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The effects of medial and lateral wedges on iliotibial band strain during overground running

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The effects of medial and lateral wedges on iliotibial band strain during overground running

by

Evan M. Day

A thesis submitted to the graduate faculty in partial fulfillment of the requirements for the degree of

MASTER OF SCIENCE

Major: Kinesiology

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Iowa State University
Ames, Iowa
2015

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<td>ITB</td>
<td>Iliotibial Band</td>
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<tr>
<td>ITBS</td>
<td>Iliotibial Band Syndrome</td>
</tr>
<tr>
<td>LFE</td>
<td>Lateral Femoral Epicondyle</td>
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<tr>
<td>ASIS</td>
<td>Anterior Superior Iliac Spine</td>
</tr>
<tr>
<td>CFO</td>
<td>Custom Foot Orthotic</td>
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<tr>
<td>TFL</td>
<td>Tensor Fascia Latae</td>
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<tr>
<td>LW7</td>
<td>Lateral 7° Wedge</td>
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ABSTRACT

Background: Iliotibial band syndrome (ITBS) is the leading cause of lateral knee pain in runners. Previous research has theorized that higher ITB strain rate leads to ITBS development. Orthotics are commonly used to correct gait mechanics and may reduce ITB strain and strain rate. The purpose of this research was to investigate how wedge inserts and gender affect kinematics, kinetics, and ITB strain and strain rate during running. Methods: Thirty (15 male, 15 female) participants ran with lateral 7°, lateral 3°, no wedge, medial 3°, and medial 7° wedges. A motion capture system and force platform were used to collect kinematic and kinetic data. Joint angles and joint moments were calculated during the stance phase. ITB strain and strain rate were determined using a six degree of freedom musculoskeletal model. Findings: There were no significant differences for ITB strain or strain rate between wedge conditions or genders. The lateral 7° wedge resulted in significantly higher ankle eversion angles and lower ankle plantar flexion moments than no wedge. The medial 7° wedge resulted in significantly lower ankle eversion angles, higher hip internal rotation angles, lower ankle plantar flexion moments, lower ankle inversion moments, and higher external knee varus moments. Males had significantly higher knee valgus angles, knee internal rotation angles, ankle plantar flexion moments, ankle inversion moments, and knee extension moments. Interpretation: Results indicate that wedge inserts do not have a significant effect upon ITB strain and strain rate for healthy runners. While wedge orthotics may correct ankle/foot alignment problems, higher external knee varus moments with a medial 7° wedge are of concern.
CHAPTER 1

INTRODUCTION

Iliotibial band syndrome (ITBS) is the second most prevalent running injury behind patellofemoral pain syndrome and is the leading cause of lateral knee pain in runners (Taunton et al., 2002). Among all running injuries, ITBS accounts for 12% of cases (Fredericson et al., 2000). Onset of ITBS has been theorized to be caused by the ITB snapping across the LFE at 20-30° of knee flexion, known as the impingement zone (Orchard et al., 1996). An alternate theory is that ITBS results from compression of the ITB against the LFE in the impingement zone, as opposed to snapping of the ITB across the epicondyle, thus meaning ITBS is not a true friction syndrome (Fairclough et al., 2006). It remains unclear which theory is the true etiology of ITBS.

Multiple biomechanical factors can contribute to the development of ITBS. Messier et al. (1995) reported individuals with ITBS exhibited greater rearfoot motion throughout stance when compared to healthy runners. Contradicting results have been reported that individuals with ITBS do not exhibit a larger peak eversion angle (Ferber et al., 2010b; Miller et al., 2007; Noehren et al., 2007). Miller et al. (2007) reported that individuals with ITBS had greater foot inversion at heel strike, potentially leading to a larger rearfoot eversion excursion. However, it remains undetermined if excessive rearfoot motion is associated with development of ITBS.

Prospective and retrospective studies have reported that females who developed ITBS exhibited greater hip adduction and knee internal rotation than runners who did not (Ferber et al., 2010b; Noehren et al., 2007). On the contrary,
retrospective results report that runners with ITBS exhibit less hip adduction
throughout stance than healthy runners (Grau et al., 2011; Noehren et al., 2014).
This could be due to the fact that runners with ITBS have a tight ITB, reducing range
of motion. Individuals with ITBS have also been found to exhibit greater knee
flexion and knee internal rotation velocity (Miller et al., 2007). However,
contradicting results of no observed difference in peak knee flexion have been
reported (Ferber et al., 2010b). Adduction and internal rotation at the hip and
flexion, internal rotation, and varus at the knee cause an increase in ITB length due
to its position on the lateral aspect of the thigh. Greater range of motion of these
movements can be harmful because they increase strain in the ITB (Hamill et al.,
2008; Miller et al., 2007).

Strain and strain rate of the ITB has been reported to be higher in individuals
with ITBS, with strain rate being proposed to be a major factor in the development
of ITBS (Hamill et al., 2008; Miller et al., 2007). Previous studies that have assessed
ITB strain and strain rate have used a one degree of freedom hinge joint model of
the knee (Hamill et al., 2008; Meardon et al., 2012; Miller et al., 2007). High strain
rate can be caused by an increase in velocities of movements that will lengthen the
ITB, notably increased hip adduction velocity, knee internal rotation velocity, knee
flexion velocity, and tibial internal rotation velocity. It is possible that modification
of running form that decreases segment velocities could result in decreased ITB
strain rate during stance phase.

Orthotic use is commonly used in clinical settings to facilitate injury
rehabilitation and prevention (Taunton et al., 2002). However, the effect of orthotics
has drawn inconclusive results, with variable findings potentially due to dissimilar materials and alignments being used in studies. It is commonly accepted that medially aligned wedges or orthotics reduce rearfoot eversion, in turn reducing tibial internal rotation. Stackhouse et al. (2004) also found that medial wedges reduced knee flexion velocity. Decreases in tibial internal rotation and knee flexion velocity can lead to a decrease in ITB strain rate.

Joint moments at the ankle and knee are affected by orthotic use. Medially aligned wedges and custom foot orthotics often reduce ankle inversion moments (Lewinson et al., 2013a; Maclean et al., 2009; Nigg et al., 2003; Stackhouse et al., 2004), but increase knee external rotation moments and external knee varus moments, shifting the loading of the knee to the medial compartment (Boldt et al., 2013; Lewinson et al., 2013a; Maclean et al., 2009; Nigg et al., 2003). Use of lateral wedges has been shown to increase ankle inversion moments and external knee valgus moments, shifting the loading of the knee to the lateral compartment (Lewinson et al., 2013b).

An increase in external knee varus moments and external rotation moments result in greater strain on soft tissues that cross the lateral aspect of the knee and increases loading of the medial knee compartment (Boldt et al, 2013; Maclean et al., 2009; Mundermann et al., 2003). These results suggest that if kinetics play a role in the etiology of ITBS, then medial wedges could be harmful. However, it is unknown how kinetics effect ITBS development (Noehren et al., 2007).

The majority of ITB research has separately investigated males or females. This may be because females run with greater hip internal rotation, hip adduction,
and knee valgus (Chumanov et al., 2008; Ferber et al., 2003). Hip internal rotation and hip adduction both increase the length of the ITB; with increased hip adduction previously reported as a potential risk factor for ITBS development (Noehren et al., 2007). Chumanov et al. (2008) reported that gluteus maximus activity was greater in females than males when running. Because the ITB serves as a connection for the gluteus maximums, increased activity may lead to increased tension in the ITB. Due to females exhibiting gait mechanics that may influence ITB strain and we chose to analyze males and females separately.

No study to date has evaluated the effect of wedged shoe inserts on ITB strain. Existing literature indicates the use of medial wedge orthotics can alter kinematic patterns that could be beneficial in prevention and/or rehabilitation of ITBS. However, use of medial wedges may lead to greater knee external varus moments, increasing ITB strain (Boldt et al., 2013; Lewinson et al., 2013a; Maclean et al., 2009; Nigg et al., 2003). Our first hypothesis is that medial wedges will decrease ITB strain rate despite a potential increase in knee varus. Our second hypothesis is that lateral wedges will increase ITB strain rate due to influencing larger movements in the kinematic chain.
Thesis Format

This thesis is organized into Introduction, Review of the Literature, Manuscript, and General Conclusions. The manuscript will be submitted to *Clinical Biomechanics*. The primary author for this article is Evan Day. Dr. Jason Gillette contributed to the experimental design, data analysis, and preparation of the manuscript. Dr. Tim Derrick contributed to the experimental design and data analysis.
CHAPTER 2

REVIEW OF THE LITERATURE

Epidemiology and Anatomy

The iliotibial band (ITB) is a sheet of thick tendinous fascia on the lateral aspect of the thigh. Iliotibial band syndrome (ITBS) is a common injury among cyclists, runners, and military recruits. ITBS accounts for 12% of all running injuries (Fredericson et al., 2000). It is the second most common injury behind patellofemoral pain syndrome (PFPS), and is the leading cause of lateral knee pain in runners (Taunton et al., 2002). Pain from ITBS has a slow onset, progressively increasing during activity.

