Lower extremity mechanics of iliotibial band syndrome during an exhaustive run

Ross Herbert Miller
Iowa State University

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Lower extremity mechanics of iliotibial band syndrome during an exhaustive run

by

Ross Herbert Miller

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

MASTER OF SCIENCE

Major: Exercise and Sport Science (Biological Basis of Physical Activity)

Program of Study Committee:
Jason C. Gillette, Major Professor
Timothy R. Derrick
Philip Dixon

Iowa State University
Ames, Iowa
2006

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Graduate College
Iowa State University

This is to certify that the Master’s thesis of

Ross Herbert Miller

has met the thesis requirements of Iowa State University

Signatures have been redacted for privacy
Dedicated to my fellow HHP graduate students.
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<tr>
<td>$D$</td>
<td>stiffness coefficient (Hooke’s law)</td>
</tr>
<tr>
<td>$F$</td>
<td>force</td>
</tr>
<tr>
<td>$L$</td>
<td>length</td>
</tr>
<tr>
<td>$S$</td>
<td>transformation matrix component</td>
</tr>
<tr>
<td>$i$</td>
<td>anterior-posterior direction</td>
</tr>
<tr>
<td>$j$</td>
<td>medial-lateral direction</td>
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<td>$k$</td>
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<td>sample size</td>
</tr>
<tr>
<td>$t$</td>
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</tr>
<tr>
<td>$u$</td>
<td>uniaxial deformation (Hooke’s law)</td>
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</tr>
<tr>
<td>$y$</td>
<td>global medial-lateral coordinate</td>
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<tr>
<td>$z$</td>
<td>global vertical coordinate</td>
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Greek Symbols

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<th>Symbol</th>
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<tbody>
<tr>
<td>$\Delta$</td>
<td>change in a quantity</td>
</tr>
<tr>
<td>$\alpha$</td>
<td>statistical significance level (Type I error rate)</td>
</tr>
<tr>
<td>$\beta$</td>
<td>Cardan vertical rotation angle</td>
</tr>
<tr>
<td>$\epsilon$</td>
<td>strain</td>
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\( \theta \)  
Cardan anterior/posterior rotation angle, or general angular displacement

\( \phi \)  
Cardan medial/lateral rotation angle

\( \omega \)  
angular velocity

**Subscripts**

0  
indicates an initial value

\( T \)  
tension

\( n \)  
time step index or local coordinate system

\( max \)  
indicates a maximum value

**Abbreviations**

3D  
three-dimensional

ACL  
anterior cruciate ligament

ASIS  
anterior-superior iliac spine

COP  
center of pressure

CRP  
continuous relative phase

GCS  
global coordinate system

GRF  
ground reaction force

ITB  
iliotibial band

ITBS  
iliotibial band syndrome

LCS  
local coordinate system

LFE  
lateral femoral epicondyle

OT  
Ober test

SD  
standard deviation

SIMM  
Software for Interactive Musculoskeletal Modeling
ACKNOWLEDGEMENTS

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ABSTRACT

Purpose: To determine the role of fatigue in the biomechanics of iliotibial band syndrome (ITBS) in distance running. Methods: Sixteen recreational runners ran to voluntary fatigue on a motorized treadmill at a self-selected speed (average time: 16.2 min). Eight runners had a history of ITBS. The positions of twenty-three retro-reflective markers were recorded by an eight-camera 120 Hz motion capture system. Joint and segment angles during stance were exported to a numerical musculoskeletal model (SIMM 4.0) with the iliotibial band (ITB) modeled as a passive structure in order to determine the tension and strain in the ITB. Results: With the onset of fatigue: (1) knee flexion at heel-strike was significantly greater in the ITBS runners (20.6° vs. 15.3°, \( p = 0.01 \)), (2) the number of knees with the ITB impinged upon the lateral femoral epicondyle increased from six to eleven in the ITBS runners, and (3) tension in the ITB at heel-strike rose by 10.3%. Strain and tension in the ITB was higher in the ITBS runners throughout stance. Maximum knee internal rotation velocity and hip internal rotation were also significantly different between groups. Conclusions: Runners with a history of ITBS showed higher levels of strain and tension in the ITB and more ITB impingement than control runners during a run to fatigue. Higher strain and tension appear to be related to changes in knee-flexion at heel-strike and transverse kinematics at the hip and knee. Key Words: ILIOTIBIAL BAND SYNDROME, RUNNING, FATIGUE, STRAIN, TRANSVERSE KINEMATICS
CHAPTER 1 INTRODUCTION

1.1 General Introduction

Knee injuries are a common source of pain to many athletes, especially runners. About 60% of all runners are injured in an average year, and one third of those injuries occur at the knee due to the large cyclical loads placed on the knee during running [1]. Iliotibial band syndrome (ITBS) is an overuse injury of the lateral knee that has been frequently documented in long-distance runners, but also observed in other populations who make repetitive movements, such as cyclists, weight-lifters, and military personnel. ITBS is believed to develop when the iliotibial band (ITB) becomes inflamed due to friction with the lateral femoral epicondyle (LFE) as the two structures impinge on one another during running.

1.1.1 Symptoms, causes, and treatments

The primary symptom of ITBS is pain on the lateral aspect of the knee, near the LFE. The pain is not constant, but follows an arc that peaks near heel-strike (approximately 30° of knee flexion) as the ITB shifts to the posterior side of the LFE. Running tends to aggravate ITBS, causing the pain to increase as the run continues until the runner is forced to stop. In severe cases, the pain can be so intense that it prevents normal walking [2]. Friction between the ITB and the LFE is unavoidable and occurs naturally in healthy runners. ITBS is believed to develop when the friction is elevated beyond the physiological norm. Friction can be elevated due to factors that cause additional
impingements of the ITB over the LFE, such as a sudden increase in training mileage or speed, and factors that increase the tension in the ITB and consequently increase the compression between the ITB and the LFE. These factors include, but are not limited to, long stride lengths, high volumes of downhill running, a lack of ITB flexibility, hip abductor weakness, and cavus feet [2–4].

