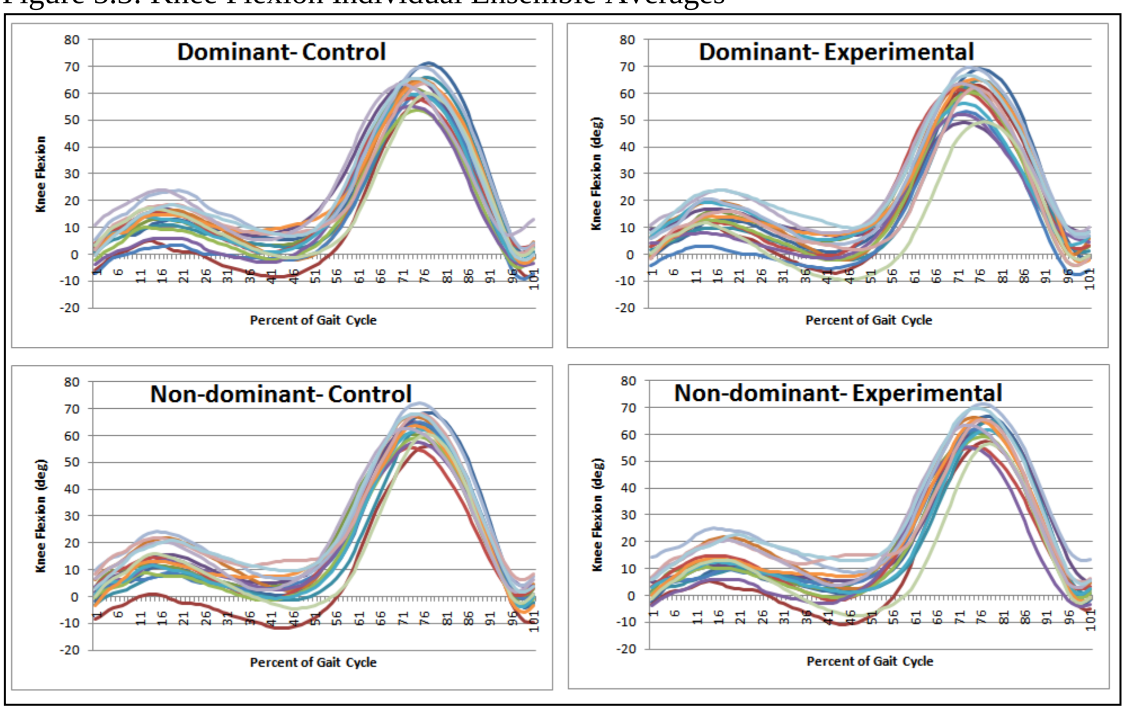
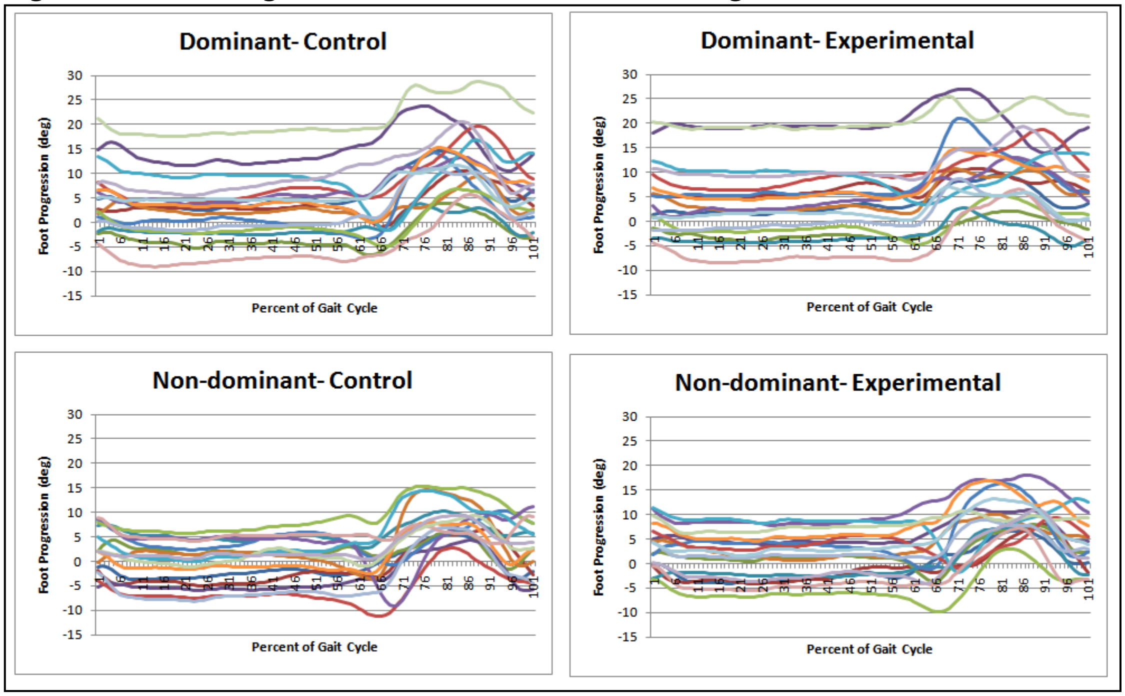