The anatomy and biomechanical function of the ITB has been extensively studied. At the proximal portion of the leg, the ITB has superficial and deep connections. The deep connection attaches to the hip joint capsule (Birnbaum et al., 2004). The superficial connections of the ITB attach to the gluteus maximus and tensor fascia latae. Proximally, the ITB is anchored to the anterior superior iliac spine (Birnbaum et al., 2004). Due to its connections to hip abductor muscles, the proximal portion of the ITB functions as a hip stabilizer, providing resistance to hip adduction and internal rotation (Fredericson et al., 2000). The ITB lies on the lateral portion of the thigh along the linea aspera (Falvey et al., 2010). Distally, the ITB is anchored to the lateral femoral condyle and crosses the knee joint, where it inserts onto Gerdy’s Tubercle on the tibia. The ITB also has distal attachments to the head of the fibula, lateral intermuscular septum, and the transverse and longitudinal
retinacula of the patella (Birnbaum et al., 2004). These attachments result in the ITB functioning to resist knee external varus moments and internal rotation of the tibia (Hamill et al., 2008).

**Etiology**

Iliotibial band syndrome causes pain at the lateral portion of the knee. There are two theories regarding the development of ITBS: friction and compression. The first states that as the knee moves through flexion, the ITB snaps posteriorly across the lateral femoral epicondyle (LFE). Moving through extension, the ITB passes back anteriorly across the LFE. This movement across the LFE occurs at 20-30° of knee flexion and causes an increase in friction, resulting in lateral knee pain. This movement range is referred to as the impingement zone (Orchard et al., 1996). When the ITB is in the impingement zone the gluteus maximus and tensor fascia latae contract eccentrically to slow the leg down, creating tension in the ITB (Orchard et al., 1996). It is unknown if the duration of time in the impingement zone has an effect on the development of ITBS. Hamill et al. (2008) stated that being able to model the time in the impingement zone could allow for a better understanding of the intrinsic factors leading to ITBS.

The second theory states that lateral knee pain from ITBS can also be a result of compression of the layer of fat between the ITB and LFE (Fairclough et al., 2006). At approximately 30° of knee flexion the ITB is drawn medially towards the epicondyle, resulting in compression, as opposed to a friction buildup. The adipose layer between the ITB and LFE is highly vascular and contains pacinian corpuscles,
resulting in the onset of pain. It is yet to be concluded which theory is more accepted as the cause of pain in ITBS.

Development of ITBS can also occur at the hip (Ilizaliturri et al., 2006). Snapping hip is a result of the ITB moving anteriorly and posteriorly across the greater trochanter. The bursa of the greater trochanter becomes inflamed, causing pain. However, most cases of ITB injury result in development of ITBS at the knee and not snapping hip.

There are two clinical tests to determine if an individual has ITBS: the Noble’s test and the Ober’s test. The Noble’s test is used to elicit pain in the ITB at the spot of impingement (Noble, 1980). The patient lies supine, flexes their knee to 90°, and pressure is applied proximal to the LFE. The leg is then extended towards neutral anatomical position. If the individual has ITBS, they should feel pain at relatively the same degree of knee flexion as when they are running.

The Ober’s test is used to assess length and flexibility of the ITB (Ober, 1936). The patient lies on their uninjured side and the examiner then guides the injured leg through a range of movement. With the pelvis stabilized, the injured leg is abducted, extended, and then adducted towards the examination table. If the leg remains slightly abducted, then the ITB is exhibiting tightness. Traditionally, the injured leg is set to 90° knee flexion during the test. Gajdosik et al. (2003) proposed that performing the Ober’s test at 0° flexion, as opposed to 90° knee flexion, results in greater hip adduction results. While not statistically significant, Miller et al. (2007) found that runners with ITBS exhibited slightly less range of motion in the Ober’s test than healthy controls, proposing a link between a tight ITB and ITBS.
**Treatment and Rehabilitation**

Treatment and rehabilitation protocols for ITBS can be lengthy. There are both operative and non-operative options to alleviate ITBS. Fredericson & Wolf (2005) developed a commonly used protocol to aid in ITBS rehabilitation. The protocol consists of three phases: acute, subacute, and recovery/strengthening. In the acute phase, individuals modify the level of activity to reduce inflammation. This is done by monitoring extrinsic factors such as mileage or running terrain. The use of ice and massage is recommended to reduce inflammation. If no improvement occurs, corticosteroid injections can be implemented to alleviate pain. Gunter & Schwellnus (2004) reported a decrease in pain within two weeks using corticosteroid injections.

During the subacute phase, the focus is on increasing flexibility of the ITB. Increased flexibility allows for more effective strength training. Loosening of the gluteus maximus and tensor fascia latae can decrease strain in the ITB, relieving compression over the lateral portion of the knee.

The last phase, recovery and strengthening, is not to be started until range of motion has been improved. The gluteus medius is the muscle targeted during this phase. Strengthening hip abductors can decrease the amount of hip adduction during gait. Larger degrees of hip adduction have been identified as a potential causative factor of ITBS (Ferber et al., 2010b; Noehren et al., 2007). By reducing hip adduction, there is potential for ITB strain to decrease. Recommended exercises to strengthen the gluteus medius are single leg hip abduction lifts, pelvic drops, and multi-planar lunges. Falvey et al. (2010) support the idea that the tensor fascia latae
and gluteus maximus should be targeted for treatment as opposed to treatment of local symptoms.

If recovery and strengthening exercises do not work to relieve ITBS, multiple surgical methods can be implemented. The mesh technique involves positioning the knee to 30° flexion, cutting a 3cm incision over the LFE, and cutting multiple 2mm slits in the ITB (Sangkaew, 2007). Martens et al. (1989) details a similar procedure, but with removal of a triangular portion of the ITB directly over the LFE. Both methods focus on relieving tension and compression of the ITB over the LFE. Z-lengthening is the last commonly performed surgical method. Richards et al. (2003) describes Z-lengthening as making a 5cm incision over the ITB so that it can be isolated from other soft tissues. Two transverse incisions are then made in the ITB, connected by a longitudinal incision. Multiple small horizontal stitches are then applied for repair. The goal of Z-lengthening is to relieve tension in the ITB. While surgical methods are effective, individuals with ITBS are recommended to go through a lengthy non-operative protocol before deciding upon operative treatment.

Various gait manipulation methods have potential to alleviate ITBS. A case study reported by Allen (2014) found that a female with ITBS was able to become pain free after a 4-week intervention by increasing her step rate by 5%. Increasing step rate changed the subject from a rearfoot to midfoot striker, resulting in decreased hip adduction, decreased hip internal rotation angles, and reduced joint loading at the knee. However, the subject also underwent a dynamic hip abductor strengthening regimen, so it is unknown if the improvement in relieving ITBS pain was strictly from the increased step rate or from the strengthening protocol.
Manipulation of step width can also be utilized for treatment and prevention of ITBS. Meardon et al. (2012) found that a narrow step width increases ITB strain and strain rate during running. Increased strain rate is hypothesized to be a causative factor of ITBS (Hamill et al., 2008). A wider step width can decrease ITB strain rate and may be useful in prevention and treatment of ITBS.

**Intrinsic Factors Contributing to ITBS**

Intrinsic factors predisposing ITBS can be muscular deficits, kinematic patterns, and anatomical abnormalities. Muscular deficits of the hip abductors are commonly associated with ITBS. Noble et al. (1980) and Fredericson et al. (2000) proposed that weak hip abductors (gluteus medius, gluteus minimus, tensor fasciae latae) and a tight ITB lead to the development of ITBS. However, in a retrospective study by Grau et al. (2008), ITBS individuals did not exhibit a difference in static or dynamic strength measurements of the hip abductors when compared to healthy controls. Noehren et al. (2014) reported similar results that men with ITBS did not have weaker hip abductors when compared to healthy controls, but they did have weaker hip external rotators. Individuals with ITBS did not exhibit differences in strength between the injured and non-injured leg either. Fredericson et al. (2000) reported distance runners with ITBS had weaker hip abductors on the injured side and became pain free after a six-week strengthening protocol. These confounding results could be due to varying equipment, strength tests, and the degree of knee flexion during the tests.
Other potential intrinsic causes of ITBS are oversized femoral condyles, a medially aligned tibia, varying leg length, and genu varum of the legs. These factors lead to a potential increase in ITB strain throughout stance phase.

**Kinematics of ITBS**

Many studies have focused on attempting to identify kinematic patterns that can act as a predisposition to ITBS. Results have been inconclusive however, potentially due to some studies being retrospective and some studies being prospective in nature.

Although ITBS can be alleviated with proper treatment and rehabilitation techniques, it has been found that healed females still exhibited different kinematic and kinetic patterns than healthy controls (Ferber et al., 2010b). Females that previously had ITBS still exhibited greater peak knee internal rotation and hip adduction when compared to healthy controls. Greater peak knee internal rotation and hip adduction have been prospectively identified as potential kinematic factors leading to the development of ITBS (Noehren et al., 2007). These results suggest that despite the subject pool being healthy at the time of data collection, they are at risk of re-developing ITBS unless gait patterns are changed.