ITBS is usually alleviated in one to three weeks by conservative treatments of rest, ice, stretching, and training programs to correct muscular weaknesses. Local steroidal injections have also been successful in treating ITBS [5]. In cases where prolonged conservative treatments have failed, surgical resectioning of the ITB can relieve symptoms by removing a small section of the ITB near the LFE [6, 7].

1.1.2 Biomechanics of ITBS

While clinical investigations of ITBS have been numerous in recent years [5, 8–11], the biomechanics of ITBS have received comparatively little attention. Orchard et al. [12] performed a dynamic treadmill study of ITBS runners and found no differences in knee flexion angles between the runners’ symptomatic and asymptomatic legs. Comparisons for angles in other planes and at other joints were not reported. Messier et al. [13] demonstrated that runners with a history of ITBS reduced maximal rearfoot pronation when compared to runners in a control group. This finding was contrary to the usual assumption of excessive rearfoot motion as a mechanism of overuse injuries. The ITBS runners also showed faster maximum supination velocities, smaller maximum breaking forces, and were shorter and lighter than the control group.

Unfortunately studies such as Orchard et al. [12] and Messier et al. [13] are rare, and both are nearly a decade old. There is presently a lack of in-depth investigations of ITBS biomechanics that have followed up on these early studies. There has also been little research on mathematical modeling of ITBS. While studies on human subjects can provide kinematic, kinetic, and muscle electromyography data, they cannot explain how
individual muscles and tissues contribute to motion at every instant during running. Modern computational power allows for simulations of human movement that are an order of magnitude more detailed than what was possible in the past [14]. Mathematical models can also allow for detailed investigations of factors that are difficult or impossible to manipulate with human subjects, such as incremental anthropometric changes or various muscle force optimization schemes.

1.1.3 Motivation and goals

The present research is motivated by the prevalence of ITBS in runners and the lack of biomechanical studies in this area. While previous research has shown significant associations between kinematic and kinetic variables and ITBS, these studies are few in number, and many important injury mechanisms are in need of further investigation. While ITBS is recognized as an overuse injury, the effects of fatigue on the etiology of ITBS have not been reported. Fatigue has long been suggested as a component of lower extremity injuries [18]. Additionally, some researchers have suggested an association between excessive internal tibial rotation and lower extremity injuries in runners [20–22].

The goal of the present research is to expand the base of knowledge of ITBS biomechanics using both videography and mathematical modeling. The videographic component will measure the kinematics of runners with and without a history of ITBS during a run to voluntary fatigue. It is hypothesized that with the onset of fatigue, runners with a history of ITBS will exhibit larger changes in internal tibial rotation and rearfoot motion than an age-matched control group. A mathematical model of the ITB and its associated anatomy will also be constructed to investigate the effects of a fatigued run on the strain and tension in the ITB. A second hypothesis is that ITB strain and tension will be higher in the ITBS runners, and that this difference will increase with fatigue.
1.2 Literature Review

1.2.1 General description and frequency of ITBS

Iliotibial band syndrome, also referred to as the iliotibial band friction syndrome, was first described by Colson and Armour [23] and subsequently by Renne [24], Orava [25], and Noble [2, 26]. It is an overuse injury of the outer knee and is frequently observed in long-distance runners and other athletes who make repetitive movements, such as cyclists and weight-lifters. The frequency of ITBS can be quite high. Kirk et al. [10] reported that the incidence rate of ITBS ranges from 2% to 52% depending on the population under consideration. Among populations of runners, Messier et al. [13] reported a frequency of 1.6% to 12%. Pinshaw et al. [19] reported a frequency of 12% during a clinic on 196 injured runners and cited ITBS as the third most common running injury behind patellofemoral pain and shin splints. In spite of its observed frequency, however, ITBS has not traditionally received a great deal of attention from researchers.

1.2.2 Anatomy of the iliotibial tract

The first step in understanding ITBS is to recognize the associated anatomy of the iliotibial tract. The ITB is a thick sheet of connective tissue that runs from the iliac crest down the lateral side of the thigh to the lateral tibial epicondyle (Gerdy’s tubercle). The primary function of the ITB is to provide lateral stability to the knee during motion. Near its proximal end, the ITB provides insertions for the gluteus maximus and the tensor fascia lata. At its distal end, the ITB terminates at Gerdy’s tubercle and also attaches to the anterior tibialis and the tendon of the biceps femoris. A bursa located at the tibial attachment site enables the band to slide over the LFE during knee flexion and extension. When the knee flexes past an angle of approximately 30°, the ITB shifts to the posterior side of the LFE. When the knee extends back under 30°, the ITB shifts back to the anterior side of the LFE. During running, the ITB shifts back and forth
across the LFE repeatedly and generates friction. ITBS is diagnosed when the ITB becomes inflamed, irritated, and causes pain during running.

Birnbaum et al. [27] performed an extensive investigation of the anatomy and mechanical properties of the ITB using eighteen fresh cadavers (average age seventy-three years). The average length of the ITB between the iliac spine and Gerdy’s tubercle was 510 mm. The average width of the ITB at the iliac spine, sub-trochanter region, and tibial insertion was 59 mm, 90 mm, and 43 mm, respectively. The anatomical investigation revealed the following pieces of information:

1. The ITB splits into a superficial portion and a deep portion. The deep portion covers the tensor fasciae latae.

2. The tensor fasciae latae has an insertion at the ITB.

3. The ITB is not fixed to the greater trochanter, but slides over it during movement.

4. The ITB passes over the vastus lateralis.

5. Part of the gluteus maximus inserts at the ITB.

6. The ITB terminates in bundles at Gerdy’s tubercle, the fibular head, the lateral intermuscular septum, and the transverse and longitudinal patellar retinaculum.