Figure 5.4: Foot Progression Individual Ensemble Averages



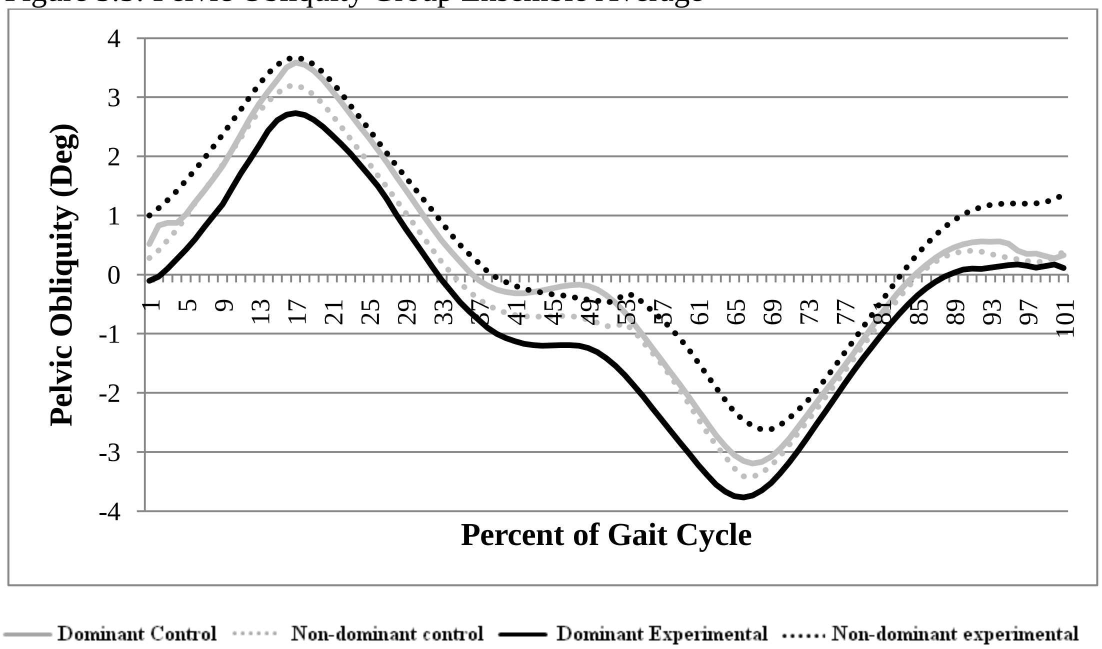
Ensemble averages were then averaged across all participants to determine if group compensatory patterns existed (Figures 5.5-5.8).