Noehren et al. (2007) performed a prospective study of 200 female runners over the span of two years. Subjects who developed ITBS exhibited greater hip adduction and knee internal rotation. Increased hip adduction and knee internal rotation are associated with greater ITB strain and compression against the LFE. Greater knee internal rotation in the ITBS group was determined to be caused by greater external femoral rotation, while internal tibial rotation was lower than the
controls. Bauer & Duke (2012) also reported a lack of excessive internal tibial rotation in individuals with ITBS. Excessive external femoral rotation can be caused by muscular imbalances at the hip. Further study investigating hip strengthening protocols and femoral rotation is needed.

Rearfoot eversion is coupled with internal tibial rotation (Inman, 1976; McClay & Manal, 1997). Increased internal tibial rotation is expected to increase ITB strain. Messier et al. (1995) reported runners with ITBS exhibited greater rearfoot motion than the controls. Miller et al. (2007) found that ITBS individuals had greater rearfoot inversion at heel strike than the healthy controls. A larger degree of inversion can result in greater rearfoot eversion excursion. Large excursion values can affect joint coupling patterns in a negative manner, resulting in mistiming of segment movement. Due to contrasting results, it is unknown if rearfoot motion is a causative factor to development of ITBS.

Grau et al. (2011) reported confounding results in hip motion compared to Noehren et al. (2007). Subjects with ITBS exhibited less hip adduction during stance phase than healthy controls. These contrasting results could be due to differentiating methods. Noehren et al. (2007) only examined females, whereas Grau et al. (2011) studied males and females. Females tend to run with greater hip adduction, hip internal rotation, and knee valgus when compared to men (Chumanov et al., 2008; Ferber et al., 2003). Increased hip adduction and hip internal rotation in women are consistent with potential kinematic patterns that can lead to ITBS, thus studying men and women can lead to different results. Grau et al. (2011) was also a retrospective study, so the individuals had already developed
ITBS. ITBS is associated with tight hip abductors and a tight ITB, which would resist hip adduction. A tight gluteus maximus can also create tension and restrict movement of the ITB. This tension resists the range of motion of hip adduction during gait, supporting the findings by Grau et al. (2011) that individuals with ITBS exhibited less hip adduction than healthy individuals. In support of no differences in hip motion in ITBS individuals, Noehren et al. (2014) found that men with ITBS did not exhibit greater hip adduction but had greater hip internal rotation, knee varus, and an anatomically shorter ITB.

The effect of fatigue on running gait has been extensively studied. ITBS has a slow onset of pain during a run, thus changes in kinematics due to ITBS may not be prevalent at the beginning of exercise. Miller et al. (2007) found that at the end of an exhaustive run on a treadmill, ITBS individuals exhibited greater knee flexion at heel strike, ITB strain throughout stance phase, foot adduction during stance, foot inversion at contact, and maximum knee internal rotation velocity, when compared to runners without ITBS. An increase in knee flexion is a potential injury mechanism for ITBS because the ITB will be in the impingement zone at the time of heel strike. Higher knee internal rotation velocity results in a greater ITB strain rate, which can serve as a primary cause of ITBS (Hamill et al., 2008).

Further up the kinematic chain, Foch & Milner (2014) examined how trunk movement in the frontal plane could potentially affect development of ITBS. They hypothesized that weak lateral core muscles will result in greater peak contralateral pelvic drop, trunk contralateral flexion, hip adduction, and external knee varus moment. Greater degrees of hip adduction have previously been hypothesized to be
a causative factor of ITBS (Noehren et al. 2007). However, Foch & Milner (2014) found that there were no differences in peak frontal plane movement of the pelvis, trunk, and hip angles, and no difference in external knee varus moment, suggesting that pelvic and trunk motion may not affect development of ITBS. Ford et al. (2013) investigated the relationship between hip strength and trunk motion. They determined that hip extensor strength is inversely correlated with thorax axial rotation range of motion, and hip abductor strength inversely correlated with pelvic obliquity range of motion. Muscle groups assessed in both studies were different, but their findings regarding trunk motion contradict each other. Therefore, it remains unknown if excessive trunk motion affects development of ITBS.

Measures of joint coupling have the potential to identify abnormal phasic patterns of movement that can lead to onset of injury. Miller et al. (2008) examined coordination variability between individuals with ITBS and healthy controls during an exhaustive run. Data were analyzed at the start, middle, and end of the run. They found that the ITBS group was less variable in couplings between thigh adduction/abduction-foot inversion/eversion, and thigh adduction/abduction-tibia internal/external rotation. The ITBS group was more variable in knee flexion/extension-foot adduction/abduction than the healthy controls during stance phase. Miller et al. (2008) proposed that low variability of thigh adduction/abduction coupling could be indicative of dysfunctional hip abductors, a potential pre-determinant cause of ITBS. However, low variability can lead to a smaller range motion that allows for pain-free running (Hamill et al., 1999). In contrast, low variability of movement, notably coupling of thigh
abduction/adduction, can result in segmental motion patterns that result in continuous stress of the ITB (Miller et al., 2008). Thus, investigation of coupling variability and injury prevalence requires further investigation.

**Kinetics of ITBS**

It is unknown how loading of the ITB affects development of ITBS. Noehren et al. (2007) found no differences in kinetic patterns between those that developed ITBS and healthy individuals. However, Ferber et al. (2010b) found that healed females who previously had ITBS exhibited a greater rearfoot invertor moment. Rearfoot inversion moment is associated with control of foot pronation, and can thus affect internal tibial rotation. Further examination of kinetic patterns is needed, notably of variables that can affect strain of passive tissues at the knee and hip.

**Strain and Strain Rate of the ITB**

Strain of the ITB is defined as the percent change in length of the ITB during stance phase. Averaged modeled ITB strain ranges from 6-9% (Hamill et al., 2008), with reports of failure from cadaver studies averaging 13% (Birnbaum et al., 2004). Hamill et al. (2008) prospectively examined ITB strain and strain rate in runners that developed ITBS over a two-year period. Maximum strain values were assessed at touchdown and maximum knee flexion. Though not statistically significant, ITBS individuals exhibited greater strain throughout stance phase when compared with the matched limb of the control group. Strain values were 7.3% versus 6.4% at touchdown, and 9.0% versus 7.3% at maximum knee flexion. Although not statistically significant, within subjects’ comparison of the injured and non-injured leg in the ITBS individuals yielded higher strain values. ITBS individuals exhibited
strain of 7.3% versus 5.5% at touchdown, and 9.0% versus 7.7% at maximum knee flexion. Miller et al. (2007) reported similar results. ITBS individuals exhibited an average peak strain of 8.5% at the beginning of the run versus 7.5% in the controls. Strain at the end of the run was similar, 8.4% for ITBS versus 7.5% for controls. ITB strain was higher in ITBS individuals for the final 90% of stance.

Hamill et al. (2008) suggests that while strain was higher in ITBS affected limbs it is strain rate that should be looked at as an injury mechanism, not peak strain. Strain rate is defined as the change in strain over the change in time. Individuals with ITBS exhibited significantly higher strain rate (25.1%/s versus 12.4%/s) when compared to the control group. Implications can be made that if strain rate can be controlled, incidence of ITBS can be reduced. Manipulating velocities of various kinematic patterns, notably transverse plane rotations and knee flexion, can change strain rate. These manipulations can potentially occur by altering foot movement.

**Extrinsic Factors Contributing to ITBS**

The most common extrinsic factor leading to the development of ITBS is a quick onset of workload (Messier et al., 1995). Increasing mileage, whether running or on the bicycle, at too fast of a rate creates a stimulus that the musculoskeletal system cannot adjust to and thus injury occurs. Downhill running can also lead to development of ITBS (Orchard et al., 1996). When running downhill, the knee is more extended at heel strike, causing impact to potentially occur while the ITB is in the impingement zone. Landing with the ITB in the impingement zone leads to an increase in pain. To avoid heel strike occurring while the ITB is in the impingement
zone, sprinting has been determined to help individuals recover from ITBS because an individual lands with more knee flexion and thus the ITB has moved past the impingement zone by the time heel strike occurs (Orchard et al., 1996). However, this is not supported by researchers who found that ITBS patients had increased knee flexion at heel strike while running (Miller et al., 2007; Bauer & Duke, 2012).

**Orthotics**

Use of orthotics has been recommended for treating running injuries and is commonly used in clinical settings (Taunton et al., 2002). Orthotics are traditionally prescribed to promote skeletal alignment to fix gait abnormalities. However, it has been proposed that as opposed to skeletal realignment, orthotics support the preferred movement of the foot and thus reduce lower extremity muscle activity. Lower muscle activity would then result in less fatigue and energy expenditure (Nigg et al. 1999; Boldt et al, 2013). It has been extensively studied how medial and lateral wedges and various models of custom foot orthotics (CFO’s) affect running gait, but results have been inconclusive. Varying results have been due to the use of different materials, degree of wedge, half or full foot alignment, or using a CFO.