In six of the cadavers, the ITB was dissected between the iliac spine and Gerdy’s tubercle and subjected to uniaxial tensile strength testing by a tearing apparatus. The dissected ITB had an average stiffness coefficient of 17 N/mm and sustained an average ultimate tensile force of 860 N. The stress-strain response of the ITB was approximately linear in the elastic region, suggesting that ITB tissue deforms linearly according to Hooke’s law:

\[ F = D \cdot u \]  
(1.1)
where $F$ is the applied tensile force, $D$ is the stiffness coefficient, and $u$ is the uniaxial deformation. After verifying the linear stress-strain response of the ITB, Birnbaum et al. [27] created a finite-element model of the femur with the ITB modeled as a set of five linear springs in parallel. The goal of the finite-element model was to calculate the hip-centralizing forces that the ITB places on the greater trochanter. The force placed on the greater trochanter by the ITB was introduced by high-stiffness contact spring elements. A similar technique could likely be used to model the interaction between the ITB and LFE.

### 1.2.3 Symptoms and treatments

The primary symptom of ITBS is poorly localized pain on the lateral side of the knee near the LFE. The pain typically onsets after a sudden change in training protocol. Of the forty-eight runners diagnosed with ITBS in Sutker et al. [28], most had significantly altered their weekly mileage, training pace, training surface or terrain, or footwear prior to the onset of ITBS symptoms.

The symptoms of ITBS are frequently alleviated by conservative treatments. Barber and Sutker [29] reported successful non-surgical treatment of ITBS in nineteen athletes over a period of three years by prescribing rest, training reduction, footwear changes, gait modifications, anti-inflammatories, local steroidal injections, and stretching. Gunter and Schwellnus [5] demonstrated the capability of local steroidal injections to reduce pain in runners with recent onset of ITBS. Participants ran on a treadmill for thirty minutes at their 10-km race pace and reported pain on a visual analog scale every minute, with the total pain during running taken as the area under the pain-time curve. Runners who received a corticosteroid injection near the LFE reported greater reductions in total pain during follow-up runs one week and two weeks after the injection, in comparison to a placebo control group.

Unfortunately, conservative treatments are not always successful. Noble [2] reported
initial success in only thirty of seventy-three ITBS patients using local steroidal injec-
tions. The remaining patients required a second or third injection before they were able
to return to running pain-free. Fourteen patients required four to six weeks of total rest
following the third injection before returning to running, and only nine patients returned
to running with no recurrence of pain.

Surgery has been effective in relieving the pain of ITBS when prolonged conservative
treatments have failed. The most frequently-used surgical procedure involves a technique
called band resectioning whereby the surgeon removes a small triangular piece of the
posterior fibers of the ITB near the LFE. In an early report on surgical treatment of
ITBS, Martens et al. [7] performed band resectioning surgery on twenty-three patients
whose symptoms were resistant to conservative treatments. The surgery was highly
successful in the majority of the patients. Few follow-up complications arose and the
patients returned to activity quickly. Drogset et al. [30] reported that of the forty-five
ITBS patients operated on at Trondheim University Hospital between 1989 and 1996,
only one patients had poor results (a minor post-operative infection).

The severity of ITBS cases is typically established by the degree of inflammation
present in the ITB. Due to the prohibitive costs of advanced medical scanning tech-
nologies, diagnoses are usually made based on patients’ medical histories and physical
examinations. These methods cannot determine if the severity of ITBS symptoms are
such that conservative treatments will be sufficient for recovery. Magnetic resonance
imaging (MRI) has been successful in revealing ITB inflammation and making definitive
diagnoses. Ekman et al. [31] compared the MRI findings in ITBS knees to those in
control knees. The ITBs in the ITBS knees were twice as thick on average compared to
the ITBs in the control knees. The comparison also showed the presence of fluid accu-
mulation beneath the ITB and near the LFE in the ITBS knees. In another MRI study
of ITBS, Nishimura et al. [32] did not show increased ITB thickness in the symptomatic
knees. Rather, the MRI evidence suggested that the posterior fibers of the ITB (near
the LFE) where tighter than the anterior fibers (away from the LFE). Murphy et al. [33] demonstrated the ability of MRI to confirm or establish diagnoses of ITBS in patients with lateral knee pain. Six patients with clinical histories and physical examinations indicative of ITBS underwent MRI scanning. The imaging results were consistent with those of Ekman et al. [31] and indicated inflammation beneath the ITB and near the LFE.

MRI is clearly an effective tool for diagnosing the severity of ITBS in individual patients. Unfortunately, it is also an expensive technology with limited availability. National Imaging Associates reported in 2004 that a single MRI scan costs a patient $700 to $900. The machinery required for the scan costs around $2 million, with an additional $800,000 spent annually in training and maintenance costs [34].

1.2.4 Biomechanical studies of ITBS

The difficulty in dealing with ITBS lies not in diagnosis and treatment, but in prevention. Biomechanical investigations of runners prone to developing ITBS can help to reveal injury mechanisms and suggest prevention strategies. In comparison to the number of studies focusing on treatments, however, there have been very few studies on the biomechanics of ITBS. To date there have been only two major studies focusing on the biomechanics of ITBS in distance runners. Messier et al. [13J studied the etiology of ITBS in distance runners with a series of clinics that investigated training variables, anthropometrics, isokinetic strength of the knee, rearfoot movement, and ground reaction forces (GRF). Knee strength was measured using an isokinetic dynamometer. Rearfoot movement data were recorded during a fifteen minute treadmill run by a 200 Hz video camera system. GRF data were collected while subjects ran over a runway equipped with a force platform. The ITBS runners put in more weekly mileage, were less experienced runners, had more recently changed their training program, and spent more time running on composite tracks and swimming, than the control group. Height was the only
anthropometric discriminator (ITBS runners were shorter), although weight was also significant when height was removed from the analysis. The tendency for ITBS runners to be shorter and lighter lent support to the hypothesis that larger runners are better able to support the impacts of running and attenuate shocks to prevent injury [35]. The ITBS runners were isokinetically weaker in a number of strength and endurance variables for knee flexion/extension and generated less peak torque and performed less muscular work. There were no significant discriminators between the injured and non-injured legs of the ITBS runners. The kinematic analysis revealed that ITBS runners had smaller calcaneal to vertical touchdown angles and faster maximum supination velocities. The ITBS runners also utilized smaller maximum braking forces. A combined discriminant analysis revealed that weekly mileage and maximum normalized breaking force were the most significant discriminators between the ITBS and control groups.

