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Loading of the Lower Extremity and Low Back When Using Wedge Orthotics during Walking and Stair Negotiation

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Loading of the lower extremity and low back when using wedge orthotics during walking and stair negotiation

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This Master's thesis is dedicated to my father, Gary Meyer (1945-2003), who taught me the value of dedication, persistence, and hard work.

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ABSTRACT

Wedge orthotics are commonly prescribed for patients with hyperpronation and/or low back pain to improve lower limb alignment and to reduce pain. The purpose of this study was to examine the kinetic effects of medial and lateral wedge orthotics during walking and stair negotiation. Twenty-two healthy young adults participated in the study. Each participant wore no wedge (W0) as a baseline and lateral and medial wedge orthotics at 3 degrees and 7 degrees bilaterally (L3, L7, M3, M7) during walking, stair ascent, and stair descent. Ankle, knee, and hip joint moments were calculated using inverse dynamics during the stance phase of walking and the second step of stair ascent and stair descent. L5S1 compression forces were calculated as the sum of L5S1 joint reaction forces and low back muscle forces. Repeated measures ANOVA was used to test for significant differences ($p < 0.05$). The L7 wedge significantly reduced external knee valgus moments during walking, but increased knee extension moments during stair ascent compared to W0. The M7 wedge significantly reduced ankle inversion moments during stair ascent and descent, but increased ankle eversion moments during walking and external knee varus moments during walking, stair ascent, and stair descent compared to W0. There were no effects of wedge orthotics on L5S1 compressive forces compared to W0. These results support the recommendation that when considering the use of a wedge orthotic, an individual's foot alignment, the degree of wedge angle, and effects at the knee joint need to be considered.

CHAPTER 1: GENERAL INTRODUCTION

Mechanical low back pain is the most common cause of disability in adults less than the age of 45 and second to arthritis between the ages of 45-65 years (Loney & Stratford, 1999). At least 85% of people will suffer from low back pain at some point during their lifetime (Andersson, 1999). The peak prevalence of back pain occurs between the ages of 40-60 (Loney & Stratford, 1999; Kent & Keating, 2005). The indirect and direct costs of low back pain are high, including insurance costs and loss of production. Low back pain is the second highest reason to be seen by a doctor and the third most common cause for surgery (Andersson, 1999). For the aforementioned reasons, creating strategies to prevent and treat low back pain are of clinical interest.

Rehabilitation of low back pain often focuses on enhancing lumbar spine stability. Traditional approaches to treating low back pain have not been as effective as once thought. Increasing spinal range of motion without proper stability can increase the risk of future injury. Spinal muscles are considered to be local stabilizers and provide stability and endurance to the lumbar spine, and stability should be achieved before forces are applied to the back (McGill, 2001). Another possible mechanism for low back pain is abnormal biomechanics (Cambron et al., 2011). There is evidence of a closed kinetic chain in which lower extremity alignment can be a factor in developing low back pain (Bird et al., 2003). Abnormal subtalar joint pronation is a suggested mechanism for low back pain due to a chain reaction into the pelvis and lumbar spine. However, the research examining this coupling behavior is conflicting (Duval et al., 2010; Khamis & Yizhar, 2007; Souza et al., 2010; Tateuchi et al., 2011; Pinto et al., 2008).

A common way to correct abnormal foot biomechanics is to use custom made orthotics. In chiropractic care, 81.8% of chiropractors prescribe orthotics to 20.9% of their patients (Cambron et al., 2011). Medial wedge orthotics are typically prescribed for low back pain patients to account for abnormal pronation and enhance foot stability (Cambron et al., 2011; Castro-Mendez et al., 2013). Orthotic use has been shown to alter the recruitment patterns of spinal musculature (Bird et al., 2003) and to improve low back pain (Castro-Mendez et al., 2013). Lateral wedge orthotics are commonly used for patients with knee osteoarthritis to relieve medial compression in the knee and reduce knee varus angle (Hinman et al., 2009; Hinman et al., 2012; Kerrigan et al., 2002; Bennell et al., 2011). Research has shown that lateral wedge orthotics are effective in decreasing pain in knee osteoarthritis patients. The question is how medial and lateral wedge orthotics change loading in the lower back, knee, hip, and ankle.

The coupling relationship between the subtalar joint and lumbopelvic region has most often been studied using a static standing position. Walking and stair negotiation are common daily tasks that involve dynamic balance and loading. Studying how the foot affects joints up the chain when using orthotics may prove beneficial in creating treatment strategies for common ailments like low back pain and knee osteoarthritis. Therefore, the current study is designed to look at how medial and lateral orthotics with different wedge angles affect loading in the lower extremity and low back while performing walking tasks, stair ascent, and stair descent. Each of these movements has its own challenges and may reveal different effects of wedge orthotics on the low back as well as the ankle, knee, and hip joints.

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CHAPTER 2: LITERATURE REVIEW

2.1 Prevalence of Low Back Pain

Mechanical low back pain is one of the most common complaints in society. It is reported that 70-85% of people will experience low back pain during their lifetime (Andersson, 1999). Low back pain is the second most often cited reason why a person seeks a physician's help and the third most common body region that is surgically operated upon (Andersson, 1999). Low back pain results in high financial costs, accounting for 68% of works days lost and 76% of total compensation costs (Andersson, 1999). In 1990, annual back pain direct costs in the United States were estimated to be $24,300,000,000 (Kent & Keating, 2005).

The prevalence of low back pain increases in middle age adults, ages between 40 and 60, while a decrease is observed after age 65 (Anderson, 1999; Loney & Stratford, 1999). The peak prevalence of low back pain occurs during the ages of 45-59 (Kent & Keating, 2005), and spinal impairments are the most frequently reported musculoskeletal impairments at 51.7% in people up to age 65 (Andersson, 1999). The prevalence is high in this decade of life because of continued involvement in the workforce. Compared to professionals and managers, craftsman, laborers, service workers, salesman, clerks, and farmers have the highest rate of low back pain (Leigh & Sheetz, 1989). The aforementioned occupations involve high force trunk exertions, awkward and repetitive lifting, and prolonged periods of standing and walking, often on uneven ground, which may increase repetitive stress on spinal musculature (Shin & Mirka, 2004).

The mechanisms of mechanical low back pain still remain unclear in many instances. There are many potential causes of low back pain including damage to bone, facet joints, intervertebral discs, nerves, and muscles in the lumbopelvic region. Lower extremity alignment can be a factor in the development of low back pain and could provide clinicians a better understanding of its pathology (Bird et al., 2003). A proposed mechanism for the development of low back pain is abnormal foot function, which may increase tension in the muscles of the lower back region and affect the motion of the lumbar vertebrae during gait (Castro-Mendez et al., 2013).

2.2 Functional Anatomy

The lumbosacral joint is located where the fifth lumbar vertebra joins with the sacrum. This junction is referred to as the L5/S1 joint and is a common site of low back pain. Proper stability in this joint is crucial to prevent low back pain (Vlemming et al., 2012). Sacroiliac joint pain accounts for 55% to 61.5% of low back pain cases (Delitto et al., 1993). Movement of the pelvis and sacrum involves the L5/S1 joint, and this movement can directly impact the lumbar vertebrae up the spine (Vlemming et al., 2012). Therefore, tightness or decreased range of motion in the hip and pelvis can negatively influence loading on the L5/S1 and lead to development of low back pain. Increasing or changing forces, particularly compressive forces, at the lumbar spine may increase the risk of low back injury or pain.

There is evidence of a closed kinetic chain reaction, beginning with foot contact as ground reaction forces are transmitted through the body (McPoil & Knecht, 1985). The sacrum and lower limbs are connected by fascial and muscular connections, supporting this chain reaction from the foot to the lumbopelvic complex (Vlemming et al., 2012).