**Kinematic Changes With Orthotics**

Orthotic use manipulates foot kinematics, which is expected to result in compensatory changes at the knee and hip. This linkage is referred to as the closed kinematic chain. Orthotic use affects foot pronation (a combination of eversion, adduction, and dorsiflexion), which directly affects rotation of the tibia due to the functionality of the ankle, notably the subtalar joint. Excessive foot eversion coupled to an increase in internal rotation of the tibia has been reported in various studies
(McClay & Manal, 1997; Inman, 1976). Moving further up the kinematic chain, greater excursion of internal rotation of the tibia can lead to an increase in internal rotation of the knee.

Reports of kinematic adjustments have varied across studies due to different data collection methods. The most commonly reported differences are in rearfoot eversion and internal tibial rotation. Eng & Pierrynowski (1994) were able to reduce range of motion at the talocrural/subtalar joint by 1-3° in the frontal and transverse plane during walking using a 3° soft orthotic. These differences would be potentially larger during running due to greater impact force and segment velocities.

Nester et al. (2003) reported using a 10° medial wedge decreased rearfoot pronation, and a 10° lateral wedge increased pronation during running. Mundermann et al. (2003) found similar results using a 6mm medial post, a custom molded neutral orthotic, and a custom molded 6mm medial posted orthotic. The medial post reduced rearfoot eversion, eversion velocity, internal tibial rotation, and internal tibial rotation velocity in the first half of stance. The molded (full foot) condition affected kinematics throughout the entirety of stance phase, as opposed to the first half. Vertical impact peak and loading rate were also reduced in the molded condition. Maclean et al. (2006) had similar findings using custom foot orthotics (CFOs) in healthy individuals. CFOs produced a decrease in rearfoot eversion and rearfoot eversion velocity, supporting the findings of other studies.

As opposed to using standard external markers, Stacoff et al. (2000) surgically implanted a bone pin into the lateral malleolus and proximal tibia of
participants and attached markers to the pin to analyze lower extremity kinematics during running. Subjects ran with a rearfoot 20° wedge and a midfoot 20° wedge. Results showed a decrease in internal tibial rotation, but no change in rearfoot eversion. However, rearfoot eversion as they reported is technically calcaneal eversion, not shoe eversion. A significant difference in eversion was measured between the shoe and calcaneus.

Varying anatomical alignments or current injuries have the potential to affect the effect of orthotics. Boldt et al. (2013) examined females with and without patellofemoral pain syndrome (PFPS) running with a full foot 6° medial wedge. Subjects were further divided into groups based upon static calcaneal angle. Medial wedges had the same effect on injured and healthy individual’s running gait. Calcaneal angle did not change the effect of orthotics. The medial wedge did not result in a decrease in internal tibial rotation, contrary to reports from other studies. However, a small but significant decrease in hip adduction excursion was observed in the medially wedged conditions.

Midsole density of running shoes can change the effect of orthotics. Maclean et al. (2009) assessed the effects of three varying midsole densities with and without CFO intervention on lower extremity dynamics. The CFOs used had a 5° medial wedge alignment. In all shoe conditions, CFOs resulted in reduced rearfoot eversion, rearfoot eversion velocity, calcaneal eversion, and internal tibial rotation. Across shoe conditions, harder shoes resulted in a decreased rearfoot eversion velocity, but greater internal rotation of the tibia.
Center of pressure (COP) throughout stance phase can be manipulated using orthotics. Nigg et al. (2003) used 4.5mm half and full foot posted medial and lateral wedges to examine COP alterations during running. Despite hypothesizing that the medial wedges would move the COP laterally, and the lateral wedges would move the COP medially, the opposite was observed. The full foot lateral condition was the only condition to exhibit a difference, pushing the COP laterally. This compensation could occur due to the body fighting the medial push of the orthotic.

Despite most orthotics typically having a rearfoot alignment, they still produce changes in gait in forefoot strikers (Stackhouse et al., 2004). A 6° medial wedge reduced rearfoot eversion and knee flexion velocity by 20% in two-thirds of subjects, both rearfoot and forefoot runners. Decreases in internal rotation and valgus angle of the knee occurred for both forefoot and rearfoot strikers.

Rearfoot motion and internal tibial rotation are the two main kinematic variables controlled by medial orthotics. Maclean et al. (2006) determined that CFO intervention is most effective in the first 50-60% of stance phase. Hamill et al. (2008) determined that peak ITB strain occurs at 35-40% of stance. This overlap opens up potential that controlling kinematics by means of orthotics can manipulate ITB strain.

Larger degrees of internal tibial rotation can lead to an increase in ITB strain due to the ITB inserting on Gerdy’s Tubercle on the lateral proximal portion of the tibia (Miller et al., 2007). The decrease in knee flexion velocity with a 6° medial wedge found by Stackhouse et al. (2004) may decrease the strain rate of soft tissue that crosses the knee joint, including the iliotibial band.
Noehren et al. (2007) found that individuals who developed ITBS exhibited greater internal rotation at the knee, which was associated with excessive external rotation of the femur as opposed to internal rotation of the tibia. However, the ITBS subjects with the greatest degree of rearfoot eversion exhibited the greatest degree of internal rotation of the tibia. This shows that a distal mechanism could be involved in developing ITBS, and thus orthotic control could be beneficial for prevention or rehabilitation.

**Kinetic Changes with Orthotics**

The most common kinetic adjustments with the use of orthotics are changes in ankle inversion moment and external knee varus/valgus moment. Stackhouse et al. (2004) reported a 24% reduction in ankle inversion moment and 33% reduction in inversion work using a 6° medial wedge. Lewinson et al. (2013a) used lateral and medial wedges of 0, 3, 6, and 9mm postings and reported a decrease in external knee varus moment and angular impulses and larger ankle inversion moments using lateral wedges. Lewinson et al. (2013b) examined subjects running in 6mm medial and 6mm lateral wedges. External knee varus impulse was 19.2% lower when running in the lateral wedge than compared to the medial wedge.

Mundermann et al. (2003) using a 6mm medial posting and 6mm custom molded orthotics reported a decrease in ankle inversion moment, but an increase in knee external rotation and external knee varus moment. Nigg et al. (2003) using full and half foot medial and lateral wedges with 4.5mm posting also found a decrease in ankle inversion moment with the medial wedge, but an increase in knee external rotation moment. Boldt et al. (2013) found that females with and without PFPS both
exhibited an increase in external knee varus moment using a 6° medial wedge. Maclean et al. (2009) found that CFO’s decreased ankle inversion moment, vertical impact peak, and loading rate. However, CFO’s were found to increase the knee external rotation impulse. Overall, medially aligned orthotics results in a decrease in ankle inversion moments and external knee valgus moments, but potentially result in an increased knee external rotation moments and external knee varus moments.

It is currently unknown how kinetic variables affect the iliotibial band (Noehren et al., 2007). While medial wedges have potential to control kinematics that can be predisposing factors to developing ITBS, they also create a larger external knee varus moment, and thus a lateral wedge can be beneficial if larger knee frontal plane joint moments contribute to the development of ITBS.

The shift in loading to the medial knee compartment using medial wedges can potentially increase strain in lateral knee tissue, notably the iliotibial band (Boldt et al., 2013). The increase in knee external rotation angular moment impulse found by Maclean et al. (2009) using a 5° medial aligned orthotic can cause an increased strain on passive tissues resisting internal tibial rotation. Mundermann et al. (2003) also hypothesized that a greater external knee rotation moment from medial wedges could result in an increase in soft tissue strain in the knee.

Results from the aforementioned studies imply that medial wedges can be utilized to control kinematic variables that potentially cause ITBS. However, if kinetic variables play a role in ITBS etiology, medial wedges can be harmful if they
increase strain on lateral knee tissues. Further investigation is needed as to
determine how medial and lateral wedges affect ITB strain during running.

References


CHAPTER 3

THE EFFECTS OF MEDIAL AND LATERAL WEDGES ON ILIOTIBIAL BAND STRAIN DURING OVERGROUND RUNNING

A paper to be submitted to Clinical Biomechanics

Evan M. Day, Jason C. Gillette, Timothy R. Derrick

Abstract

Background: Iliotibial band syndrome (ITBS) is the leading cause of lateral knee pain in runners. Previous research has theorized that higher ITB strain rate leads to ITBS development. Orthotics are commonly used to correct gait mechanics and may reduce ITB strain and strain rate. The purpose of this research was to investigate how wedge inserts and gender affect kinematics, kinetics, and ITB strain and strain rate during running.

Methods: Thirty (15 male, 15 female) participants ran with lateral 7°, lateral 3°, no wedge, medial 3°, and medial 7° wedges. A motion capture system and force platform were used to collect kinematic and kinetic data. Joint angles and joint moments were calculated during the stance phase. ITB strain and strain rate were determined using a six degree of freedom musculoskeletal model.