Statistical results showed a significant difference in pelvic obliquity during the gait cycle. In normal gait, the swing-side pelvis drops below the stance-side pelvis during swing (Kuo, 2007). Previous gait studies in TTA have revealed a reversal of the pelvic obliquity pattern for the amputated side, meaning that the swing-side pelvis is raised above the stance-side pelvis during swing, also known as "hip-hiking" (Michaud et al., 2000). Results of our study indicated a significant difference in pelvic obliquity in a different manner than has been observed in TTA. Our results indicated that the pelvis on the restricted side dropped farther below the stance-side pelvis during swing in the experimental condition than in the control condition. Conversely, the pelvis on the non-restricted side did not drop as far below the stance-side pelvis during swing in the experimental condition than the control condition (see Figure 4.3). Since pelvic obliquity increased for one limb and decreased for the other, pelvic obliquity symmetry did not change significantly.

A change in pelvic obliquity during swing should be linked to observed changes in other limb-dependent gait measures during swing. However, we did not observe any significant changes in limb-dependent measures during swing which would follow the observed change in pelvic obliquity pattern. This may be due to the overall difference in pelvic obliquity measures between limbs and conditions being less than 1° .

The ensemble average graph for pelvic obliquity (Figure 5.5) reveals the results of the statistical analysis (decrease in pelvic obliquity on restricted side, increase in pelvic obliquity on non-restricted side). Swing phase begins at approximately 60% of the gait cycle. Interestingly, the ensemble average also reveals that across the full gait cycle, pelvic obliquity is decreased on the restricted side (black line is below grey line) and increased on the non-restricted side (grey dots are below black dots). This finding shows that across the gait cycle, the pelvis remained raised on the non-dominant side compared to the dominant side in the restricted condition. This has not been observed in previous studies.

Figure 5.5: Pelvic Obliquity Group Ensemble Average



Our results showed that restricted ankle ROM also contributes to compensatory patterns at the hip and knee. The restricted limb and non-restricted limb behaved differently from each other during the experimental condition. On the restricted side, the hip and knee were more extended at toe-off in the experimental condition. On the non-restricted side the hip and knee were more flexed at toe-off in the experimental condition.

Increased hip and knee extension at toe-off on the restricted side may be explained by participants' ability to transfer weight further over the foot in terminal stance. Stance time on the dominant (restricted) limb was similar between conditions. Maintaining stance time on the restricted limb at a slower velocity during the experimental condition may have allowed for participants to obtain greater extension at the toes and extension at the hip in order to compensate for lack of ankle ROM.

This is in agreement with our finding that contralateral step length did not differ between conditions.

Increased knee and hip flexion at toe-off on the non-restricted side may be explained by loss of controlled plantarflexion of the restricted limb at heel-strike. At heel-strike of the restricted limb, the foot is unable to rotate in a controlled manner from the point of heel-strike to the point where the foot is flat on the ground (flat-foot). Instead, flat-foot occurs earlier in the stance phase. If flat-foot occurs earlier in the stance phase, the body's center of mass (COM) shifts over the stance limb more quickly, rather than gradually laying the foot flat on the ground, then shifting the COM forward over the stance limb. Since the COM is shifted forward sooner in the stance phase the foot may leave the ground before the hip and knee on the non-restricted side are able to achieve maximum extension.

Results of our study indicate that restricted ankle ROM contributes to compensatory patterns at the hip and knee. In TTA increased hip and knee extension at toe-off has been observed in conjunction with hyperactivity of the hip extensors during terminal stance on the amputated side (Bateni & Olney, 2002). Previous researchers hypothesized that increased activity of the hip extensors was a compensation for loss of power generation by the foot at toe-off (Winter & Sienko, 1988). Kinetic data were not collected for our study; however, we observed a similar kinematic pattern at the hip and knee as has been observed in TTA. This indicates that restricted ankle ROM may contribute to compensatory patterns at the hip and knee in TTA.

Ensemble averages for hip and knee flexion (Figures 5.6 and 5.7) do not greatly display the statistical differences at toe-off (approximately 60% of gait cycle). However, difference in gait symmetry is apparent. In the control condition (grey dots and grey line) the pattern is very similar, with the two lines overlying each other the majority of the time (symmetrical). In the experimental condition (black dots and black line) the patterns appear different with the two lines further apart from one another (asymmetrical). Ensemble average for knee flexion also reveals a difference in pattern during swing. On the dominant limb, during the experimental condition, knee flexion is decreased (in agreement with statistical finding) however, the peak value appears to occur earlier in the swing phase than the non-dominant limb.