For this reason, it may prove beneficial to examine mechanisms of low back pain globally through the lower extremity rather than locally at the L5/S1 joint (Vora et al., 2010).

Muscles that attach to the pelvis and spine play a direct role in stabilization and motion of the lumbosacral joint. These muscles include the gluteus maximus, gluteus medius, quadratus lumborum, lower lumbar multifidus, iliocostalis, erector spinae, and internal and external obliques. Force transmission from the legs to the upper body is also generated through the thoracolumbar fascia, which covers the posterior musculature of the back (Carvalhais et al., 2013). Thoracolumbar fascia is attached to the thoracic and lumbar spinous processes and serves as a connection for muscles such as the erector spinae, latissimus dorsi, transverse abominus, internal oblique, and gluteus maximus (Bogduk et al., 1984). Changes of recruitment patterns in aforementioned muscles are evident in low back pain patients during gait (Himmelreich et al., 2008; Vogt et al., 2003; Mehta et al., 2010; Silfies et al., 2005; Nelson-Wong et al., 2008; Van Dieen et al., 2003).

The foot is the foundation to human movement, and its function is to support body mass, provide postural balance, adapt to uneven ground, absorb shock, and transmit vertical ground reaction forces during gait (Barwick et al., 2012). The foot is divided into four sections: the rearfoot, midfoot, metatarsals, and phalanges. The talus and the calcaneus make up the rearfoot, and the subtalar joint (STJ) is the articulation between the talus and the calcaneus (Rockar, 1995). The talus sits between the calcaneus and the tibia of the lower leg.

Motion at the STJ is triplanar, meaning that motion includes inversion/eversion in the frontal plane, internal/external rotation in the transverse plane, and dorsiflexion/plantar flexion in the sagittal plane. Frontal plane movement in the STJ creates transverse plane motion in the tibiofemoral joint and hip. Subtalar joint neutral is measured by using the angle between the line that bisects the distal third of the lower leg and the line that bisects the calcaneus (Tiberio, 1988). However, due to concerns about reliability in measuring foot positions, like pronation, a standard approach has not been established.

2.3 Muscle Recruitment Patterns

Low back pain patients often display changes in muscle recruitment patterns. Muscle dysfunction is defined as "an unusual pattern of muscle recruitment during a prescribed set of movements" (Danneels et al., 2002). Patients with low back pain exhibit a decrease in range of motion for hip extension, resulting in a decreased activation of the biceps femoris (Vogt et al., 2003). A key role of the biceps femoris is to maintain normal nutation of the sacroiliac joint, creating a bracing action in the sacroiliac joint during the heel contact phase of gait (Bird et al., 2003). In addition to the dysfunction of the biceps femoris, changes in activation of the iliopsoas and quadratus lumborum also serve as potential contributors to low back pain (Bird et al., 2003).

Bird et al. (2003) studied the effects of various orthotics on the onset of erector spinae and gluteus medius activity during gait. A lateral foot wedge produced a decrease in the onset time of the erector spinae, while the unilateral heel lift condition delayed gluteus medius activation. The gluteus medius functions to abduct the hip, stabilize lateral pelvic tilt, and decelerate internal rotation at the hip during heel contact and stance phase in gait (Bird et al., 2003). In healthy individuals there is a synergistic relationship between the right and left gluteus medius during the gait cycle, while patients with low back pain exhibit more of an agonist/antagonist relationship (Nelson-Wong et al., 2008).

During normal gait, muscles contribute support for vertical ground reaction forces, with ankle dorsiflexors, gluteus maximus, vasti, and gluteus medius providing the most support during the early stance phase and from flat foot to contralateral toe-off (Anderson & Pandy, 2002). The use of lateral wedge orthotics results in changes for muscle recruitment patterns during gait (Bird et al., 2003) and can be a reliable tool to manipulate foot alignment (Tillman et al., 2003). Lateral wedge inserts has been used to alleviate medial knee osteoarthritis by reducing peak knee adduction moments (Hinman et al., 2012).

Changes in activation of hip extensor and pelvic stabilizing musculature may be a causative factor in development of low back pain (Himmelreich et al., 2008). Gluteal muscle recruitment patterns of chronic low back pain patients during level and incline walking, as well as stair ascent, differ from that of healthy individuals. Disturbances in the neuromuscular control of the gluteus maximus can cause pain in the sacroiliac joint, which is a form of chronic low back pain. The gluteus maximus is vital in providing stability of the lumbar spine during walking, running, lifting, and stair ambulation.

Himmelreich et al. (2008) used electromyography (EMG) of the gluteus maximus to determine recruitment patterns in chronic low back pain patients during various ambulatory tasks. During stair ascent, there was a prolonged recruitment of the gluteus maximus throughout the stance phase in the chronic low back pain patients compared to the healthy controls. This may be due to the increased need for pelvic stability during the push-up phase of stair ascent, where gluteus maximus recruitment is high (McFayden & Winter, 1988). EMG indicated that the recruitment of the erector spinae at the lumbar level and the gluteus maximus is prolonged during the stance phase of walking (Vogt et al., 2003), which is similar to the stair ascent results by Himmelreich et al. (2008).

The lumbar multifidus muscle provides local segmental stabilization to the lumbar vertebrae during movement, whether it is walking, stair negotiation, or lifting objects (Danneels et al., 2002). As a local stabilizer, the mutifidus muscle is designed for prolonged activation to provide stability. However, in low back pain patients compared to healthy individuals, the lumbar multifidus fatigues at an earlier onset which decreases the amount of stability (Danneels et al., 2002). There is also evidence that the lumbar multifidus has a smaller cross-sectional area in low back pain patients, which may affect its effectiveness as a stabilizer (Danneels et al., 2002).

The results of the aforementioned studies reveal that low back pain patients compensate the neuromuscular control of spinal and pelvic stabilizers to increase stability during ambulation to decrease pain (Vogt et al., 2003). A decrease in hip range of motion, especially hip extension and medial rotation, is also seen in low back pain patients and may be a causative factor of lower back, or sacroiliac joint pain (McGreggor & Hukins, 2009; Ellison et al., 1990; Cibulka et al., 1998). Passive measurements of hip medial and lateral rotation show that patients with low back pain demonstrate a decreased range of motion in medial rotation compared to healthy cohort.

There is often asymmetry in medial rotation for low back pain patients (Ellison et al., 1990; Cibulka et al., 1998). This asymmetry and decreased range of motion may result in low back pain. The lumbar spine has a small degree of rotation and if hip range of motion is diminished, the lumbar spine often becomes more mobile to provide efficient movement. In turn, increased lumbar spine mobility can change muscle recruitment patterns in the stabilizing muscles of the lumbopelvic region.

2.4 Normal and Pathological Gait

Kinetic and kinematic analyses of normal and pathological gait have been used to study human locomotion and can be used clinically in diagnosis of gait impairments. Winter et al. (1996) studied normal locomotion using three-dimensional analysis to measure joint moments at the ankle, knee, and hip during walking in the sagittal, frontal and transverse planes. In the sagittal plane, the ankle joint has a dorsiflexion moment initially in order to eccentrically lower the foot to the ground. As the foot begins to push-off, there is a large plantarflexion moment in the ankle produced by the gastrocnemius and soleus. In the transverse plane, ankle external rotation is present during stance phase of gait, and the foot/ankle goes through periods of eversion and inversion. A small ankle eversion moment occurs just after heel contact and transitions to an inversion moment during early stance to midstance. Lastly, the inversion moment reverts back to an eversion moment during the late push-off phase.

There is a hip external rotation moment during weight acceptance of stance followed by an internal rotation moment during mid-stance to late stance. During late stance to early toe off, there is a greater range of motion needed in hip extension. These motions help to control the transverse rotation of the pelvis during the single support phase of gait (Winter et al., 1996). During gait, there is a high abductor moment at the hip to control upright posture of the head and trunk. The hip abductors and adductors control the medial/lateral balance of the body, while the ankle plantarflexors and dorsiflexors control the anterior/posterior motion during loading and unloading (Winter et al., 1996).