Findings: There were no significant differences for ITB strain or strain rate between wedge conditions or genders. The lateral 7° wedge resulted in significantly higher ankle eversion angles and lower ankle plantar flexion moments than no wedge. The medial 7° wedge resulted in significantly lower ankle eversion angles, higher hip internal rotation angles, lower ankle plantar flexion moments, lower ankle inversion moments, and higher external knee varus moments. Males had significantly higher
knee valgus angles, knee internal rotation angles, ankle plantar flexion moments, ankle inversion moments, and knee extension moments.

Interpretation: Results indicate that wedge inserts do not have a significant effect upon ITB strain and strain rate for healthy runners. While wedge orthotics may correct ankle/foot alignment problems, higher external knee varus moments with a medial 7° wedge are of concern.

Introduction

Iliotibial band syndrome (ITBS) is the second most prevalent running injury behind patellofemoral pain syndrome and is the leading cause of lateral knee pain in runners (Taunton et al., 2002). Among all running injuries, ITBS accounts for 12% of cases (Fredericson et al., 2000). Onset of ITBS has been theorized to be caused by the ITB snapping across the LFE at 20-30° of knee flexion, known as the impingement zone (Orchard et al., 1996). An alternate theory is that ITBS results from compression of the ITB against the LFE in the impingement zone, as opposed to snapping of the ITB across the epicondyle, thus meaning ITBS is not a true friction syndrome (Fairclough et al., 2006). It remains unclear which theory is the true etiology of ITBS.

Multiple biomechanical factors can contribute to the development of ITBS. Messier et al. (1995) reported individuals with ITBS exhibited greater rearfoot motion throughout stance when compared to healthy runners. Contradicting results have been reported that individuals with ITBS do not exhibit a larger peak eversion angle (Ferber et al., 2010b; Miller et al., 2007; Noehren et al., 2007). Miller et al. (2007) reported that individuals with ITBS had greater foot inversion at heel strike,
potentially leading to a larger rearfoot eversion excursion. However, it remains undetermined if excessive rearfoot motion is associated with development of ITBS.

Prospective and retrospective studies have reported that females who developed ITBS exhibited greater hip adduction and knee internal rotation than runners who did not (Ferber et al., 2010b; Noehren et al., 2007). On the contrary, retrospective results report that runners with ITBS exhibit less hip adduction throughout stance than healthy runners (Grau et al., 2011; Noehren et al., 2014). This could be due to the fact that runners with ITBS have a tight ITB, reducing range of motion. Individuals with ITBS have also been found to exhibit greater knee flexion and knee internal rotation velocity (Miller et al., 2007). However, contradicting results of no observed difference in peak knee flexion have been reported (Ferber et al., 2010b). Adduction and internal rotation at the hip and flexion, internal rotation, and varus at the knee cause an increase in ITB length due to its position on the lateral aspect of the thigh. Greater range of motion of these movements can be harmful because they increase strain in the ITB (Hamill et al., 2008; Miller et al., 2007).

Strain and strain rate of the ITB has been reported to be higher in individuals with ITBS, with strain rate being proposed to be a major factor in the development of ITBS (Hamill et al., 2008; Miller et al., 2007). Previous studies that have assessed ITB strain and strain rate have used a one degree of freedom hinge joint model of the knee (Hamill et al., 2008; Meardon et al., 2012; Miller et al., 2007). High strain rate can be caused by an increase in velocities of movements that will lengthen the ITB, notably increased hip adduction velocity, knee internal rotation velocity, knee
flexion velocity, and tibial internal rotation velocity. It is possible that modification of running form that decreases segment velocities could result in decreased ITB strain rate during stance phase.

Orthotic use is commonly used in clinical settings to facilitate injury rehabilitation and prevention (Taunton et al., 2002). However, the effect of orthotics has drawn inconclusive results, with variable findings potentially due to dissimilar materials and alignments being used in studies. It is commonly accepted that medially aligned wedges or orthotics reduce rearfoot eversion, in turn reducing tibial internal rotation. Stackhouse et al. (2004) also found that medial wedges reduced knee flexion velocity. Decreases in tibial internal rotation and knee flexion velocity can lead to a decrease in ITB strain rate.

Joint moments at the ankle and knee are affected by orthotic use. Medially aligned wedges and custom foot orthotics often reduce ankle inversion moments (Lewinson et al., 2013a; Maclean et al., 2009; Nigg et al., 2003; Stackhouse et al., 2004), but increase knee external rotation moments and external knee varus moments, shifting the loading of the knee to the medial compartment (Boldt et al., 2013; Lewinson et al., 2013a; Maclean et al., 2009; Nigg et al., 2003). Use of lateral wedges has been shown to increase ankle inversion moments and external knee valgus moments, shifting the loading of the knee to the lateral compartment (Lewinson et al., 2013b).

An increase in external knee varus moments and external rotation moments result in greater strain on soft tissues that cross the lateral aspect of the knee and increases loading of the medial knee compartment (Boldt et al, 2013; Maclean et al.,
2009; Mundermann et al., 2003). These results suggest that if kinetics play a role in the etiology of ITBS, then medial wedges could be harmful. However, it is unknown how kinetics effect ITBS development (Noehren et al., 2007).

The majority of ITB research has separately investigated males or females. This may be because females run with greater hip internal rotation, hip adduction, and knee valgus (Chumanov et al., 2008; Ferber et al., 2003). Hip internal rotation and hip adduction both increase the length of the ITB; with increased hip adduction previously reported as a potential risk factor for ITBS development (Noehren et al., 2007). Chumanov et al. (2008) reported that gluteus maximus activity was greater in females than males when running. Because the ITB serves as a connection for the gluteus maximums, increased activity may lead to increased tension in the ITB. Due to females exhibiting gait mechanics that may influence ITB strain and we chose to analyze males and females separately.

No study to date has evaluated the effect of wedged shoe inserts on ITB strain. Existing literature indicates the use of medial wedge orthotics can alter kinematic patterns that could be beneficial in prevention and/or rehabilitation of ITBS. However, use of medial wedges may lead to greater knee external varus moments, increasing ITB strain (Boldt et al., 2013; Lewinson et al., 2013a; Maclean et al., 2009; Nigg et al., 2003). Our first hypothesis is that medial wedges will decrease ITB strain rate despite a potential increase in knee varus. Our second hypothesis is that lateral wedges will increase ITB strain rate due to influencing larger movements in the kinematic chain.
Methods

Participants

Thirty (15 male and 15 female) recreational and competitive runners were recruited for this study. Subjects were recruited from the university running club, undergraduate biomechanics course, and by flyers hung in the department building. Exclusion criteria for subjects included if they were currently running less than 15 miles per week, suffered a lower extremity injury in the past 3 months, underwent any lower extremity surgeries in the past year, currently used orthotics, or were pregnant. Participants completed informed consent documents before data collection commenced. After data collection, subjects completed a questionnaire asking for their age, body mass, height, weekly mileage, 5000 m run personal best, shoe size, type of runner (recreational vs. competitive), and a general lower extremity injury history. This study was approved by the Institutional Review Board at Iowa State University.

Table 2.1 Participant characteristics. Values are mean ± standard deviation.

<table>
<thead>
<tr>
<th>Gender</th>
<th>Age (years)</th>
<th>Mass (kg)</th>
<th>Height (m)</th>
<th>Running Speed (m/s)</th>
<th>Weekly Running Mileage</th>
<th>Years Running</th>
<th>5000 m Run Best (min:s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Male</td>
<td>21 ± 2</td>
<td>72 ± 6</td>
<td>1.83 ± 0.05</td>
<td>4.0 ± 0.3</td>
<td>44 ± 16</td>
<td>7 ± 3</td>
<td>16:27 ± 1:35</td>
</tr>
<tr>
<td>Female</td>
<td>21 ± 2</td>
<td>56 ± 5</td>
<td>1.65 ± 0.08</td>
<td>3.5 ± 0.4</td>
<td>33 ± 16</td>
<td>6 ± 2</td>
<td>20:36 ± 2:54</td>
</tr>
</tbody>
</table>

Data Collection

An eight-camera Vicon Nexus motion capture system (Vicon, Centennial, CO) was used to capture three-dimensional kinematic data. The eight-camera system was mounted on an octagon frame hanging from the ceiling of the lab. Motion data
were collected at 160 Hz. Kinetic data were captured using an AMTI (AMTI, Watertown, MA) force platform mounted in the floor of a 30-meter runway. Force platform data were captured at 1600 Hz.

Participants wore tight spandex shorts or running shorts, a tight fitting short-sleeved or sleeveless shirt, and their own running shoes. Subjects were fitted with 23 retro-reflective markers. Markers were attached on the right leg and torso to the toe, lateral dorsifoot, heel, medial and lateral malleoli, anterior calf, lateral calf, medial and lateral tibial epicondyles, medial and lateral femoral epicondyles, anterior thigh, lateral thigh, left and right greater trochanters, left and right anterior superior iliac spine (ASIS), sacrum, left and right posterior superior iliac spine (PSIS), left and right acromion, and the cervicale. Markers were used to create segments for the foot, shank, thigh, and pelvis.