In an earlier study, Messier and Pittala [36] investigated the relationships between various biomechanical, anthropometric and training variables in runners suffering from common overuse injuries, including ITBS. Results showed that the ITBS runners tended to have slightly higher arches, less range of motion in ankle dorsiflexion, more pronation, more total rearfoot movement, and faster pronation velocity. However, none of these trends were statistically significant.

The second major biomechanical study of ITBS in runners was performed by Orchard et al. [12]. They studied the anatomy of the iliotibial tract with eleven cadaver knees, and also performed a dynamic study of runners who had shown symptoms of ITBS in one leg within the last six months. The asymptomatic legs served as the control group. In the cadaver study, the legs were fixed in full extension due to the embalming process. Large variations were present in ITB width and in the portion of the ITB overlaying the LFE in full extension. The dynamic study measured joint kinematics in the sagittal plane using optical motion capture of reflective markers placed on the lower extremity. Subjects performed two two-minute runs on a level treadmill at a self-selected pace.
(2.78–3.89 m/s). The second run was performed with a heel lift inserted into the shoe of the symptomatic side. There were no significant differences in knee, hip, and ankle angles in the sagittal plane for the symptomatic and asymptomatic legs in this study. The study concluded that friction between the ITB and LFE occurs at or slightly below 30° of knee flexion, and that downhill running predisposes a runner to developing ITBS by reducing the knee flexion angle at heel-strike.

Two conclusions can be drawn from the ITBS research by Messier et al. [13] and Orchard et al. [12]. First, investigations of ITBS kinematics and kinetics should include a separate control group of runners with no history of ITBS. Second, the kinematics of the transverse and frontal planes should be considered when examining the etiology of the injury.

### 1.2.5 The role of fatigue in injury

An important aspect of ITBS that has not been well-documented in the literature is the role of fatigue. Fatigue has long been recognized as a precursor to running injuries. As runners fatigue, their form breaks down and they are more likely to exhibit abnormal motion that may lead to injury. Verkerke et al. [15] induced fatigue by enforcing a standardized running pace (10 km/hr) on a treadmill. Runners who reached exhaustion demonstrated a large increase in step time variability (the time between two successive ipsilateral heel-strikes). Slawinski and Billat [16] used 3D motion analysis to measure the total mechanical cost of an overground run to exhaustion at the velocity associated with 95% of $\dot{V}O_{2,\text{max}}$ and found that while the kinetic and potential energies were unchanged, the internal mechanical energy requirement rose by 9%. Gerlach et al. [17] measured vertical GRF on female runners before and after a run to fatigue. Peak impact decreased by 6% and loading rate decreased by 11%. The changes were attributed to changes in stride length, stride frequency, and joint kinematics, although they did not report specific kinematic adjustments. An interaction between injuries during the previous year and
changes in kinetics suggested that injury-prone runners are less successful at attenuating impact forces and loading rates as they fatigue. Nyland et al. [18] demonstrated that the onset of fatigue delayed maximum knee flexion and knee flexor/extensor activations when performing a run and rapid stop movement. They suggested that the knee may be the primary site of force attenuation during fatigue.

Shock attenuation is another important parameter that may be affected by fatigue. Derrick et al. [37] examined the kinematic adjustments made to absorb shocks during a run to voluntary fatigue. Accelerometers were mounted to the head and shank, and goniometers were attached to the knee and rearfoot. Results showed that near the end of the run, knee flexion at impact and rearfoot supination at impact both increased, resulting in a smaller effective mass of the shank. Consequently, peak shank acceleration (6.11 g to 7.38 g) and impact attenuation (74.5% to 77.5%) both increased as the runners fatigued. The increased shank accelerations were not considered an injury risk due to the corresponding decrease in the effective mass of the shank. Mizrahi et al. [38] measured shank impact acceleration along with EMG activity of the ankle dorsi and plantar flexors during a thirty-minute run above the anaerobic threshold. An imbalance in shank muscle contraction developed along with the increase in shank impact accelerations. The muscular imbalance may expose the tibia to excessive bending stresses.

Because ITBS is an overuse injury, fatigue may play an important role in its development. The previous research suggests a link between fatigue and running injuries, although the role of fatigue in ITBS has not been investigated extensively.

1.2.6 Axial rotation of the lower leg

The present research hypothesizes that ITBS runners will exhibit greater fatigue-induced changes in internal rotation at the lower leg than a control group. The total range of internal tibial rotation has previously been reported to be near 15° [20, 21, 39]. Previous research has indicated that excessive internal rotation may predispose runners
to developing injuries. Nigg et al. [39] found that the transfer of foot eversion to internal tibial rotation increased with increasing arch height in runners and suggested that the transfer of movement may be related to knee pain during running. Bellchamber and van den Bogert [21] investigated the contributions of moments at the knee and ankle to tibial rotation during running and found that the majority of tibial rotation is driven at the knee-level (rather than at the ankle-level) during stance. However, their subject pool did not include runners with knee pain.

The coupling of rearfoot eversion with internal tibial rotation has been investigated as a mechanism of injury in runners [40–44]. McClay and Manal [40] investigated coupling using the ratio of rearfoot motion to tibial rotation in runners with and without abnormal rearfoot mechanics (excessive eversion). The ratio was significantly smaller for the abnormal group due to their greater internal tibial rotation, and the timing of tibial rotation and rearfoot motion were more closely matched in the control runners. Hamill et al. [41] presented an innovative approach to analyze lower extremity motion coupling by analyzing the continuous relative phases (CRP) between segments, including foot inversion-tibial rotation. Runners with patellofemoral pain demonstrated less CRP variability than healthy control runners. The authors suggested that the lack of variability in the injured runners indicated the presence of repeatable joint actions and may be useful in diagnosing and treating injuries.