During heel strike, the calcaneus everts and the STJ pronates, which allows the foot to become flexible to adapt to the terrain and absorb shock (McPoil & Knecht, 1985). At midstance, the foot supinates until toe-off, creating a rigid base to propel the body forward (McPoil & Knecht, 1985). Pronation and supination motion of the STJ is coupled with motion at the tibia and femur (McPoil and Knecht, 1985; Tiberio, 1988). At heel contact, the tibia internally rotates until 25% of stance, at which point the tibia externally rotates (Botte, 1981).

Abnormal foot motion is considered to be an excessive amount of pronation occurring beyond midstance when the foot should begin to supinate (Moseley et al., 1996). Asynchronous timing of pronation to supination affects the coupling behavior of the tibia and femur and has been liked to lower limb injuries like plantar fasciitis and patellofemoral pain (Duval et. al., 2010.). As a result of this delay in external rotation at toe-off, there is greater strain exerted on the sacroiliac and sacrolumbar joints (Botte, 1981). These joints do not permit a large amount of motion and the extra strain creates a state of hypermobility.

Following heel strike, the pelvis is negatively tilted, the lower thoracic spine is extended, and the lumbar spine is flexed. During early swing phase, the lower thoracic and lumbar spine laterally flexes toward the weight bearing limb, and the pelvis laterally flexes toward the swing leg (Crosbie et al., 1997). Around the double support phase, both the pelvis and spine return to neutral. During heel contact, the spine rotates to the side of heel contact (right rotation and right heel contact), followed by a rotation to the opposite side before the next heel strike (Callaghan et al., 1999; Crosbie et al., 1997). Schache et al. (2002) studied 3D angular kinematics of the lumbar spine and pelvis during running. At right foot heel strike, the lumbar spine and pelvis rotate right. During stance, the lumbar spine rotates left and reaches its left rotation peak by right toe-off and during swing, it begins to rotate right again (Schache et al., 2002).

The pelvis follows a slightly different path than the lumbar spine. Peak right rotation occurs just prior to mid-stance and in the latter half of stance, the pelvis begins left rotation and reaches neutral by toe-off. After toe-off, the pelvis continues to rotate left. Schache et al. (2002) found a strong significant inverse correlation of flexion/extension of the lumbar spine with anterior-posterior tilt of the pelvis (r=-.084), and lateral bending of the lumbar spine with obliquity of the pelvis (r=-0.75). As angular rotations of the pelvis were more positive, the lumbar spine rotations were more negative (Schache et al., 2002).

2.5 Stair Ambulation

Stair ascent and descent are common daily activities that produce greater magnitudes of hip, knee, and ankle kinematics and kinetics than level walking (Protopapadaki et al., 2007; Lee & Park, 2011). Vertical ground reaction forces are the greatest during stair descent at the beginning of stance phase. During stair ascent, there are greater hip extension and knee flexion moments as compared to descent (Protopapadaki et al., 2007). Stair ascent is considered to be the more demanding task in young healthy individuals. While Protopapadaki et al. (2007) looked at sagittal plane kinematics and kinematics of stair ascent and stair descent, Nadeau et al. (2003) studied frontal plane challenges during stair ascent as compared to level walking. Stair ascent required a substantial amount of effort in the frontal plane at the hip to control the pelvis. Hip abductor muscles control the lateral pelvic obliquity so that the contralateral leg can swing properly to the next step.

During stair ascent, the hip abductors demonstrate a concentric action to raise the pelvis on the contralateral side (Nadeau et al., 2003). McFayden and Winter (1988) observed that at 10 percent of stance, which is the push-up phase, the gluteus maximus is working concentrically to propel the body upward. Adequate strength in the gluteal complex is important in stabilizing the lumbopelvic region during walking as well as stair ambulation (Himmelreich et al., 2008). Hip net joint powers are characterized by more positive power compared to level walking in the frontal plane and at the knee and ankle joint in the sagittal plane (Nadeau et al., 2003).

Joint moments at the hip, knee, and ankle are different in magnitude during stair ascent compared to level walking. Knee and hip flexion/extension moments are greater during stair ascent than in level walking, with the largest difference occurring at the knee joint (Andriacchi et al., 1980). Studies show stair ascent to be more demanding task than descent and level walking due to the increase in greater knee extension moments and hip moments (Protopapadaki et al., 2007). Stair negotiation is a frequently encountered obstacle and produces different motion in the spine than level walking (Lee & Park, 2011). Normal kinetics and kinematics of walking and stair negotiation are well established. However, the question remains whether or not excessive STJ pronation changes the normal kinetic and kinematic behaviors of gait and stair negotiation and if these changes cause pain and/or injury in the knee, hip, and low back.

2.6 Subtalar Joint Motion and Injuries

STJ pronation that is greater in amplitude and prolonged in duration is considered excessive during gait (Duval et al., 2010). This abnormal function of the foot has been linked to overuse injuries such as plantar fasciitis, patellofemoral pain, and mechanical low back pain (Pohl et al., 2009). The planter fascia functions to maintain the medial longitudinal arch of the foot and absorb forces (Cheung et al., 2006). As the foot supinates during toe-off, the plantar fascia is pulled tight and the foot becomes rigid, referred to as the “windlass mechanism” (Bolga & Malone, 2004).

The appropriate timing of pronation to supination affects the stiffness of the plantar fascia during push-off. A longer duration of pronation, past midstance, creates a more flexible foot for push-off and decreases the windlass mechanism, which puts greater stress on the plantar fascia (Bolga & Malone, 2004). Foot pronation is a combination of ankle dorsiflexion, eversion, and abduction (Moseley et al., 1995). Ankle dorsiflexion is needed during gait, and a tight Achilles tendon can limit the range of motion. Excessive pronation is a common compensation for limited dorsiflexion (Pohl et al., 2009).

Excessive STJ pronation is considered to be a contributing factor to the development of patellofemoral pain syndrome (Tiberio, 1987). During the midstance and push off phase of gait when the foot is supinating, the tibia and femur begin to externally rotate to allow for knee extension. Excessive pronation delays the timing of supination and the tibia is not able to externally rotate (Tiberio, 1987). This disruption in timing increases the joint compression on the lateral surface of the patella and affects patellar tracking (Tiberio, 1987). Prolonged STJ pronation or rearfoot eversion during the stance phase of gait increases the loading forces at the knee (Levinger & Gilleard, 2007). Abnormal STJ pronation may cause plantar fasciitis and patelleofemoral pain, but it remains unclear what the effects of abnormal pronation are on the pelvis and lower back, and if there is a relation to the development of lower back pain.

2.7 Relationship of Subtalar Joint and Lumbopelvic Region

STJ pronation causes internal rotation of the tibia and femur, and supination causes external rotation of the tibia and femur (Duval et al., 2010; Khamis & Yizhar, 2007; Souza et al., 2010). Those who exhibit longer periods of foot pronation also show longer periods of hip internal rotation (Souza et al., 2010). However, it remains unclear is whether or not hip internal rotation increases lumbar lordosis or anterior pelvic tilt. Khamis and Yizhar (2007) observed a change in pelvic alignment with an increase in foot pronation, while Duval et al. (2010) did not find a significant change. Khamis and Yizhar (2007) and Duval et al. (2010) agree that excessive pronation causes greater internal rotation in the tibia, femur, and hip. However, their results conflict in regards to whether or not pronation increases lumbar lordosis. Detecting changes in the pelvis and low back may be difficult in the sagittal plane, while changes may be more readily detected in the frontal plane.