Participants performed a static trial at the start of data collection. Subjects stood in the middle of the lab on the force platform with arms outstretched, and data were collected for 2 seconds to obtain relative marker positions. Markers were removed from the medial malleolus, and the medial tibial and femoral epicondyles after completion of the static trial so that they did not interfere with running movements. The removed markers were recreated during the dynamic trials using transformations based on the relative position and orientation from the remaining markers on the segment.

There were five wedge conditions for each subject: medial 7° (MW7), medial 3° (MW3), no wedge (NW0), lateral 3° (LW3), and lateral 7° (LW7). Wedges were inserted into the shoes on top of the insole. Wedges were made from EVA (Shore
Durometer Type A-75) foam and donated by Marathon Orthotics (Eden Prairie, MN). The order of conditions was randomized between subjects to control for the effects of any sort of fatigue that may affect running mechanics. Subjects ran on the treadmill with the new wedges in their shoes at a self-selected speed for one minute before each wedge condition. Dynamic trials were performed on a 30 m runway through the video capture volume. Subjects were instructed to look straight ahead while running to avoid targeting the force platform. A successful trial was defined as hitting the force platform with the entire right foot without any visual evidence of targeting. Subjects were instructed to run at their preferred running velocity that was indicative of normal training pace. Trials within ±5% of the preferred running velocity were considered acceptable for analysis. Running velocity was monitored by calculating the average velocity of the sacral marker during stance phase.

Conditions were completed when five acceptable trials were recorded.

Data analysis

Motion and force data were imported to MATLAB (MathWorks, Natick, MA) for analysis. Custom MATLAB programs were used to calculate kinematics, kinetics, and iliotibial band (ITB) strain and strain rate.

Kinematic data were calculated throughout the stance phase. Stance phase was defined as first foot contact to toe off. First foot contact was defined as when the vertical ground reaction force exceeded 5% body weight. Toe-off was defined as when the vertical ground reaction force fell below 5% body weight.

Joint centers were calculated for the ankle, knee, and hip. The ankle joint center was calculated as the midpoint between the lateral and medial malleoli
markers. The knee joint center was calculated as the midpoint between the medial and lateral femoral epicondyle markers. The hip joint center was calculated as 25% of the distance between the left and right greater trochanter markers. Marker data were filtered using a dual-pass, fourth-order low pass Butterworth filter with a 10 Hz cutoff frequency. Joint angles were calculated using Euler/Cardan equations with a rotation order of flexion/extension, abduction/adduction, and internal/external rotation.

Joint moments were calculated using an inverse dynamics approach. A cutoff frequency of 20 Hz was used for force plate data. Segment masses, centers of mass, and moments of inertia were individually estimated (de Leva, 1996). Joint moments were transformed to the distal segment coordinate system and normalized by body mass. All joint moments were calculated as internal moments with the exception of knee varus, which is reported as an external moment due to a lack of musculature to generate this moment. Data from the stance phase were interpolated to 101 points for creation of ensemble curves.

A model of the ITB was developed by modifying the gait2392_simbody model in OpenSim (Delp et al., 1990). The knee joint was modified from a single degree of freedom joint (flexion/extension) to a three degree of freedom joint (flexion/extension, varus/valgus, and internal/external rotation). The hip joint was modeled as a three degree of freedom joint (flexion/extension, adduction/abduction, internal/external rotation). It was assumed that the ITB followed the same anatomical pathway as the tensor fascia latae (TFL). The resting length of the ITB was assumed to be the resting musculotendon length of the TFL.
For each of the three degrees of freedom of the revised knee joint and the standard hip joint, a polynomial equation was derived that calculated the ITB change in length as a function of joint angles. Joint angle limits to these equations were set to when the ITB length would go below the resting length or at physiological maximum ranges of motion. ITB length in the neutral position was the common intercept from the six ITB length equations when the joint angles were set to zero. The overall ITB change in length was the sum of ITB length changes from these six equations and neutral intercept value.

ITB strain during stance was computed using the following equation:

\[ \text{Strain}_i = \frac{L_i - L_0}{L_0} \]

Where \( L_i \) is the length of the ITB at data point ‘i’ and \( L_o \) is the resting length of the ITB. ITB strain rate was calculated at each time step using the first central difference method:

\[ \text{Strain Rate}_i = \frac{\text{Strain}_{i+1} - \text{Strain}_{i-1}}{\text{Time}_{i+1} - \text{Time}_{i-1}} \]

**Statistical Analysis**

Outliers caused by marker obscuring were detected by examining joint angle values and removed from further analysis (24 of 775 trials). Maximum values for kinematics, kinetics, peak strain, and peak strain rate were averaged across five trials per condition. Repeated measures MANOVA with between subjects’ comparisons of gender and within subjects’ comparison of wedge condition was performed on all dependent variables. Significant differences were set to alpha = 0.05. When significant main effects were detected, Tukey post hoc comparisons
were utilized to test for significant differences between wedge conditions. All statistical analyses were run in SPSS.

Results

The MANOVA indicated that there was a significant within-subjects main effect of wedge condition \( (p < 0.001) \) and a significant between subjects main effect of gender \( (p = 0.013) \). Interactions between gender and wedge condition were not statistically significant \( (p = 0.910) \).

Joint Kinematics

Maximum ankle eversion and hip internal rotation angles were significantly dependent upon wedge condition (Table 3.1). The LW7 \( (p < 0.001) \) and LW3 \( (p = 0.025) \) wedges had significantly higher and the MW7 \( (p = 0.008) \) wedge had significantly lower maximum ankle eversion angles than no wedge. The MW7 \( (p = 0.006) \) wedge had significantly higher maximum hip internal rotation angles than no wedge. Maximum knee valgus, knee internal rotation, and hip internal rotation angles were significantly dependent upon gender. Males had significantly higher maximum knee valgus angles \( (7.6 \pm 4.6^\circ \text{ vs. } 2.3 \pm 1.6^\circ, p < 0.001) \) and maximum knee internal rotation angles \( (13.4 \pm 3.5^\circ \text{ vs. } 9.0 \pm 1.6^\circ, p = 0.004) \) than females. Females had significantly higher maximum hip internal rotation angles \( (5.9 \pm 4.1^\circ \text{ vs. } 1.4 \pm 7.0^\circ, p = 0.032) \).
Table 3.1. Kinematic variables as a function of wedge conditions. Average values ± standard deviations. Significant differences between conditions are noted as superscripts below values in the tables (p < 0.05): a = significantly different from LW7, b = significantly different from LW3, c = significantly different from NW0, d = significantly different from MW3, e = significantly different from MW7.

<table>
<thead>
<tr>
<th>Angle (degrees)</th>
<th>LW7</th>
<th>LW3</th>
<th>NW0</th>
<th>MW3</th>
<th>MW7</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ankle Dorsiflexion</td>
<td>23.8 ± 4.0</td>
<td>23.9 ± 3.8</td>
<td>23.8 ± 3.6</td>
<td>23.9 ± 3.6</td>
<td>23.9 ± 3.6</td>
</tr>
<tr>
<td>Ankle Eversion</td>
<td>9.7 ± 5.5&lt;sup&gt;b,c,d,e&lt;/sup&gt;</td>
<td>7.8 ± 4.6&lt;sup&gt;a,c,e&lt;/sup&gt;</td>
<td>6.6 ± 5.2&lt;sup&gt;a,b,e&lt;/sup&gt;</td>
<td>6.8 ± 5.4&lt;sup&gt;a,e&lt;/sup&gt;</td>
<td>5.4 ± 5.4&lt;sup&gt;a,b,c,d&lt;/sup&gt;</td>
</tr>
<tr>
<td>Knee Flexion</td>
<td>47.0 ± 6.2</td>
<td>47.8 ± 5.9</td>
<td>47.1 ± 5.9</td>
<td>47.5 ± 5.6</td>
<td>47.2 ± 6.0</td>
</tr>
<tr>
<td>Knee Valgus</td>
<td>4.9 ± 4.6</td>
<td>5.1 ± 4.4</td>
<td>5.1 ± 4.4</td>
<td>4.8 ± 4.4</td>
<td>4.5 ± 3.9</td>
</tr>
<tr>
<td>Knee Internal Rotation</td>
<td>10.8 ± 5.1</td>
<td>11.1 ± 4.2</td>
<td>11.6 ± 4.3</td>
<td>11.0 ± 4.4</td>
<td>11.0 ± 4.3</td>
</tr>
<tr>
<td>Hip Flexion</td>
<td>31.3 ± 6.5</td>
<td>31.3 ± 6.0</td>
<td>31.4 ± 5.4</td>
<td>31.5 ± 5.4</td>
<td>30.9 ± 5.9</td>
</tr>
<tr>
<td>Hip Adduction</td>
<td>13.6 ± 5.6</td>
<td>13.7 ± 4.7</td>
<td>13.8 ± 4.2</td>
<td>13.5 ± 4.6</td>
<td>13.3 ± 4.7</td>
</tr>
<tr>
<td>Hip Internal Rotation</td>
<td>4.3 ± 6.4</td>
<td>3.3 ± 6.6</td>
<td>2.7 ± 5.8&lt;sup&gt;e&lt;/sup&gt;</td>
<td>3.9 ± 5.6</td>
<td>4.6 ± 6.3&lt;sup&gt;c&lt;/sup&gt;</td>
</tr>
</tbody>
</table>
Joint Kinetics