Figure 5.6: Hip Flexion Group Ensemble Average

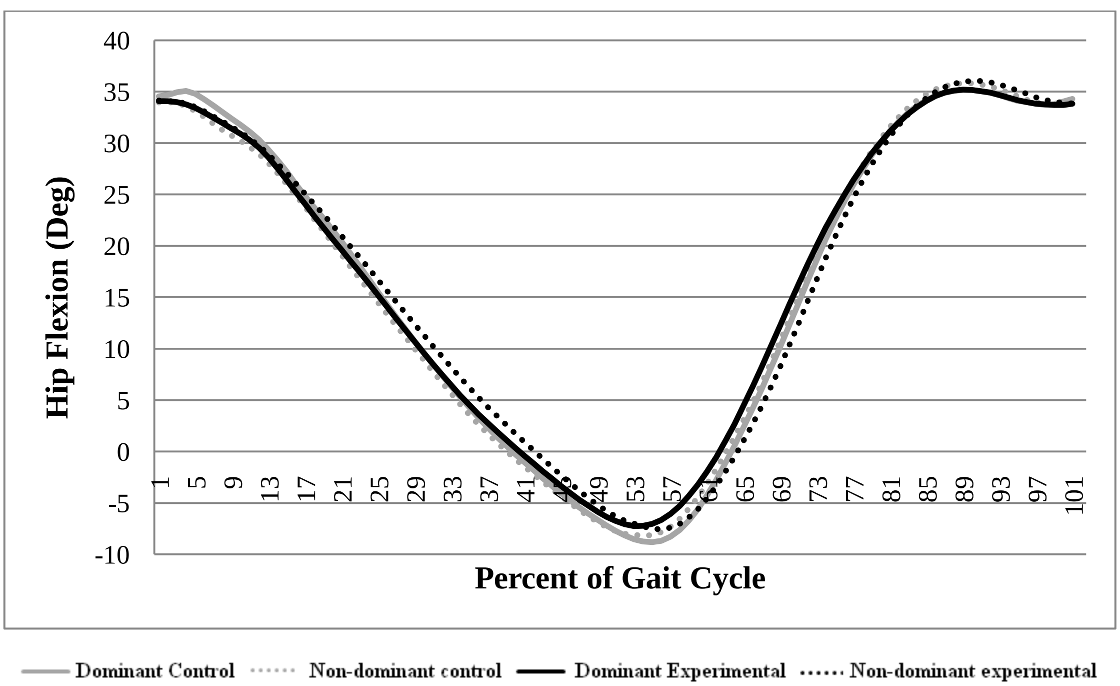