Excessive calcaneal eversion during unilateral stance increases lateral pelvic tilt toward the standing leg (Pinto et al., 2008), and an alteration in the alignment of lumbopelvic complex increases the risk of developing low back pain (Tateuchi et al., 2011). A change in pelvic alignment is considered to increase strain on muscles of the lumbopelvic region. The muscles commonly affected are the iliopsoas, piriformis, and gluteal muscles (gluteus medius and gluteus maximus). Lumbosacral instability is believed to be a risk factor for low back pain and can be caused by extra strain on the sacroiliac joint (Barwick et al., 2012). Lateral tilt of the lumbar spine and axial rotation of the thoracic spine increases during calcaneal eversion in unilateral stance.

Tateuchi et al. (2011) determined that thoracic rotation without pelvic rotation during excessive pronation increases lumbar spine rotation, and this rotation is a risk factor for mechanical low back pain. The lumbar spine, due to its larger vertebral bodies, is designed for stability and transmitting forces, not for rotation. Rotational movements are reserved for the thoracic spine. A unilateral increase in pronation creates a length discrepancy causing lateral tilt of the pelvis, resulting lumbar scoliosis which could place more stress on the facet joints of the vertebrae (Pinto et al., 2008). There is a relationship between abnormal pronation and leg length differences, and the shorter leg associated with the greatest amount of pronation (Vink & Hudson, 1988). Excessive foot pronation could cause a more anteriorly titled pelvis and place increased strain on the muscles of the hip and pelvis due to lumbosacral instability.

Coupling between calcaneal eversion and tibiofemoral internal rotation and changes in pelvic alignment are difficult to see in quite standing kinematic measurements. Measuring these coupling behaviors may be easier to see in more dynamic movements like walking and stair negotiation. Souza et al. (2010) revealed that the relationship between calcaneal eversion, tibial rotation, and hip internal rotation is strongly correlated during walking. These rotations are relatively synchronous, which supports the findings of McPoil and Knecht (1985) and Tiberio (1988). There is strong theoretical basis for this relationship; however, the empirical evidence is still lacking on the effects of changes in the lumbopelvic region during increased calcaneal eversion (Barwick et al., 2012). Abnormal and asymmetrical pronation and leg length differences may lead to mechanical low back pain due to the increased strain on the sacroiliac joint (Botte, 1981).

2.8 Effects of Orthotics

Excessive calcaneal eversion produces internal rotation at the hips and increases pelvic anterior tilt and lumbar hyperlordosis. As a result of this chain reaction, foot posture can alter pelvic and spinal alignment (Castro-Mendez et. al, 2013). Castro-Mendez and colleagues (2013) researched the effect of custom made foot orthoses with subjects with excessive foot pronation and low back pain. They found that those who wore orthotics to prevent excessive foot pronation had decreased low back pain compared to the control group, who did not wear an orthotic. Low back pain did not disappear, but it did improve significantly.

Similarly, Rothbart et al. (1995) studied the effect of medial posted orthotics on 208 low back pain patients. All participants displayed excessive pronation. Over eighty percent of the subjects reported an improvement in low back pain when using medial posted orthotics (Rothbart et al., 1995). Dananberg and Guiliano (1999) studied 32 patients with low back pain, and those that used foot orthotics had twice the improvements in pain compared to the control group. The results of the Castro-Mendez et al. (2013), Rothbart et al. (1995) and Dananberg and Guiliano (1999) studies provide indirect evidence that foot function plays a part in spinal alignment and low back pain.

Orthotics are commonly used to correct/control excessive STJ pronation during the stance phase of gait (Castro-Mendez et al., 2013) and are commonly used in treatment of low back pain (Bird et al., 2003). By changing the alignment of the foot and lower limb, orthotics can change loading patterns and alleviate pain or discomfort in those with low back pain and knee osteoarthritis (Kelaher et al., 2000; Bennell et al., 2011). Similar to Castro-Mendez et al. (2013), Kelaher et al. (2000) also observed an improvement in low back pain/comfort while wearing orthotics during a fatiguing exertion.

Orthotics generally span the length of the foot to the first metatarsal head (Kelaher et al., 2000; Castro-Mendez et al., 2012; Hinman et al., 2009). Ethylene vinyl acetate with a Shore durometer type A reading of 75 is a common material used to make orthotics because it resist deformation over time (Hinman et al., 2009; Bennell et al. 2011; Bird et al., 2003). Durometer is a measure of hardness of a material or the materials resistance to deformation. Shore type A durometer is used to measure softer rubbers and plastics that include vinyls. Shore A scale larger than 100 is used for the hardest materials. A reading of 75 is commonly used in shoe heel inserts.

A flattened arch can cause the talus to move medially and the calcaneus to evert. This posture causes poor body weight transfer through the medial longitudinal arch (Kelaher et al., 2000). Foot pronation is evident during the heel contact and stance phase of gait, but it should not be present in quiet standing. Greater pronation in one foot compared to the other can create a leg length difference causing lateral pelvic tilt and increasing lateral shear forces on the lumbar spine (Danbert, 1988). Orthotics that control foot posture, especially excessive STJ pronation or flattened arch, may alleviate low back pain. Foot orthotics can change the onset of muscle activity of the spine and pelvis, particularly the delayed onset of erector spinae and the gluteus medius muscles during walking (Bird et al., 2003).

In summary, a review of the literature indicates that the peak prevalence of chronic low back pain is between the ages of 45-65 and increases both economic and sociological burdens on the sufferer as well as employers. Instability in the low back musculature is a factor in creating low back pain. What remains to be answered is how lower extremity posture, particularly the foot, influences lower back loading. Previous research points to the existence of a kinetic chain that links abnormal foot pronation to injuries of the lower limb, such as plantar fasciitis and patellofemoral pain. The theory that this chain reaction affects the mechanics of the lumbosacral region has been studied in static standing conditions with conflicting results.

Research on orthotics shows differences in medial and lateral shoe inserts on muscle recruitment patterns and changes in ankle, knee and hip joint moments. Using orthotics is a viable way to manipulate foot posture and motion of the STJ to measure the effects foot posture has on the knee and hip, as well as the lumbosacral joint. Low back pain is one of the leading pain issues affecting industrialized populations, and up to 80% of people will suffer from back pain at some point in their lifetime. As a result of this high prevalence, it would be beneficial to research how a kinetic chain reaction of distal joints, beginning with the STJ, affect loading in the lower back. This information may provide clinicians greater understanding of causative factors of low back pain and help in developing assessment and treatment strategies.

The purpose of this study was to examine the effects of different orthotic wedge angles on the lower extremity joint moments and L5/S1 compression forces during level walking, stair ascent, and stair descent. Our hypotheses were: 1) medial wedge orthotics would reduce ankle inversion moments, 2) lateral wedge orthotics would reduce external knee varus moments, and 3) medial wedge orthotics would reduce L5/S1 compression forces during walking, stair ascent, and stair descent. Restoring normal gait patterns is an important goal in rehabilitation of injuries and chronic pain. Finding a link between excessive foot pronation and increased low back loading would inform evaluation and correction of foot mechanics that may help in reducing the prevalence of low back pain. A better understanding of the effects of foot function on the lower back could lead to improvements in treatments or prevention strategies and decrease the associated economic costs.