1.2.7 Left-right asymmetry

Several studies have investigated kinetic and kinematic symmetry between the left and right legs during running [45–49]. The degree of symmetry reported has varied due to differences in methodology, participant population, and the definition of symmetry. Vagenas and Hoshizaki [47] studied the kinematic asymmetry of the lower leg during running on a motorized treadmill and found asymmetries in calf angle, rearfoot angle, pronation/supination velocity, and the timing of some kinematic variables. In an earlier
study, the same authors found significant kinematic asymmetries in the lower leg and foot, and proposed that a runner’s degree of asymmetry may be altered by injury or running style [46].

Tashman et al. [22] studied the effectiveness of ACL reconstruction on knee kinematics during running. While running downhill at least four months after surgery, the runners’ reconstructed knees were slightly more externally rotated (3.8°) and adducted (2.8°) than their healthy knees. Although the change were small, they were statistically significant, consistent across a small subject pool (n = 6), and suggest that left-right leg asymmetry is worthwhile to investigate when dealing with runners with an injury history.

1.3 Organization

This thesis is organized into five chapters: (1) Introduction, (2) Methods, (3) Results, (4) Discussion, and (5) Manuscript. The introduction included a review of relevant literature on iliotibial band friction syndrome and related topics in running biomechanics. The methods chapter details the collection and analysis of data, and the results and discussion chapters present the important findings of the study and discuss their implications. The manuscript is formatted according to Gait and Posture specifications. The manuscript includes cited references, and a full bibliography is included at the end of the thesis. Appendices containing extended results, the informed consent document, a participant questionnaire, and MATLAB programs are also included.
CHAPTER 2 METHODS

2.1 Participants

Sixteen participants were recruited for the study via e-mail contact and word-of-mouth. Eight of the participants had a history of iliotibial band syndrome (ITBS) in one or both legs and were classified in the "ITBS" group. The remaining eight participants had no history of ITBS and were classified in the "control" group. The required group sample size was estimated at \( n = 7 \) by a power calculation to detect a 4.4° difference in knee flexion angle, with a standard deviation of 2.9° [37], power of 0.80, and a significance level of 0.05. The control runners were age-matched to within four years of an ITBS runner. Average age, height, and weight for both groups are given in Table 2.1. The ITBS runners were slightly older, shorter, and lighter than the control group; however, none of the differences were statistically significant \((p < 0.05)\).

Participants in the ITBS group were required to be free of any ITBS symptoms for at least three months prior to participating in the study. Participants in either group were required to be symptom-free of any lower extremity injury for at least three months before participating. The study was approved by the Institutional Review Board at Iowa State University. Participants gave informed written consent prior to participating in the study.

<table>
<thead>
<tr>
<th>ITBS ((n = 8))</th>
<th>27.5 ± 9.0</th>
<th>170.1 ± 6.9</th>
<th>68.7 ± 15.9</th>
</tr>
</thead>
<tbody>
<tr>
<td>Control ((n = 8))</td>
<td>26.4 ± 7.7</td>
<td>172.8 ± 8.9</td>
<td>71.3 ± 14.4</td>
</tr>
</tbody>
</table>
study. The informed consent document is provided in Appendix B.

After entering the lab, participants completed a questionnaire addressing their training regimen and injury history. Runners with a history of ITBS reported their diagnoses (by medical doctor, physical therapist, or other), pain patterns (during and after running, after running only, while running only, or while running when fatigued) and symptomatic legs. The questionnaire is provided in Appendix C.

Participants’ iliobial band (ITB) flexibilities were assessed by the Ober test [50, 51]. Participants were positioned on their sides while the tester adducted the thigh until the pelvis began to rotate. Flexibility was quantified by the angle displaced by the thigh prior to pelvic rotation, and measured from the level horizontal. Angles were measured using a mechanical inclinometer. The Ober test was performed on both legs and at 0° and 90° of knee flexion. Leg lengths were measured as the distance from the anterior-superior iliac spine (ASIS) to the ipsilateral medial malleolus.

2.2 Kinematic Methods

This section details the experimental setup, collection of marker coordinate data, and calculation of joint and segment angles.

2.2.1 Experimental setup

An eight-camera Peak Motus optical motion capture system (Vicon Peak, Centennial, CO) was used to capture three-dimensional optical data. The cameras were mounted on an octagonal frame attached to the lab ceiling. The sampling rate of the cameras was 120 Hz. The Peak Motus system tracked the orientation of body segments with a set of twenty-three spherical retro-reflective markers 2 cm in diameter. The markers were attached on both sides of the body at the toe, lateral dorsi-foot, heel, medial and lateral malleoli, anterior calf, medial and lateral knee, anterior thigh, greater trochanter,
and ASIS. A marker was also placed on the L5/S1 region of the lower back. Figure 2.1 shows a participant running on the treadmill with the marker set illuminated. Figure 2.2 shows the marker set as viewed from the frontal plane.

The markers defined body segments for the feet, shanks, thighs, and torso. Each segment was defined by at least three markers in order to allow for the calculation of axial rotations. The Peak Motus system was calibrated before each data collection such that errors in marker positions were less than 2 mm. To minimize the chance of markers falling off during running, many markers were secured using black athletic tape and a spray adhesive.

### 2.2.2 Data collection

A Quinton treadmill (Stairmaster Inc., Kirkland, WA) was situated on the laboratory floor, at the center of the octagonal camera frame. Figure 2.3 shows a schematic of the treadmill and camera system. Participants first performed a “static” trial by standing on the treadmill and facing the front of the room (towards camera 1 in Fig. 2.3), with their feet shoulder-width apart and their arms across their chest. The Peak Motus system recorded marker locations in this position for one second. The resulting set of data consisted of \((x,y,z)\) spatial coordinates captured at 120 Hz.