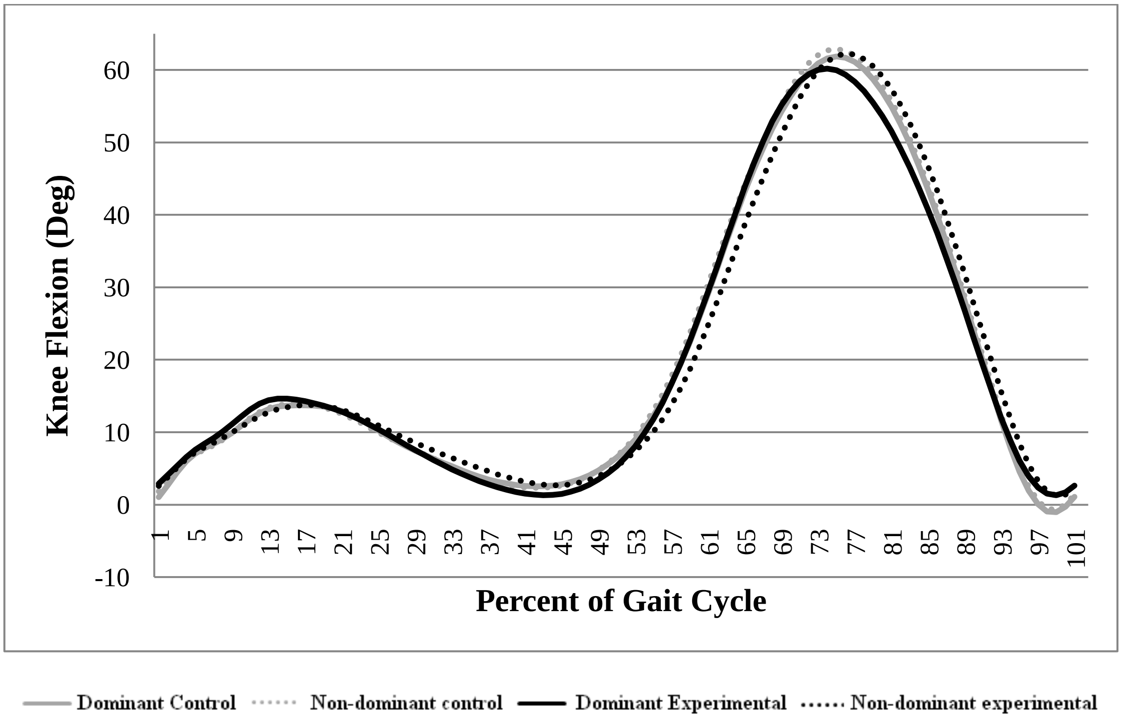
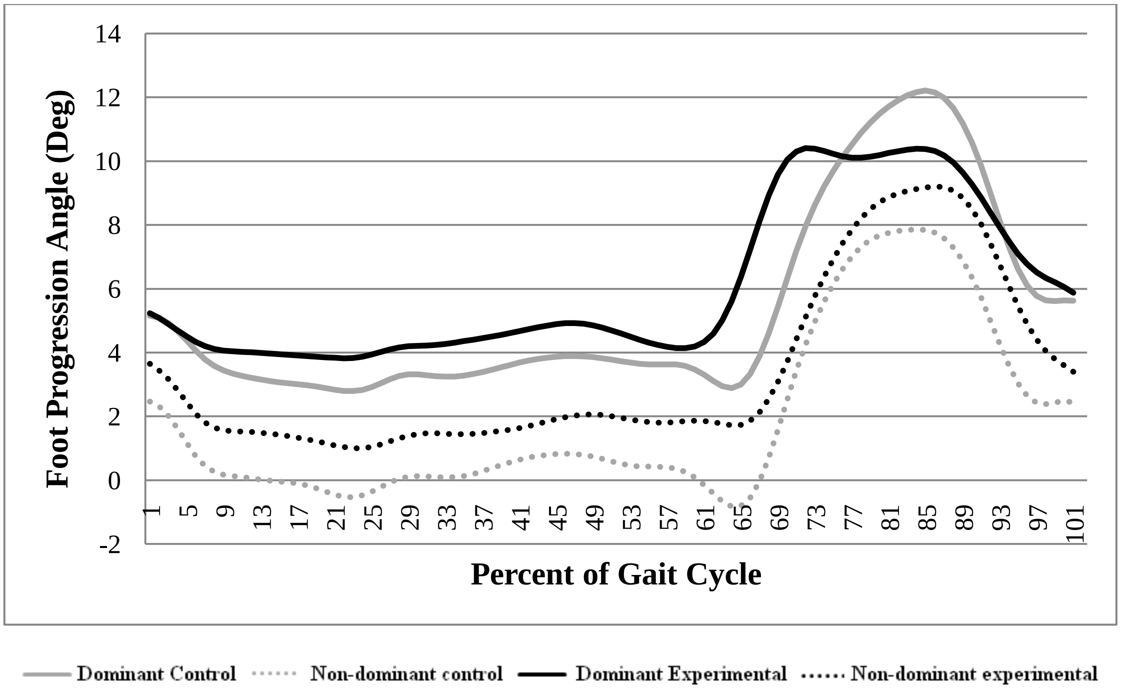