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CHAPTER 3: LOADING OF THE LOWER EXTREMITY AND LOW BACK WHEN USING WEDGE ORTHOTICS DURING WALKING AND STAIR NEGOTIATION

A paper to be submitted to *Gait and Posture*

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Abstract

Wedge orthotics are commonly prescribed for patients with hyperpronation and/or low back pain to improve lower limb alignment and to reduce pain. The purpose of this study was to examine the kinetic effects of medial and lateral wedge orthotics during walking and stair negotiation. Twenty-two healthy young adults participated in the study. Each participant wore no wedge (W0) as a baseline and lateral and medial wedge orthotics at 3 degrees and 7 degrees bilaterally (L3, L7, M3, M7) during walking, stair ascent, and stair descent. Ankle, knee, and hip joint moments were calculated using inverse dynamics during the stance phase of walking and the second step of stair ascent and stair descent. L5/S1 compression forces were calculated as the sum of L5/S1 joint reaction forces and low back muscle forces. Repeated measures ANOVA was used to test for significant differences ($p < 0.05$). The L7 wedge significantly reduced external knee valgus moments during walking, but increased knee extension moments during stair ascent compared to W0. The M7 wedge significantly reduced ankle inversion moments during stair ascent and descent, but increased ankle eversion moments during walking and external knee varus moments during walking, stair ascent, and stair descent compared to W0. There were no effects of wedge orthotics on L5/S1 compressive forces compared to W0. These results support the recommendation that

when considering the use of a wedge orthotic, an individual's foot alignment, the degree of wedge angle, and effects at the knee joint need to be considered.

1. Introduction

Low back pain accounts for 51.7% of musculoskeletal impairments (Andersson, 1999), and the economic burdens on industrialized populations are high. Work disability and absenteeism create significant indirect costs of low back pain for the employee as well as the employer by increasing compensation costs and decreasing productivity. Sixty-eight percent of work days lost are due to low back pain, and those lost work days account for 76% of total work compensation costs (Andersson, 1999). In 2005, low back pain accounted for $17.7 billion in direct healthcare costs, which include physician visits, medications, physical therapy, and other treatments (Andersson, 1999; Parthan et al., 2006). The peak prevalence of low back pain occurs between the ages of 45 to 60 years (Andersson, 1999; Kent & Keating, 2005; Loney & Stratford, 1999). Occupations that involve a great deal of awkward and repetitive lifting, high trunk exertion, and/or prolonged periods of standing and walking may increase repetitive stress on spinal musculature (Shin & Mirka, 2004; Barwick et al., 2012).

Spinal instability in the lumbopelvic region is a contributor to mechanical low back pain (McGill, 2001; Panjabi, 2003; Barwick et al., 2012). Lower limb alignment and mechanics may also be a factor in the development of low back pain (Bird et al., 2003; Castro-Mendez et al., 2013). However, the effects of lower limb mechanics, specifically foot mechanics, on the lumbopelvic complex and the development of low back pain remain unclear. Previous studies have examined the coupling relationship between the subtalar joint (STJ) and the tibiofemoral joint and have found a chain reaction from the foot to the femur. Improper mechanics of the STJ, such as hyperpronation, can lead to lower limb injuries including patellofemoral pain, plantar fasciitis, and iliotibial band syndrome (Duval et al., 2010; Pohl et. al., 2009; Bolga & Malone, 2004; Moseley et al., 1996; Tiberio, 1987; Levinger & Gilleard, 2007).

It is established that STJ pronation causes internal rotation of the tibia and femur, while supination causes external rotation (Duval et al., 2010; Khamis & Yizhar, 2007; Souza et al., 2010). Whether or not greater STJ pronation/supination and tibia/femur internal/external rotation create a chain reaction that extends to the pelvis and low back is not well documented. If greater anterior tilt and lumbar lordosis results from increased STJ pronation or supination, then loading may be increased at the L5/S1 joint (Duval et al., 2010; Khamis & Yizhar, 2007). Studies examining relationships between lower extremity and lumbopelvic movement are often done in static standing (Tateuchi et al., 2011, Pinto et al., 2008), looking solely at kinematic changes in the sagittal plane. Measuring the kinetic relationships between the STJ and lumbopelvic region during walking and stair negotiation may provide important explanations for mechanisms of low back pain.

Orthotics have been widely used in patients with knee osteoarthritis and low back pain (Castoro-Mendez et al., 2013; Rothbart et al., 1995; Dananberg & Guiliano, 1999; Kelaher et al., 2000; Hinman et al., 2009; Hinman et al., 2012; Bennell et al., 2011) to correct for lower limb misalignments and decrease pain. The use of orthotics is a reliable way to manipulate foot posture (Tillman et al., 2003) and can serve as a tool to studying and understanding the effects of foot alignment on the lumbopelvic region. A medial wedge orthotic is expected to reduce STJ overpronation and may be prescribed to help reduce low back pain. In contrast, a lateral wedge orthotic is expected to increase STJ pronation, but may help correct knee varus alignment associated with knee osteoarthritis.

The purpose of the current study was to examine the kinetic changes placed on the lower extremity and low back during walking and stair negotiation by manipulating foot alignment using bilateral medial and lateral shoe orthotics with different wedge angles. Our expectation was that foot alignment creates a chain reaction through the lower extremity to the lumbopelvic region during walking and stair negotiation. Our hypotheses were: 1) medial wedge orthotics would reduce ankle inversion moments, 2) lateral wedge orthotics would reduce external knee varus moments, and 3) medial wedge orthotics would reduce L5/S1 compression forces during walking, stair ascent, and stair descent.

2. Methods

2.1 Research Participants

Twenty-two subjects (10 male and 12 female, age 49 $\pm$ 6 years, height 1.77 $\pm$ 0.10 m, mass 80.3 $\pm$ 14.4 kg) participated in the study. Participants were between 40-60 years of age and free from low back and lower limb injuries or pain. Individuals were excluded if they had any current injuries or balance conditions that would affect walking or stair negotiation. Each participant provided informed consent and filled out a health history questionnaire prior to any data collection. The research protocol was approved by the Institutional Review Board of Iowa State University.

2.2 Medial and lateral wedge orthotics

Medial and lateral wedge orthotics were placed in the participant's left and right shoes to manipulate foot pronation and supination angles. Angles analyzed were no wedge (W0), three degrees of medial and lateral wedge (M3, L3) and seven degrees of medial and lateral wedge (M7, L7). The W0 condition was used to measure unadjusted foot motion of the subjects. Wedges were made from ethylene vinyl acetate with a shore durometer type A reading of 75. The wedges were designed, made, and donated for this research study by Marathon Orthotics, Minneapolis, Minnesota.

2.3 Data collection

After participants provided informed consent and filled out the health history questionnaire, body mass and height were recorded. Twenty-one reflective markers, 1.9 cm in diameter, were placed on the skin and clothing. Participants were instructed to wear spandex shorts and shirt to minimize movement of clothing for data collection. Participants wore their normal walking or running shoes. Two subjects had custom made orthotics or over the counter orthotics which were removed in order to use the wedge orthotics provided for the study. Markers were placed on the right toe, fifth metatarsal head, heel, medial and lateral malleoli, lateral and anterior calf, medial and lateral knee joint line, and lateral and anterior thigh. Bilateral markers were placed on the greater trochanters, anterior superior iliac spines (ASIS), posterior superior iliac spines (PSIS), and acromion processes. Single markers were placed on the sacrum and cervicale. Following a static standing trial, the heel, medial malleoli, and medial knee markers were removed. The removed markers were recreated during the dynamic trials using transformations based on the static model.

Participants performed three trials of walking, stair ascent, and stair descent with no wedge and with the four wedges for a total of 45 trials at their own preferred walking speed. The walkway used in the research was 6 meters in length with a force platform (AMTI, Watertown, MA, USA) imbedded in the laboratory floor to measure ground reaction forces during walking. A successful walking trial was defined as a clean right foot strike on the force platform without any visible signs of targeting. The staircase consisted of three steps (step height 18.5 cm, tread depth 29.5 cm) with a force platform on the first and second step (AMTI, Watertown, MA, USA). Participants completed stair ascent and stair descent with a left foot lead and step-over-step technique. The order of the wedges tested was randomized as was the order of walking and stair negotiation.