Maximum ankle plantar flexion, ankle inversion, and external knee varus moments were significantly dependent upon wedge condition (Table 3.2). The LW7 (p = 0.001), LW3 (p = 0.017), and MW7 (p < 0.001) wedges had significantly lower maximum ankle plantar flexion moments than no wedge. The MW7 (p < 0.001) wedge had significantly lower maximum ankle inversion moments than no wedge. In addition, the MW7 (p < 0.001) wedge had significantly higher maximum external knee varus moments than no wedge. Males had significantly higher maximum ankle plantar flexion moments (2.91 ± 0.35 Nm/kg vs. 2.39 ± 0.24 Nm/kg, p < 0.001), maximum ankle inversion moments (0.47
± 0.13 Nm/kg vs. 0.36 ± 0.11 Nm/kg, p = 0.017), and maximum knee extension moments (3.11 ± 0.51 Nm/kg vs. 2.68 ± 0.42 Nm/kg, p = 0.018) than females.

Table 3.2. Kinetic variables as a function of wedge conditions. Average values ± standard deviations. Significant differences between conditions are noted as superscripts below values in the tables (p < 0.05): a = significantly different from LW7, b = significantly different from LW3, c = significantly different from NW0, d = significantly different from MW3, e = significantly different from MW7

<table>
<thead>
<tr>
<th>Moment (Nm/kg)</th>
<th>LW7</th>
<th>LW3</th>
<th>NW0</th>
<th>MW3</th>
<th>MW7</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ankle Plantar Flexion</td>
<td>2.62 ± 0.39&lt;sup&gt;c&lt;/sup&gt;</td>
<td>2.64 ± 0.40&lt;sup&gt;c,e&lt;/sup&gt;</td>
<td>2.68 ± 0.39&lt;sup&gt;a,b,e&lt;/sup&gt;</td>
<td>2.66 ± 0.40&lt;sup&gt;e&lt;/sup&gt;</td>
<td>2.60 ± 0.39&lt;sup&gt;b,c,d&lt;/sup&gt;</td>
</tr>
<tr>
<td>Ankle Inversion</td>
<td>0.44 ± 0.12&lt;sup&gt;d,e&lt;/sup&gt;</td>
<td>0.43 ± 0.13&lt;sup&gt;d,e&lt;/sup&gt;</td>
<td>0.42 ± 0.13&lt;sup&gt;e&lt;/sup&gt;</td>
<td>0.41 ± 0.13&lt;sup&gt;a,b,e&lt;/sup&gt;</td>
<td>0.36 ± 0.14&lt;sup&gt;a,b,c,d&lt;/sup&gt;</td>
</tr>
<tr>
<td>Knee Extension</td>
<td>2.91 ± 0.51</td>
<td>2.91 ± 0.57</td>
<td>2.84 ± 0.52</td>
<td>2.87 ± 0.51</td>
<td>2.91 ± 0.50</td>
</tr>
<tr>
<td>Knee External Varus</td>
<td>1.00 ± 0.38&lt;sup&gt;e&lt;/sup&gt;</td>
<td>1.02 ± 0.38</td>
<td>1.01 ± 0.35&lt;sup&gt;e&lt;/sup&gt;</td>
<td>1.04 ± 0.38</td>
<td>1.10 ± 0.36&lt;sup&gt;a,c&lt;/sup&gt;</td>
</tr>
<tr>
<td>Hip Extension</td>
<td>1.91 ± 0.35</td>
<td>1.94 ± 0.36</td>
<td>1.94 ± 0.35</td>
<td>1.97 ± 0.38</td>
<td>1.99 ± 0.37</td>
</tr>
<tr>
<td>Hip Abduction</td>
<td>1.92 ± 0.30</td>
<td>1.91 ± 0.33</td>
<td>1.90 ± 0.27</td>
<td>1.89 ± 0.27</td>
<td>1.89 ± 0.32</td>
</tr>
</tbody>
</table>

**ITB Strain**

ITB strain (p = 0.943) and strain rate (p = 0.313) were not dependent upon wedge condition (Table 3.3). In addition, ITB strain (male 4.66 ± 0.75, female 4.98 ± 0.97, p = 0.244) and strain rate (male 42.4 ± 12.2 s<sup>−1</sup>, female 47.2 ± 9.8 s<sup>−1</sup>, p = 0.232) were not dependent upon gender.
Table 3.3. ITB strain and strain rate as a function of wedge conditions. Average values ± standard deviations.

<table>
<thead>
<tr>
<th></th>
<th>LW7</th>
<th>LW3</th>
<th>NW0</th>
<th>MW3</th>
<th>MW7</th>
</tr>
</thead>
<tbody>
<tr>
<td>ITB Strain</td>
<td>4.82 ± 1.03</td>
<td>4.84 ± 0.78</td>
<td>4.84 ± 0.88</td>
<td>4.75 ± 0.75</td>
<td>4.87 ± 0.99</td>
</tr>
<tr>
<td>ITB Strain Rate</td>
<td>45.1 ± 12.2</td>
<td>45.8 ± 11.0</td>
<td>43.6 ± 10.4</td>
<td>44.6 ± 11.4</td>
<td>45.3 ± 11.6</td>
</tr>
</tbody>
</table>

Discussion

The purpose of this study was to investigate the effects of wedge orthotics on ITB strain and strain rate. Our hypotheses were that ITB strain rate would increase with the use of lateral wedges and decrease with medial wedges. Our results fail to
support our hypotheses: ITB strain rate was not significantly different when comparing wedge conditions. The main reason for this finding was a lack of knee and hip kinematic changes when wearing different wedges. Five of the kinematic variables that were inputs to the ITB strain model were only minimally affected by wedge condition: knee flexion angles, knee varus/valgus angles, knee internal/external rotation angles, hip flexion-extension angles, and hip adduction/abduction angles. The exception was the maximum hip internal rotation angle, which was significantly increased with the medial 7° wedge, although the change was less than 2°.

As expected, the type of wedge produced systematic changes in maximum ankle eversion (Table 3.1). Clinicians should be aware of the significantly increased ankle eversion angles if lateral wedges are prescribed. Angle eversion angles were reduced by a total of 4.2° between the lateral 7° wedge and medial 7° wedge. It has been reported that the ratio for subtalar coupling is 1° of tibial internal rotation per 1.2-1.8° of eversion (McClay & Manal, 1997). According to that ratio, there would be a change of 2.3–3.5° of tibial internal rotation between the lateral 7° wedge and medial 7° wedge due to differences in ankle eversion. However, in this study there were no significant differences in maximum knee internal rotation angles as a function of wedge condition. The relationship between ankle eversion and knee internal rotation does not appear to be explained by a simple kinematic coupling when using wedge orthotics during running.

Maximum ITB strain and strain rate were statistically similar when comparing males and females. However, there were gender differences observed in
maximum knee valgus, knee internal rotation, and hip internal rotation angles. Males had higher maximum knee valgus angles by an average of 5.3° and higher maximum knee internal rotation angles by an average of 4.4°. The lower knee valgus angles would reduce ITB strain, while the higher knee internal rotation angles would increase ITB strain. Females had higher maximum hip internal rotation angles by an average of 4.5°, which would increase ITB strain. Strain and strain rate values were slightly larger for females than males, but the overall effect of kinematic differences due to gender appeared to cancel out. There were no significant interaction effects of wedge condition and gender, so even with kinematic differences wedges had similar effects on ITB strain and strain rate for males and females.

Not surprisingly, ankle inversion moments were affected by the type of wedge. The medial 7° wedge had significantly lower maximum ankle inversion moments as compared to no wedge (Table 3.2). Furthermore, there was a systematic decline in ankle inversion moments from the lateral 7° wedge to the medial 7° wedge. The decline in ankle inversion moments followed a similar pattern as the decline in maximum ankle eversion angles mentioned previously (Table 3.1). The type of wedge had a less predictable effect on ankle plantar flexion moments. The lateral 7° wedge, the lateral 3° wedge, and the medial 7° wedge all had significantly lower maximum ankle plantar flexion moments than no wedge (Table 3.2). It appears that adjusting the foot’s alignment either medially or laterally may reduce one’s capability to functionally generate ankle plantar flexion moments during running.
It is of concern that maximum external knee varus moments were significantly higher when using the medial 7° wedge as compared to no wedge. Soft tissues that cross the lateral portion of the knee are potentially put under greater strain by external knee varus moments (Boldt et al., 2013). Increased external knee varus moments may also increase ITB compression against the lateral femoral epicondyle, a theorized mechanism for ITBS development (Fairclough et al., 2006). The current kinematic model estimates ITB strain, but does not estimate ITB compression. A structural or finite element model of the ITB may provide further insight into whether increased external knee varus moments produce increased ITB compression. If a medial wedge is prescribed, then it is suggested that increased external knee varus moments are considered along with the reduced ankle eversion angles and ankle inversion moments.