Figure 5.7: Knee Flexion Group Ensemble Average



Finally, our results showed that restricted ankle ROM contributes to a compensatory pattern at the foot. During the experimental condition, foot progression angle (external rotation of the foot and lower leg) of the restricted limb during swing decreased, and the foot progression angle of the non-restricted limb increased. These results were the opposite of patterns that have been observed in TTA. In TTA foot progression angle was observed to increase during swing on the prosthetic side in order to achieve better ground clearance of the foot (Grumillier et al., 2008). Participants in our study showed a significantly different foot progression angle pattern as a whole. This pattern is best observed by ensemble average (see Figure 5.8).

Figure 5.8: Group Foot Progression Ensemble Average



Chapter 6: CONCLUSION

The purpose of this study was to examine the effects of restricted ankle ROM on three components of gait: spatiotemporal parameters, gait symmetry and compensatory patterns. We observed alterations in temporal parameters of gait, decrease in gait symmetry and a compensatory pattern of the pelvis, hip, knee and foot.

In reference to TTA gait patterns, we observed both similarities and differences in gait patterns with restricted ankle ROM compared to those observed in TTA. Decreased velocity and cadence were similar, as well as an overall decrease in gait symmetry. Increased stance time on the un-restricted/intact side was also similar. Additionally we observed a similar decrease in knee and hip flexion at toe-off on the restricted/involved side. These findings indicate that restricted ankle ROM may contribute to some of the alterations in gait pattern that are observed in TTA.

Overall, we did not observe either of the particular gait patterns (hip-hike or steppage) which have been associated with TTA. This indicates that ankle ROM may be a smaller component to alterations in gait patterns in TTA than previous researchers have hypothesized. This may be due to a) proprioception, b) other prosthetic components and/or c) ROM and power generation at the MCP/toes during terminal stance.

Individuals with TTA lack proprioceptive sense in the prosthetic limb. This may contribute to TTA's greater step width in order to improve balance. Maintaining proprioception and balance in the restricted limb may have allowed our participants to advance further over the foot during terminal stance than TTA and therefore not demonstrate a shorter contralateral step length. Having the ability to create a force with the toes during terminal stance may have also prevented further alterations in the gait pattern in our study.

Other prosthetic components, including the pylon, socket and residual limb condition, may contribute more significantly to alterations in TTA gait patterns than ankle ROM. Likely the combination of other components along with restricted ankle ROM, loss of proprioception and inherent individual differences contribute to differences in TTA gait patterns.

Concerning range of motion at the MCP joints, since range of motion at the MCP joints was not restricted in our study, participants were able to generate a force at the MCP joint during terminal stance. This power generation may have decreased asymmetry and prevented other compensatory patterns from occurring. Further research might benefit from collecting kinetic data when ankle ROM is restricted. Comparing power generation by the plantarflexors at the MCP joints only with power generated by the plantarflexors at the ankle and MCP joints may allow researchers to better understand how a powered AFP might function. Currently researchers are attempting to create powered AFPs with power generation at the ankle joint. However, if power generated at the MCP joints, as was present in this study, decreases the incidence of compensatory gait patterns, a simplified device may be more effective.

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APPENDICES

Appendix A: IRB Approval and Informed Consent

Oregon State University

Institutional Review Board • Office of Research Integrity
B308 Kerr Administration Building, Corvallis, Oregon 97331-2140
Tel 541-737-8008 | Fax 541-737-3093 | IRB@oregonstate.edu
http://oregonstate.edu/research/ori/humansubjects.htm

NOTIFICATION OF APPROVAL

December 10, 2010

Principal Investigator:	Joonkoo Yun, PhD	Department:	Nutrition and Exercise Science
Study Team Members:	Mike Pavol, PhD		
Student Researcher:	Susan Silverman, ATC, CSCS		
Study Number:	4811		
Study Title:	Effects of restricted ankle range of motion on human walking: An application to transtibial amputee gait patterns		
Funding Source:	None		
Submission Type:	Initial Application received 10/25/2010		
Review Category:	Expedited	Category Number:	4
Waiver(s):	None	Number of Participants:	20
Risk level for children[1]:	N/A		

The above referenced study was reviewed and approved by the OSU Institutional Review Board (IRB).