Kinematic data were measured using an 8-camera, three-dimensional motion analysis system (Vicon Nexus, Los Angeles, USA) at a sampling rate of 160 Hz. Noise was reduced in the kinematic and kinetic data using a fourth order, symmetric Butterworth filter with a cut-off frequency of 6 Hz. Segment masses, centers of mass, and moments of inertia were estimated on an individual basis (de Leva, 1996). Ankle, knee, and hip moments were calculated using inverse dynamics during the stance phase of walking and during the stance phase of the second step of stair ascent and descent. Joint moments were expressed in the distal segment coordinate system and normalized by body mass. Knee varus moments were expressed as external moments (opposite sign of internal moment) as commonly reported in knee osteoarthritis studies. Kinetic and kinematic data were processed using custom code created in Matlab (Mathworks Inc. Natick, MA, USA).

2.4 Lumbar spine model

A lumbar spine model was developed to estimate L5/S1 compression forces. L5/S1 joint reaction forces and joint moments were calculated during single leg stance using inverse dynamics. Pelvic segment mass, center of mass, and moment of inertia were defined using the lower part of the trunk in the de Leva (1996) anthropometric model. The L5/S1 joint center was estimated to be located 33% from the midpoint of the PSIS markers to the midpoint of the ASIS markers. Erector spinae (5.6 cm) and rectus abdominus (5.9 cm) muscle moment arms were determined using OpenSim (Delp et al., 2007). Erector spinae muscle forces were calculated by dividing the L5/S1 extension moments by the muscle moment arm, and rectus abdominus muscle forces were calculated by dividing the L5/S1 flexion moments by the muscle moment arm. L5/S1 compression forces were then calculated as the sum of L5/S1 joint reaction forces and muscle forces, then normalized by body mass.

2.5 Statistical analysis

The average of the three trials for each condition was used in the statistical analysis. Repeated measures ANOVA was used to compare peak joint moments at the ankle, knee, and hip and L5/S1 compressive forces during the five wedge conditions in the frontal and sagittal plane. A separate analysis was performed for walking, stair ascent, and stair descent. Statistical analyses were programmed using SPSS for Windows (Verison 12.0; SPSS Chicago, IL., USA). A Bonferroni adjustment for multiple wedge comparisons was applied and statistical significance was set at $p \leq 0.05$.

3. Results

3.1 Walking

The L7 wedge orthotic significantly reduced the maximum ankle plantarflexion moment compared to the M3 wedge ($p = 0.041$) and the external knee valgus moment compared to the W0 condition ($p = 0.025$) during walking (Table 1). The M7 wedge significantly increased the maximum external knee varus moment compared to W0 ($p = 0.002$), L3 ($p = 0.001$), L7 ($p < 0.001$), and M3 ($p = 0.038$) conditions. In addition, the M7 wedge increased the maximum hip extension moment ($p = 0.016$) and the maximum L5/S1 compression force ($p = 0.050$, Table 2) compared to the M3 wedge. The M7 wedge also significantly increased the maximum ankle eversion moment compared to the W0 condition ($p = 0.033$).

3.2 Stair Ascent

The L7 wedge orthotic significantly increased the maximum knee extension moment compared to W0 ($p = 0.027$), M3 ($p = 0.012$), and M7 ($p = 0.10$) conditions during stair ascent (Table 3). The M7 wedge significantly reduced the maximum ankle inversion moment compared to the W0 condition ($p = 0.005$). However, the M7 wedge significantly increased the maximum external knee varus moment compared to W0 ($p = 0.049$), L3 ($p = 0.002$), and L7 ($p < 0.001$) conditions. There were no significant differences in maximum L5/S1 compression force when comparing the W0 condition and four wedge angles for stair ascent (Table 2).

3.3 Stair Descent

The M7 wedge orthotic significantly reduced the maximum ankle inversion moment compared to W0 ($p < 0.001$), L3 ($p < 0.001$), L7 ($p = 0.001$), and M3 ($p = 0.004$)

conditions during stair descent (Table 4, Figure 1). However, the M7 wedge significantly increased the maximum external knee varus moment compared to W0 ($p = 0.012$) and L7 ($p = 0.006$) conditions. There were no significant differences in maximum L5/S1 compression force when comparing the W0 condition and four wedge angles for stair ascent (Table 2).

Table 1. Peak joint moments as a function of wedge orthotic angle during walking. Average values are reported with standard deviations. Statistical significance ($p < 0.05$): a – greater than W0, b – greater than L3, c – greater than L7, d – greater than M3.

Moment (Nm/kg)	**W0**	**L3**	**L7**	**M3**	**M7**
Ankle Plantarflexion	1.43 ± 0.15	1.43 ± 0.16	1.42 ± 0.17	**1.45 ± 0.17**[c]	1.44 ± 0.17
Ankle Eversion	0.08 ± 0.10	0.08 ± 0.09	0.08 ± 0.08	0.08 ± 0.10	**0.09 ± 0.10**[a]
Ankle Inversion	0.08 ± 0.06	0.08 ± 0.05	0.09 ± 0.05	0.08 ± 0.05	0.07 ± 0.05
Knee Extension	0.47 ± 0.16	0.47 ± 0.16	0.48 ± 0.16	0.47 ± 0.16	0.46 ± 0.15
Knee Flexion	0.31 ± 0.09	0.31 ± 0.09	0.30 ± 0.08	0.30 ± 0.09	0.30 ± 0.09
Knee Varus	0.51 ± 0.24	0.51 ± 0.23	0.51 ± 0.23	0.52 ± 0.23	**0.54 ± 0.23**[abcd]
Knee Valgus	**0.07 ± 0.03**[c]	0.06 ± 0.03	0.06 ± 0.03	0.06 ± 0.04	0.07 ± 0.04
Hip Extension	0.77 ± 0.14	0.77 ± 0.14	0.76 ± 0.14	0.76 ± 0.14	**0.79 ± 0.15**[d]
Hip Flexion	0.82 ± 0.21	0.82 ± 0.21	0.81 ± 0.21	0.82 ± 0.21	0.83 ± 0.22
Hip Abduction	0.89 ± 0.09	0.90 ± 0.09	0.90 ± 0.09	0.90 ± 0.09	0.90 ± 0.09

Table 2. Peak L5/S1 compression forces as a function of wedge orthotic angle during walking, stair ascent, and stair descent. Average values are reported with standard deviations. Statistical significance ($p < 0.05$): d – greater than M3.

L5S1 Force (BW)	**W0**	**L3**	**L7**	**M3**	**M7**
Walking	2.61 ± 0.25	2.60 ± 0.25	2.59 ± 0.28	2.58 ± 0.25	**2.64 ± 0.26**[d]
Stair Ascent	2.61 ± 0.43	2.63 ± 0.47	2.60 ± 0.42	2.67 ± 0.46	2.64 ± 0.44
Stair Descent	2.41 ± 0.34	2.44 ± 0.41	2.48 ± 0.41	2.44 ± 0.37	2.46 ± 0.41

Table 3. Peak joint moments as a function of wedge orthotic angle during stair ascent. Average values are reported with standard deviations. Statistical significance ($p < 0.05$): a – greater than W0, b – greater than L3, c – greater than L7, d – greater than M3, e – greater than M7.