Following the static trial, the four medial markers at the ankles and knees were removed to avoid collisions between these markers during running. Nineteen markers remained for the “dynamic” trials. The removed markers were later recreated during the dynamic trial data analysis (see Section 2.2.3). To begin the dynamic trial, participants selected a running pace (reported in minutes per mile) that would induce moderate fatigue after ten to twenty minutes of running. The selected pace was typically near a recent 5-km race pace. Participants were allowed to warm up at a slower pace for two minutes before the treadmill was set to the selected pace. The pace was measured by placing a Biddle 359985 contact hand tachometer (AVO International Instruments, Blue
Figure 2.1  A participant running on the treadmill during dynamic trial data collection. Markers are illuminated by the camera flash.
Figure 2.2 Frontal plane view of the marker set. Black markers are on the right leg, white markers are on the left leg. 1 = toe; 2 = dorsifoot; 3 = heel; 4,5 = lateral and medial malleoli; 6 = anterior calf; 7,8 = lateral and medial knee; 9 = anterior thigh; 10 = greater trochanter; 11 = ASIS; 12 = L5S1.
Figure 2.3  Schematic of the data collection setup (camera system and treadmill).
Bell, PA) on the treadmill belt while the participant was running. The dynamic trials began once the treadmill reached the selected pace. Marker positions were recorded in ten second blocks at the beginning of the dynamic trial and at every two minutes thereafter until the participant reached voluntary fatigue. The resulting blocks of dynamic trial data consisted of \((x,y,z)\) spatial coordinates for each of the nineteen markers captured at 120 Hz. Participants ran in their own usual running shoes and wore the black laboratory shirt and shorts shown in Fig. 2.1.

### 2.2.3 Data analysis

Data from the static and dynamic trials were analyzed with a custom-made program written in MATLAB 7.0.4 (The MathWorks Inc., Natick, MA) [52]. The code for this program is given in Appendix D. Using the temporal averages of the static marker coordinates, the program estimated joint center locations for the hips, knees, and ankles, and defined segment coordinate axes on the feet, calves, thighs, and torso. Neutral position joint angles for the hips, knees, and ankles were calculated based on the orientations of the segment axes relative to one another during the static trial. Neutral segment angles were calculated based on the absolute orientations of the segment axes with respect to a global coordinate system (GCS) during the static trial. The GCS was defined in the Peak Motus system calibration and was consistent across all subjects and trials. Joint and segment angles were calculated according to standardized Canadian Society for Biomechanics definitions [53].

Joint and segment angles were calculated as Cardan angles using three successive rotations: flexion/extension, adduction/abduction, and internal/external rotation. The rotation sequence was selected to conform to standard definitions of physiological motion [54]. Using the segment coordinate axes defined by the marker coordinates, Cardan angles were calculated using the following transformation:
where $S_{11}$ through $S_{33}$ are the transformation matrix components, $\hat{i}_n$, $\hat{j}_n$, and $\hat{k}_n$ are the local coordinate system (LCS) axes in the anterior-posterior, medial-lateral, and vertical directions, respectively, and $\hat{i}$, $\hat{j}$, and $\hat{k}$ are the GCS axes in the anterior-posterior, medial-lateral, and vertical directions, respectively. For joint angles, the transformation matrix was set equal to the product of the GCS segment axes of the superior segment and the transpose of the GCS segment axes of the inferior segment. For segment angles, the transformation matrix was set equal to the GCS segment axes. All segments were modeled with three degrees of rotational freedom. A Cardan angle matrix was derived by multiplying the initial rotation about the medial-lateral axis by the rotation about the anterior-posterior axis, and finally by the rotation about the vertical axis:

\[
\begin{bmatrix}
\hat{i}_n \\
\hat{j}_n \\
\hat{k}_n
\end{bmatrix} =
\begin{bmatrix}
1 & 0 & 0 \\
0 & \cos \phi & \sin \phi \\
0 & -\sin \phi & \cos \phi
\end{bmatrix}
\times
\begin{bmatrix}
\cos \beta & \sin \beta & 0 \\
-\sin \beta & \cos \beta & 0 \\
0 & 0 & 1
\end{bmatrix}
\times
\begin{bmatrix}
\cos \theta & 0 & -\sin \theta \\
0 & 1 & 0 \\
\sin \theta & 0 & \cos \theta
\end{bmatrix}
\times
\begin{bmatrix}
\hat{i} \\
\hat{j} \\
\hat{k}
\end{bmatrix}
\] (2.2)
The functions $k_p$ and $k_n$ preserve the signs of the numerator and denominator for the \texttt{atan2()} function [55].

Dynamic trial data blocks were analyzed in the same fashion as the static trial data, except that the analysis was carried out at each time step rather than with a temporal average. Since the medial markers were removed for the dynamic trials, the program constructed virtual points to represent these markers based on their relative positions in the static trials. Each reconstructed marker was calculated using three dynamic markers located on an adjacent segment. The dynamic trial data were smoothed with a 4th-order low pass symmetric Butterworth filter with a cutoff frequency of 10 Hz. Filtering was carried out using the MATLAB functions \texttt{butter()} and \texttt{filtfilt()} [56]. To account for differences in anthropometrics and marker placements that may have influenced the dynamic angles, the neutral angles were subtracted from the dynamic angles at each time step.

Within the dynamic trials, the stance phase of each stride (both left leg and right leg) was extracted by finding successive minimum $z$-coordinates for the heel and toe of the appropriate leg. Local minima for the heel and toe were located using a moving search window based on the average stride length during the trial. Stride length was initially estimated by averaging the number of time steps between each successive pair of minima on the dynamic hip flexion angle. The sets of stance phase dynamic angles were fit to 100-point splines using the MATLAB \texttt{spline()} function. Ensemble curves
for hip, knee, and ankle joint angles and thigh, calf, and foot segment angles during stance were calculated by averaging the spline-fit angle-time curves together across all strides.

Joint and segment angular velocities during stance were calculated with a first-order central difference equation:

\[ \omega_n = \frac{\theta_{n+1} - \theta_{n-1}}{2\Delta t} \]  

where \( \omega \) is the angular velocity, \( \theta \) is the angular position of the joint or segment, \( n \) is the time step index (from 1 to 100), and \( \Delta t \) is the time step (1/120 s).

### 2.3 Modeling Methods

This section details the construction of a musculoskeletal model of the ITB and its associated anatomy, and the measurement of ITB impingements, strains, and tensions during the stance phase.