Approval Date: 12/10/2010
Expiration Date: 12/09/2011

Annual continuing review applications are due at least 30 days prior to expiration date

Documents included in IRB approval:

- ☒ Protocol
- ☒ Consent forms
- ☐ Assent forms
- ☐ Grant/contract
- ☐ Project revisions:
- ☒ Recruiting tools
- ☐ Test instruments
- ☐ Attachment A: Radiation
- ☐ Letters of support
- ☐ External IRB approvals
- ☐ Translated documents
- ☐ Attachment B: Human materials
- ☐ Other:

Comments:

Principal Investigator responsibilities for fulfilling the requirements of approval:

- All study team members should be kept informed of the status of the research.
- Any changes to the research must be submitted to the IRB for review and approval prior to the activation of the changes.
- Reports of unanticipated problems involving risks to participants or others must be submitted to the IRB within three calendar days.
- Only consent forms with a valid approval stamp may be presented to participants.
- Submit a continuing review application or final report to the IRB for review at least four weeks prior to the expiration date. Failure to submit a continuing review application prior to the expiration date will result in termination of the research, discontinuation of enrolled participants, and the submission of a new application to the IRB.

If you have any questions, please contact the IRB Office at IRB@oregonstate.edu or by phone at (541) 737-8008.

[1] Where parental permission is to be obtained, the IRB may find that the permission of one parent is sufficient for research to be conducted under §46.404 or §46.405. Where research is covered by §§46.406 and 46.407 and permission is to be obtained from parents, both parents must give their permission unless one parent is deceased, unknown, incompetent, or not reasonably available, or when only one parent has legal responsibility for the care and custody of the child.

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INFORMED CONSENT FORM

Project Title:	Effects of restricted ankle range of motion on human walking: An application to transtibial amputee gait patterns.
Principal Investigator:	Joonkoo Yun, Ph.D.
Student Researcher:	Susan Silverman ATC, CSCS
Co-Investigator(s):	Mike Pavol, Ph.D.
Sponsor:	Oregon State University department of Nutrition and Exercise Science
Version Date:	29November2010

1. WHAT IS THE PURPOSE OF THIS FORM?

This form contains information you will need to help you decide whether to be in this study or not. Please read the form carefully and ask the study team member(s) questions about anything that is not clear.

2. WHY IS THIS STUDY BEING DONE?

The purpose of this study is to see how people walk when they cannot move one of their ankles. This is because people who use prosthetic legs cannot move the prosthetic ankle when they are walking. This study is being done to see how people with two normal legs walk when they cannot move one of their ankles. This is to help researchers understand how people with leg amputations walk with their prosthetic legs and how to improve them. This study is a part of the master's degree thesis for Susan Silverman. The results of this study may be used for professional presentations and publications.

Up to 20 people (10 men and 10 women) will be invited to take part in this study.

3. WHY AM I BEING INVITED TO TAKE PART IN THIS STUDY?

You are being invited to take part in the study because you represent a "normal" walking pattern and we will be able to see how you change the way you walk when you cannot move one ankle.

4. WHAT WILL HAPPEN IF I TAKE PART IN THIS RESEARCH STUDY?

The study activities include recording your motion with a motion capture system. This will involve you coming to the Biomechanics lab in the Women's Building at Oregon State University one time for about 1.5 hours.

Once you arrive at the lab we will have you change into tight fitting clothing and your choice of athletic shoe. We will cover up anything shiny on your clothes or shoes. For the motion capture

we will tape 19 small silver Styrofoam spheres to your skin, clothing and shoes on your chest, back, pelvis, thighs, knees, shins and feet.
We will have you stand in the middle of our lab and record the markers while you are standing still. You will be asked to walk a few times across the walkway to get comfortable with walking in the lab. Then, you will be asked to walk across the platform 10 times in two different conditions. One condition will be walking normally, the other condition will be walking with your ankle held in place by a plaster cast.

We will hold your ankle in place using a plaster cast. This means we will put a cotton stockinette (like a sock with no toes) from the base of your toes to the middle of your calf, then put a layer of padding over the stockinette. Then we will put your ankle in a natural position and put plaster cast material (like plaster of Paris) over the stockinette and padding. It will take about 3-5 minutes for the plaster to dry. Once the plaster dries we will have you put your shoe back on over the cast and re-apply the markers. With the plaster cast on you will have 5 minutes to walk around the lab to get used to the cast before we have you walk in front of the motion analysis system.