Moment (Nm/kg)	**W0**	**L3**	**L7**	**M3**	**M7**
Ankle Plantarflexion	1.41 ± 0.15	1.42 ± 0.17	1.40 ± 0.15	1.44 ± 0.16	1.44 ± 0.16
Ankle Eversion	0.07 ± 0.06	0.07 ± 0.06	0.07 ± 0.06	0.07 ± 0.06	0.08 ± 0.06
Ankle Inversion	**0.08 ± 0.06**[e]	0.08 ± 0.06	0.08 ± 0.06	0.07 ± 0.05	0.06 ± 0.05
Knee Extension	0.88 ± 0.23	0.90 ± 0.24	**0.94 ± 0.24**[ade]	0.90 ± 0.26	0.90 ± 0.24
Knee Flexion	0.23 ± 0.11	0.23 ± 0.11	0.24 ± 0.14	0.23 ± 0.13	0.22 ± 0.12
Knee Varus	0.52 ± 0.22	0.50 ± 0.20	0.51 ± 0.22	0.53 ± 0.21	**0.55 ± 0.21**[abc]
Knee Valgus	0.03 ± 0.03	0.03 ± 0.03	0.03 ± 0.03	0.03 ± 0.03	0.03 ± 0.03
Hip Extension	0.77 ± 0.21	0.79 ± 0.24	0.78 ± 0.22	0.81 ± 0.25	0.79 ± 0.22
Hip Flexion	0.16 ± 0.11	0.15 ± 0.09	0.15 ± 0.09	0.17 ± 0.12	0.15 ± 0.08
Hip Abduction	0.93 ± 0.12	0.93 ± 0.11	0.93 ± 0.13	0.94 ± 0.12	0.94 ± 0.11

Table 4. Peak joint moments as a function of wedge orthotic angle during stair descent. Average values are reported with standard deviations. Statistical significance ($p < 0.05$): a – greater than W0, c – greater than L7, e – greater than M7.

Moment (Nm/kg)	**W0**	**L3**	**L7**	**M3**	**M7**
Ankle Plantarflexion	1.21 ± 0.16	1.20 ± 0.17	1.20 ± 0.16	1.22 ± 0.21	1.21 ± 0.18
Ankle Eversion	0.11 ± 0.06	0.10 ± 0.06	0.11 ± 0.06	0.11 ± 0.06	0.12 ± 0.07
Ankle Inversion	**0.10 ± 0.06**[e]	**0.09 ± 0.06**[e]	**0.09 ± 0.06**[e]	**0.09 ± 0.07**[e]	0.07 ± 0.06
Knee Extension	1.09 ± 0.19	1.08 ± 0.21	1.11 ± 0.21	1.10 ± 0.22	1.10 ± 0.25
Knee Flexion	0.16 ± 0.08	0.16 ± 0.08	0.16 ± 0.08	0.16 ± 0.07	0.17 ± 0.07
Knee Varus	0.61 ± 0.25	0.62 ± 0.27	0.62 ± 0.26	0.64 ± 0.26	**0.65 ± 0.26**[ac]
Knee Valgus	0.02 ± 0.03	0.01 ± 0.03	0.02 ± 0.03	0.01 ± 0.03	0.01 ± 0.03
Hip Extension	0.43 ± 0.16	0.44 ± 0.20	0.47 ± 0.18	0.44 ± 0.20	0.45 ± 0.19
Hip Flexion	0.35 ± 0.16	0.33 ± 0.15	0.33 ± 0.15	0.34 ± 0.17	0.33 ± 0.16
Hip Abduction	0.95 ± 0.16	0.97 ± 0.13	0.97 ± 0.13	0.96 ± 0.12	0.97 ± 0.14

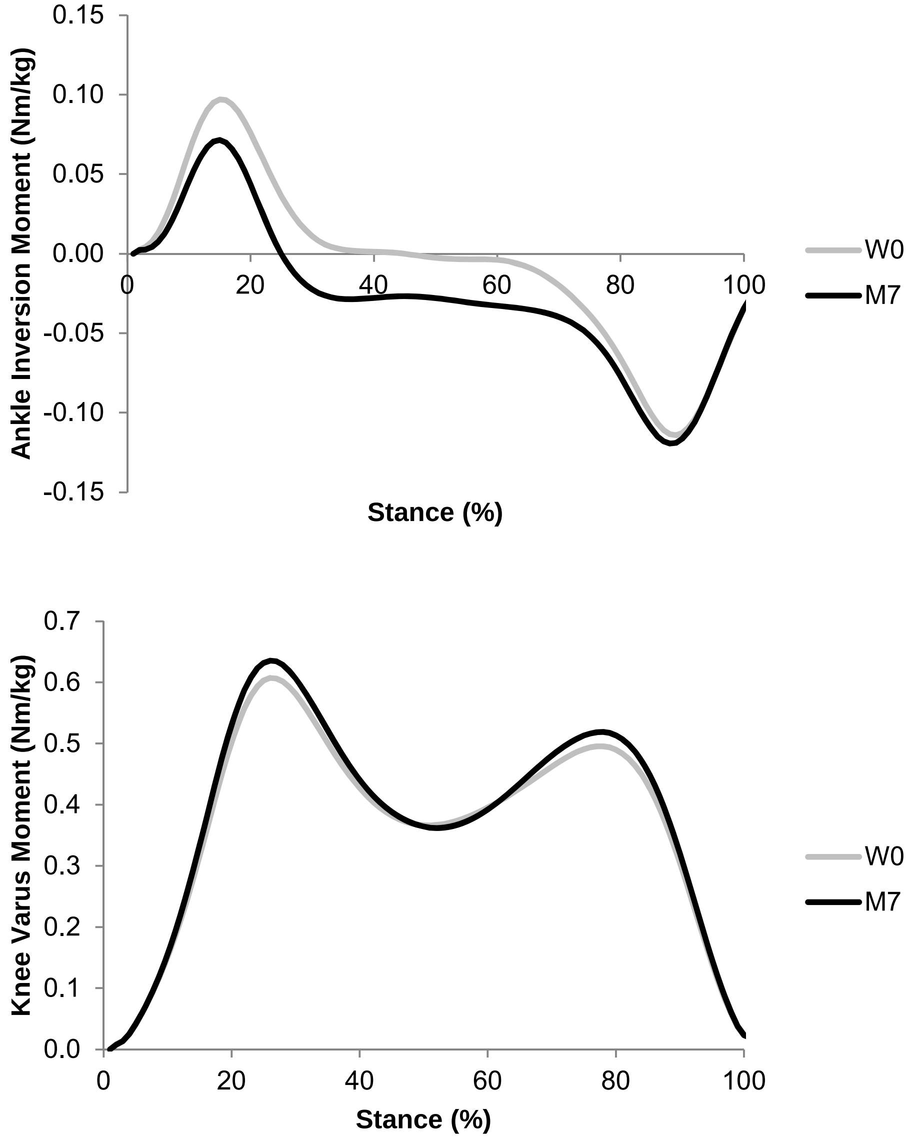

Figure 1. Ensemble curves for ankle inversion and external knee varus moments comparing W0 condition and the M7 wedge orthotic during stair descent.

4. Discussion

The purpose of this study was to utilize medial and lateral wedge orthotics to manipulate foot supination and provide greater knowledge of how foot alignment can affect joint moments up the kinetic chain. Abnormal foot biomechanics have been linked to lower extremity injuries (Pohl et al., 2009; Tiberio 1987; Bolga & Malone, 2004) that may be caused by a chain reaction from the STJ to the tibia and femur (Duval et al., 2010; Khamis & Yizhar, 2007; Souza et al., 2010; Mosely et al., 1996). Lateral wedge orthotics have been prescribed to reduce knee osteoarthritis pain by increasing STJ pronation with the goal of decreasing external knee varus moments (Bennell et al., 2011, Tuck et al., 2011; Hinman et al., 2009; Kerrigan et al., 2002). Medial wedges have been prescribed to decrease low back pain by reducing STJ overpronation with the goal of improving lumbopelvic alignment (Barwick et al., 2012; Castro-Mendez et al., 2013; Botte et al., 1981; Dananberg et al., 1999).