#### 2.3.1 Model construction

To determine how the kinematics affected the mechanics of the ITB, a numerical musculoskeletal model was created using SIMM 4.0 (MusculoGraphics Inc., Santa Rosa, CA). SIMM is a graphics-based software package for the development and analysis of musculoskeletal models. Details on the development of SIMM can be found in the literature [56–58].

The model included the pelvis/sacrum, femur, tibia/fibula, patella, talus, calcaneus, and metatarsal bones. The model also included forty-three muscles of the lower extremity. Muscles followed a basic Hill-type muscle model, with tendon force-length, active force-length, passive force-length, and force-velocity relationships defined for each structure [60, 61]. While there are some aspects of muscle mechanics that cannot be
accurately represented by the Hill model, it is generally adequate for describing muscular action during voluntary human movement [62].

The ITB was included in the model by defining three additional structures that originated at the iliac crest and terminated at the femoral insertion of the gluteus maximus, the fibular head, and Gerdy’s tubercle. The maximum isometric force of the structures, based on the cadaveric data of Birnbaum et al. [27], was 860 N. Since the ITB consists of passive tissue, the active force-length and the force-velocity relationships of the ITB structures were not incorporated. Figure 2.4 shows a sagittal-plane view of the SIMM model, and Figure 2.5 shows a close-up view of the SIMM model’s knee with the ITB impinged.

Impingement between the ITB and the lateral femoral epicondyle (LFE) was modeled by defining a wrapping sphere whose surface was flush against the outer surface of the LFE. Impingement was defined as instances during stance when the ITB was in contact with the LFE wrapping sphere (see Fig. 2.5). The diameter of the LFE wrapping sphere in the default model was 2.24 cm. To prevent the ITB structures from passing through the femur during motion, wrapping spheres were also defined at the greater trochanter and the femoral neck (diameters 1.64 cm and 4.48 cm, respectively) and wrapping cylinders were defined along the body of the femur. The diameters of the wrapping spheres were chosen to approximately match the diameters of the bone geometry in the default SIMM model.

Each participant’s ensemble angles at the left and right hip, knee, and ankle joints were imported into SIMM motion files. The bone and muscle geometries were scaled by the ratio between the participant’s leg length (see Section 2.1) and the default leg length length of the SIMM model (91.5 cm from ASIS to ipsilateral medial maleolus). Maximum isometric forces of the muscles and ITB structures were scaled by the ratio between the participant’s body mass and the default body mass of the SIMM model (75 kg).
Figure 2.4 Wrapping geometry of the SIMM model.
2.3.2 Model data collection

The motion files defined by the joint angles were imported into the scaled musculoskeletal models. The following analyses were made from each motion file:

1. The knee flexion angle at heel-strike,

2. Whether the ITB impinged with the LFE at heel-strike,

3. The strain in the ITB during stance, and

4. The tension in the ITB during stance.

Impingement was determined if the ITB structure had begun to wrap around the LFE wrapping sphere at heel-strike (see Fig. 2.5). Strain was calculated by dividing the change in length of the ITB by the length of the ITB when the model was in the anatomical position:

\[ \epsilon = \frac{L - L_0}{L_0} \cdot 100 \]  

(2.8)

where \( \epsilon \) is the ITB strain, \( L \) is the ITB length, and \( L_0 \) is the ITB length in the anatomical position. Tension in the ITB was calculated by applying the passive force-length relation shown in Fig. 2.6 to the strain data.

Figure 2.5 Close-up view of the SIMM model’s knee with the ITB impinged.
The passive force-length curve shown in Fig. 2.6 is characterized by the following equation:

\[ \frac{F_T}{F_{T,\text{max}}} = -\sqrt{1 - \frac{\epsilon^2}{(0.174)^2}} + 1 \]  

(2.9)

where \( F_T \) is the tension in the ITB, \( F_{T,\text{max}} \) is the ultimate tensile force of the ITB [27], and the value 0.174 is the maximum sustained strain in the ITB before rupture, normalized to the resting ITB length [27, 63]. The ultimate tensile force \( F_{\text{max}} \) was scaled by the ratio of the body mass of the participants to the body mass of the default SIMM model (75 kg).

### 2.4 Statistical Analysis

Left-right asymmetries of the joint and segment angles were assessed using matched pair \( t \)-tests, with a pair defined as the difference between the left-side angle and the same right-side angle for a particular subject. Asymmetry was assessed for the ITBS group \( (n = 8) \) and control group \( (n = 8) \), and for the entire participant pool grouped together.
(n = 16).

Because it was desired to determine which of the dependent variables accounted for the most variability between groups, discriminant analysis was an appropriate statistical test [64]. Discriminant analyses were performed on the following data sets:

1. Questionnaire responses, anthropometrics, and flexibility measurements,
2. Angles from the first and last 10 s blocks of the dynamic trials,
3. Angular velocities from the first and last 10 s blocks of the dynamic trials,
4. The timing of maximums and minimums for the dynamic trials, and
5. Changes in kinematics that occurred during the dynamic trials.

Two-by-two factorial ANOVA tests with group (ITBS, control) and fatigue status (rested, fatigued) as the independent variables were performed on the dependent variables from the discriminant analyses that accounted for the most variability in the data. Post-hoc comparisons between the variables that were significantly different between groups were performed with Tukey adjustments to account for multiple comparisons. The ITB strains and tensions calculated by the SIMM model were compared using t-tests. A difference between groups was considered significant at the $\alpha = 0.05$ level. SAS 9.1.3 (SAS Institute, Cary, NC) was used for the statistical analysis.
CHAPTER 3 RESULTS

3.1 Questionnaire Responses and Measurements

Of the eight iliotibial band syndrome (ITBS) runners, six had suffered from ITBS symptoms in both legs. One runner had symptoms only in the right leg, and one runner had symptoms only in the left leg. Two ITBS runners had a history of patellar tendonitis. Six ITBS runners had been diagnosed by a physical therapist and two had been diagnosed by a medical doctor. None of the control runners had a history of lower extremity injury. All sixteen runners in the study described their training intensity as either “Recreational” (defined as running only for fitness and/or enjoyment) or “Recreational/Competitive” (defined as running for fitness and training for one to three races per year).