Males displayed significantly higher maximum ankle plantar flexion moments, maximum ankle inversion moments, and maximum knee extension moments. However, no significant interactions were found between wedge condition and gender, so the type of wedge had a similar effect on males and females despite kinetic differences. Higher joint moments in males may be attributed to a faster average running speed (4.0 m/s) than females (3.5 m/s).

One limitation of this study is the use of standard non-fitted wedges for all participants. Use of custom fitted orthotics may lead to benefits not observed in this study. A second limitation is the analysis of healthy participants. Ideally, this study would have been completed prospectively with participants who developed ITBS symptoms or with participants who currently suffer from ITBS symptoms. A third
limitation was the participants were not fatigued during running, which has been associated with ITBS symptom onset. Previous research has shown that kinematics of individuals with ITBS change when run to voluntary exhaustion (Miller et al., 2007). Finally, the model is limited by individual differences and the complexity of the ITB structure. The estimated ITB strain does not reflect if participants have tighter or stiffer ITB properties and does not take into account any potential effects of gluteus maximus activation.

A six degree of freedom ITB model was developed to analyze the effects of wedge orthotics on ITB strain rate during running. There was no evidence that the use of lateral or medial wedges reduced ITB strain rates. Medial 7° wedges resulted in potentially beneficial reductions in maximum ankle eversion angles and ankle inversion moments, but with the potential disadvantage of increased external varus knee moments. In contrast, lateral 7° wedges resulted in potentially unfavorable increases in maximum ankle eversion angles.

References


CHAPTER 4
GENERAL CONCLUSIONS

A six degree of freedom ITB model was developed to analyze the effects of wedge orthotics on ITB strain rate during running. There was no evidence that the use of lateral or medial wedges reduced ITB strain rates. Medial 7° wedges resulted in potentially beneficial reductions in maximum ankle eversion angles and ankle inversion moments, but with the potential disadvantage of increased external varus knee moments. In contrast, lateral 7° wedges resulted in potentially unfavorable increases in maximum ankle eversion angles. Further study using custom fitted orthotics, advanced modeling of ITB compression, individual ITB stiffness measures, fatigue effects, and a subject pool including individuals with ITBS is needed in order to further understand if and how orthotics can aid in the prevention or rehabilitation of ITBS.
APPENDIX

INFORMED CONSENT, QUESTIONNAIRE, RECRUITMENT FLYER

The following documents are the informed consent and questionnaire signed and completed by participants and the recruitment flyer used for this study.

INFORMED CONSENT DOCUMENT

Title of Study: The effects of medial and lateral wedges on iliotibial band strain during overground running.

Investigators: Evan Day, Dr. Jason Gillette

This form describes a research project. It has information to help you decide whether or not you wish to participate. Research studies include only people who choose to take part—your participation is completely voluntary. Please discuss any questions you have about the study or about this form with the project staff before deciding to participate.

Introduction

The purpose of this study is to investigate how using medial and lateral wedge shoe inserts affect iliotibial band strain during running. Data from this study will be used to further knowledge about prevention and rehabilitation of iliotibial band syndrome and other related running injuries.

You are being invited to participate in this study because you are a healthy recreational or competitive runner that runs at least 15 miles per week. You should not participate if you: have suffered a lower extremity injury in the past 3 months, have had surgery on a lower extremity in the past 12 months, currently wear orthotics, or are currently pregnant.

Description of Procedures

Before participation, you will fill out a medical history questionnaire to ensure you are eligible for participation. If you agree to participate, then you will be asked to visit the Biomechanics Lab (178N) in the Forker building once to complete the following tasks. You will wear tight fitting clothes, such as compression shorts or short running shorts, and a tight fitting top or sleeveless shirt or jersey. If you do not have clothes that meet these criteria, then the lab can provide them for you. We will record your age and measure your height and weight prior to data collection. Retroreflective markers will be placed on you at specific landmarks on your right foot, right leg, pelvis, and trunk. Marker movement will be tracked by an 8-camera system in the lab. You will be instructed to hit the force platform in the middle of
the lab with your right foot. The force platform records the forces produced during foot contact. Data collection will consist of 25 over-ground running trials with a short treadmill warm-up in between conditions. There will be 5 different degrees of wedges that you will run with. The wedge conditions are 7° medial, 3° medial, no wedge, 3° lateral, and 7° lateral. You will run through each condition 5 times at a pace representative of normal training pace. You are allowed practice trials to get comfortable with the equipment. You will run in your own shoes. You will complete a short questionnaire at the end of data collection. Your participation will last for approximately 45-60 minutes.

**Risks or Discomforts**

While participating in this study you may experience the following risks or discomforts:

- Muscle soreness
- Fatigue
- Skin irritation from markers

**Benefits**

If you decide to participate in this study, there will be no direct benefit to you. It is hoped that the information gained in this study will benefit society by advancing our knowledge on how wedged inserts can treat running injuries.

**Costs and Compensation**

You will not have any costs from participating in this study. You will not be compensated for participating in this study. If you are a student in Kin 355, then you may receive 1 extra credit point for your participation.

**Participant Rights**

Participating in this study is completely voluntary. You may choose not to take part in the study or to stop participating at any time, for any reason, without penalty or negative consequences.

If you have any questions about the rights of research subjects or research-related injury, please contact the IRB Administrator, (515) 294-4566, [IRB@iastate.edu](mailto:IRB@iastate.edu), or Director, (515) 294-3115, Office for Responsible Research, Iowa State University, Ames, Iowa 50011.

**Research Injury**

Emergency treatment of any injuries that may occur as a direct result of participation in this research is available at the Iowa State University Thomas B. Thielen Student Health Center and/or referred to Mary Greeley Medical Center or another physician or medical facility at the location of the research activity.
Compensation for any injuries will be paid if it is determined under the Iowa Tort Claims Act, Chapter 669 Iowa Code. Claims for compensation should be submitted on approved forms to the State Appeals Board and are available from the Iowa State University Office of Risk Management and Insurance.

**Confidentiality**

Records identifying participants will be kept confidential to the extent permitted by applicable laws and regulations and will not be made publicly available. However, federal government regulatory agencies auditing departments of Iowa State University, the Food and Drug Administration (FDA), and the Institutional Review Board (a committee that reviews and approves human subject research studies) may inspect and/or copy study records for quality assurance and data analysis. These records may contain private information.

To ensure confidentiality to the extent permitted by law, the following measures will be taken:

Your name will not be used in the data collection, and you will be assigned an alphanumeric number instead. Your name and information/data will be kept in a separate secure location. Computerized records for motion analysis will be kept on password protected computers of Evan Day and Dr. Gillette, while all other information will be kept in Dr. Gillette’s office. The individuals who will have immediate access to the identifiable research records are Dr. Gillette and Evan Day. If the results are published, then your identity will remain confidential.

**Questions**

You are encouraged to ask questions at any time during this study. For further information about the study, contact Evan Day (eday@iastate.edu) or Dr. Jason Gillette (gillette@iastate.edu).

If you have any questions about the rights of research subjects or research-related injury, please contact the IRB Administrator, (515) 294-4566, IRB@iastate.edu, or Director, (515)294-3115, Office for Responsible Research, Iowa State University, Ames, Iowa 50011.

**Consent and Authorization Provisions**

Your signature indicates that you voluntarily agree to participate in this study, that the study has been explained to you, that you have been given the time to read the document, and that your questions have been satisfactorily answered. You will receive a copy of the written informed consent prior to your participation in the study.
Participant's Name (printed) __________________________________________

__________________________________________
Participant's Signature                     Date
Participant Questionnaire

Subject Number:

Personal Information

Age:

Body Mass:

Height:

Current Weekly Mileage:

How many years running:

Type of runner: Recreational        Competitive

Shoe size:

5000m personal record:

Lower extremity injury/surgery history:

Thoughts on comfort of wedge inserts:
Department of Kinesiology

Volunteers Needed For Running Study

The Effects of Medial and Lateral Wedges on Iliotibial Band Strain During Overground Running

Who: Recreational or competitive runners who run at least 15mi/week and are currently free of any injuries.

What: Participants will visit the Biomechanics Lab once. Data collection will consist of over-ground running trials with varying degrees of medial and lateral wedges in their shoes to evaluate how shoe inserts affect iliotibial band strain during running. Data collection will last 45-60 minutes.

Where: Testing will occur in the Biomechanics Laboratory at 178N in the Forker Building.

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