Our first hypothesis that medial wedge orthotics would reduce maximum ankle inversion moments was partially supported. The M7 wedge significantly reduced ankle inversion moments by 30% during stair ascent and by 26% during stair descent as compared to the W0 condition. In fact, the M7 wedge significantly reduced ankle inversion moments as compared to all other wedges tested during stair descent. There was evidence that the M7 wedge produced an overcorrection, since the maximum ankle eversion moment was increased by 16% during walking compared to the W0 condition. However, the M3 wedge was not enough of a correction to result in significantly different ankle inversion moments during walking, stair ascent, or stair descent. It would be of interest to test a 5° medial wedge to see if reduction in ankle inversion moments could be achieved without increases in ankle eversion moments. The L3 and L7 wedges did not produce any significant changes in ankle inversion or eversion moments.

Our second hypothesis that lateral wedge orthotics would reduce maximum external knee varus moments was not supported. The L3 and L7 wedges did not significantly change external knee varus moments as compared to the W0 condition during walking, stair ascent, or stair descent. These results are in disagreement with Kerrigan et al. (2002) and Hinman et al. (2012), who found that a 5° lateral wedge reduced external knee varus moments by 6% during walking. A likely explanation for the discrepancy in results is that the previous studies included participants with knee osteoarthritis, while the current study involved young healthy adults. There is also evidence that the L7 wedge might be an excessive angle, since the knee extension moments increased by 7% during stair ascent during the W0 condition. With no other increases or decreases in joint moments during stair ascent, the increase in knee extension moment with the L7 wedge seems to indicate a less efficient movement generation pattern.

Our third hypothesis that medial wedge orthotics would reduce maximum L5S1 compression forces also was not supported. The M3 and M7 wedges did not produce any significant changes in L5/S1 compression forces as compared to the W0 condition for walking, stair ascent, or stair ascent. In addition, L3 and L7 wedges did not produce any significant changes in the L5/S1 compression forces. With the simple model developed in this study, L5/S1 compression forces are affected by L5/S1 reaction forces, L5/S1 flexion/extension moments, and pelvic tilt. Considering the model, it is not surprising that the L5/S1 compression forces did not change since hip flexion and extension moments did not change with the medial or lateral wedges for walking or stair negotiation. The effect of wedge angle on L5/S1 loading requires further study as Castro-Mendez et al. (2013) saw a reduction in low back pain using custom made orthotics to control foot pronation.

A result that causes concern is that the M7 wedge significantly increased maximum external knee varus moments by 6% during walking, 5% during stair ascent, and 7% during stair descent as compared to the W0 condition. Increased external knee varus moments are associated with medial compression of the knee joint (Kerrigan et al., 2002). In addition, increased external knee varus moments are linked with increased risk of knee osteoarthritis progression (Miyazaki et al., 2002). Any benefits of reduced ankle inversion moments with the M7 wedge are likely outweighed by the potentially negative consequences at the knee joint. Figure 1 illustrates the compromise between ankle inversion moments and external knee varus moments with the M7 wedge. These results emphasize the importance of examining both the ankle and knee moments when evaluating orthotics. As mentioned previously, this may be a case of overcorrection with the M7 wedge and further study with a 5° medial wedge is warranted.

One limitation of this study was that the participants were young, healthy adults without low back pain or lower extremity injuries that would affect walking or stair negotiation. Instead of correcting a problem, the use of wedge orthotics in this study was to simulate abnormal foot alignment. Future research including participants with abnormal foot and/or knee alignment with symptoms such as knee osteoarthritis or low back pain is warranted. Another limitation of this study was that the subjects only wore the different orthotics for a short period of time. It is possible that wearing the orthotics for days or weeks may reveal longer term compensations. However, it is difficult to justify wearing a more extreme wedge orthotic such as the M7 or L7 for an extended period of time without initial short-term gait analysis. A final limitation was the simple L5/S1 compression model. Low back models with additional muscles may provide additional insight into the mechanisms of low back pain.

There are practical conclusions that can be drawn from this study. Use of a wedge orthotic such as the M7 reduced ankle inversion during stair negotiation. However, the same M7 wedge increased external knee varus moments during walking and stair negotiation. When considering the use of a wedge orthotic, an individual's foot alignment, the degree of wedge angle, and effects at the knee joint need to be considered.

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CHAPTER 4: GENERAL CONCLUSIONS

The major finding of the current study was that the M7 wedge significantly increased external knee varus moments by 6% during walking, 5% during stair ascent, and 7% during stair descent. An increase in knee varus moment is a risk factor for the development of knee osteoarthritis since it is associated with an increase in medial compression of the knee (Kerrigan et al., 2002; Miyazaki et al., 2002). Medial wedge orthotics are commonly used to correct for hyperpronation in order to stabilize the foot. Although ankle inversion moments were reduced during stair ascent descent with the M7 wedge, the negative effects on the knee joint likely outweigh the benefits.

Excessive foot pronation may be a risk factor in the development of low back pain, and patients with low back pain report a reduction in pain while wearing medial orthotics to control for the excessive pronation (Castro-Mendez et al., 2002; Rothbart et al., 1988). Due to the increase in external knee varus moments while wearing the M7 wedge, prescribing medial wedges to reduce pronation in low back pain patients should be done with caution. The M7 wedge could be an overcorrection and studying the effects of a 5° wedge would be a logical next step.

During gait there is a coupling relationship with the STJ, tibia, and femur. During initial foot contact, ground reaction forces cause the STJ to pronate (Castro-Mendez et al., 2012). Previous studies support that there is a chain reaction from STJ pronation that leads to internal rotation of the tibia and femur (Khamis et al., 2007; Duval et al., 2012; Tateuchi et al., 2011). Excessive pronation over time may lead to the development of low back pain due to changes in muscle recruitment patterns resulting from abnormal alignment of the pelvis (Himmelreich et al., 2008; Bird et al., 2003).

Results of Bird et al. (2003) showed a delay in muscle recruitment patterns in the erector spinae while wearing shoe orthotics. The current study focused on kinetic changes in the lower extremity and low back. Further study using EMG to measure spinal musculature recruitment patterns during walking and stair negotiation while wearing various shoe inserts may yield further insights into low back pain.

Definitions of excessive pronation vary as do the methods for measuring pronation. Rothbart & Estabrook (1988) define excessive pronation as greater than 6°, while Botte (1981) defines hyperpronation as greater than 10°. In the current study, L5/S1 compressive forces did not increase when wearing medial and lateral orthotics up to a 7° wedge angle. However, previous studies have shown improvement in low back pain patients from wearing medial wedge orthotics to control for hyperpronation (Castro-Mendez et al., 2012; Rothbart & Estabrook, 1988). Excessive pronation in one foot can create a leg length discrepancy that increases stress in the lumbar spine by inducing scoliosis to maintain postural balance (Botte 1981; Gurney 2002; Tateuchi et al., 2012). The current study looked at effects of manipulating foot alignment bilaterally with orthotics, and future research on unilateral orthotics may prove beneficial to examine the effects of leg length discrepancy.

In the current study, L5/S1 compression forces increased while wearing the M7 wedge orthotic as compared to the M3 during walking. A possible explanation of this could be that the M7 increased maximum hip extension moments compared to the M3 during walking. Another possible explanation of increased L5/S1 compression with M7 is that hip internal rotation was limited. Low back pain patients exhibit a reduction in hip internal rotation (Cibulka et al., 1998; Ellison et al., 1990). The current study only looked at changes in sagittal and frontal plane joint moments. Examining effects of medial and lateral wedge orthotics on transverse plane kinetics may provide greater insight on the effects foot pronation and supination have internal/external rotation moments at the hip and the low back. It is possible that increases in left/right rotation moments at the L5/S1 may be a factor in low back pain development.

The fact that the M7 wedge orthotic increased ankle eversion moments in walking and external knee varus moments in walking, stair ascent, and stair descent indicates that 7° is too much of a correction for foot hyperpronation. Clinicians should consider foot alignment and the effects on the knee joint when prescribing orthotics for those who have low back pain. The current study provides evidence that there is a chain reaction from the foot to at least the hip joint and assessment of foot alignment may be of value for patients who have knee osteoarthritis.

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