When we are recording the marker movements we will have you walk across a walkway about 10 meters long at whatever speed you would comfortably walk at. You will walk across the walkway 10 times with your ankle held in place and 10 times normally.
Once you have walked across the platform 10 times normally and 10 times with your ankle held in place we will remove the markers, you can change clothes and then you are all done.

We will store the motion capture data which is the movement of the markers only to use for the results of this study. An image of your face or body will not be recorded and your name will not be connected to your data. We will keep this information on a computer which is in a locked room. Once the study is over we will delete this data.

The results of your data will not be shared directly with you but will be available as part of the thesis of Susan Silverman *Effects of restricted ankle range of motion on human walking: An application to transtibial amputee gait patterns.*

5. WHAT ARE THE RISKS AND POSSIBLE DISCOMFORTS OF THIS STUDY?

This study has minimal risk to you. You may feel some warmth while the plaster cast dries and feel slightly uncomfortable walking with the plaster cast. There is a minimal risk that you could fall during the walking trials.

6. WHAT HAPPENS IF I AM INJURED?

Oregon State University has no program to pay for research-related injuries. If you think that you have been injured as a result of being in this study please contact Dr. Joonkoo Yun jk.yun@oregonstate.edu 541-737-8584 or Susan Silverman silversu@onid.orst.edu 847-312-0407

6. WHAT ARE THE BENEFITS OF THIS STUDY?

This study is not meant to benefit you directly, but results from this study may help people with amputations by helping to design prosthetic legs.

7. WILL I BE PAID FOR BEING IN THIS STUDY?

You will receive a $10.00 gift card for being in this research study.

8. WHO IS PAYING FOR THIS STUDY?

N/A

9. WHO WILL SEE THE INFORMATION I GIVE?

The information you provide during this research study will be kept confidential to the extent permitted by law. Research records will be stored securely and only researchers will have access to the records. Federal regulatory agencies and the Oregon State University Institutional Review Board (a committee that reviews and approves research studies) may inspect and copy records pertaining to this research. Some of these records could contain information that personally identifies you. This information will be kept for three years after the completion of the study.

If the results of this project are published your identity will not be made public.

To help ensure confidentiality, we will use an ID number to track your data. Data will be kept on a password protected computer in a locked room.

Participation in this study is voluntary. If you decide to participate, you are free to withdraw at any time without penalty. You will not be treated differently if you decide to stop taking part in the study. If you choose to withdraw from this project before it ends, the researchers may keep information collected about you and this information may be included in study reports.

10. WHO DO I CONTACT IF I HAVE QUESTIONS?

If you have any questions about this research project, please contact: Dr. Joonkoo Yun jk.yun@oregonstate.edu 541-737-8584 or Susan Silverman silversu@onid.orst.edu 847-312-0407

If you have questions about your rights or welfare as a participant, please contact the Oregon State University Institutional Review Board (IRB) Office, at (541) 737-8008 or by email at IRB@oregonstate.edu

Your signature indicates that this research study has been explained to you, that your questions have been answered, and that you agree to take part in this study.

11. WHAT DOES MY SIGNATURE ON THIS CONSENT FORM MEAN?

Your signature indicates that this study has been explained to you, that your questions have been answered, and that you agree to take part in this study. You will receive a copy of this form.

Participant's Name (printed): ____

______________________________________ ____________________________
(Signature of Participant) (Date)

______________________________________ ____________________________
(Signature of Person Obtaining Consent) (Date)

Appendix B: Participant Eligibility Screening

Participant ID:

Height: ______________________________

Weight: ______________________________

Age: ________________________________

Gender: ______________________________

Knee Width:___________________________

Ankle Width: Dominant_______ Non-dominant_________Dominant with cast___________

Shoe length:___________________________

Cast weight:___________________________

	Yes	No
Participant is pain free in the lower extremities today	☐	☐
Participant has been diagnosed with a leg length discrepancy	☐	☐
Participant has a pair of comfortable athletic shoes with them today	☐	☐
Participant is allergic to latex	☐	☐

Limb Dominance	**R**	**L**
Step-up:	☐	☐
Fall recover:	☐	☐
Ball kick:	☐	☐

Dominant limb:_________________