Table 3.1 shows the training data for the ITBS and control groups. The ITBS group trained at a considerably higher average weekly mileage than the control group (23.7 mi/week for ITBS vs. 11.8 mi/week for control, p = 0.06). The control group had maintained their current training mileage for a slightly longer period of time (6.0 months for ITBS vs. 6.3 months for control) and ran their most recent 5k race at a slightly faster pace (23.4 min for ITBS vs. 22.9 min for control).

Table 3.1 also shows the Ober test measurements for iliotibial band (ITB) flexibility, taken on both legs at 0° and 90° of knee flexion. On average, the control group was more flexible than the ITBS group in all four measurements, however none of the differences in flexibility were statistically significant.

Five of the eight ITBS runners had a leg-length discrepancy of at least 3 mm, com-
Table 3.1  Training data and Ober test (OT) measurements. Values are mean ± SD. n = 8 for both groups.

<table>
<thead>
<tr>
<th></th>
<th>ITBS</th>
<th>Control</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Weekly mileage</td>
<td>23.7 ± 15.1</td>
<td>11.8 ± 5.9</td>
<td>0.06</td>
</tr>
<tr>
<td>Months at mileage</td>
<td>6.0 ± 3.2</td>
<td>6.3 ± 2.5</td>
<td>0.87</td>
</tr>
<tr>
<td>5k time (min)</td>
<td>23.4 ± 5.3</td>
<td>22.9 ± 2.6</td>
<td>0.84</td>
</tr>
<tr>
<td>OT right leg, 0° (deg.)</td>
<td>17.4 ± 6.8</td>
<td>19.9 ± 11.5</td>
<td>0.62</td>
</tr>
<tr>
<td>OT right leg, 90° (deg.)</td>
<td>12.6 ± 7.4</td>
<td>13.9 ± 11.7</td>
<td>0.79</td>
</tr>
<tr>
<td>OT left leg, 0° (deg.)</td>
<td>15.9 ± 5.2</td>
<td>17.0 ± 8.4</td>
<td>0.56</td>
</tr>
<tr>
<td>OT left leg, 90° (deg.)</td>
<td>13.2 ± 5.5</td>
<td>15.2 ± 7.1</td>
<td>0.54</td>
</tr>
</tbody>
</table>

Table 3.2  Dynamic trial pace and duration data. Values are mean ± SD. n = 8 for both groups.

<table>
<thead>
<tr>
<th></th>
<th>ITBS</th>
<th>Control</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Time (min)</td>
<td>16.14 ± 4.96</td>
<td>16.31 ± 6.03</td>
<td>0.95</td>
</tr>
<tr>
<td>Pace (min/mi)</td>
<td>9.48 ± 2.16</td>
<td>7.84 ± 0.98</td>
<td>0.08</td>
</tr>
</tbody>
</table>

pared to three out of eight control runners. Subotnick suggested that a leg-length discrepancy of more than 3 mm may require intervention for runners [65].

3.2  Dynamic Trials and Kinematics

Table 3.2 shows the times to fatigue and the paces selected by the runners during their dynamic trials. Both groups ended their runs after approximately sixteen minutes (16.1 min for ITBS vs. 16.3 min for control). The control group tended to run at a faster pace than the ITBS group (9.48 min/mi for ITBS vs. 7.84 min/mi for control, p = 0.08). Two of the slower ITBS runners did not report a 5k time on the questionnaire, which helps explain the large difference in group paces compared to the small difference in group 5k times. One ITBS runner reported mild lateral knee pain near the end of the dynamic trial.
3.2.1 Left-right asymmetry

There were significant left-right asymmetries in the following combinations of trials, groups, and kinematic angles:

1. ITBS rested: minimum thigh adduction, maximum foot adduction
2. ITBS fatigued: minimum thigh adduction
3. Control rested: minimum thigh adduction, minimum foot inversion
4. Control fatigued: maximum thigh adduction, minimum foot inversion
5. All rested: minimum thigh adduction, minimum foot inversion, maximum calf adduction
6. All fatigued: maximum thigh adduction, minimum foot inversion

In light of the presence of left-right asymmetries between all combinations of subject groups and trials, the remaining statistical analyses were carried out with each leg, rather than each runner, as an observational unit (n = 16 for the ITBS group and n = 16 for the control group).

**A note on terminology:** For joint angles at the hip and knee, and segment angles at the thigh and calf, motion in the frontal plane is referred to as “adduction/abduction” and motion in the transverse plane is referred to as “internal/external rotation.” For ankle joint angles and foot segment angles, motion in the frontal plane is referred to as “inversion/eversion” and motion in the transverse plane is referred to as “adduction/abduction.” The inconsistency in the definition of adduction/abduction was chosen to keep the terminology consistent with clinical conventions.
3.2.2 Kinematics

Kinematic angles during stance and angular velocities during stance were compared between the ITBS group and the control group. During the rested trials, there were significant differences in maximum foot adduction (2.6 ± 7.9° for ITBS vs. -6.1 ± 7.6° for control, p = 0.005) and minimum knee flexion (12.5 ± 3.6° for ITBS vs. 7.7 ± 3.8° for control, p = 0.01). At heel-strike during the rested trials, the ITBS runners’ feet were significantly more adducted (internally rotated) than the control feet (4.8 ± 6.3° for ITBS vs. -2.3 ± 5.7° for control, p = 0.001). During the fatigued trials, there were significant differences in maximum knee flexion (43.8 ± 7.5° for ITBS vs. 36.5 ± 9.2° for control, p = 0.01), minimum knee flexion (11.1 ± 4.2° for ITBS vs. 7.4 ± 6.2° for control, p = 0.05), and maximum foot inversion (3.3 ± 10.9° for ITBS vs. -9.5 ± 19.6° for control, p = 0.02). When comparing the changes that occurred between the rested and fatigued trials, there was a significant difference in maximum knee flexion (1.3° for ITBS vs. -1.4° for control, p = 0.008).

During the rested trials, there was a significant difference in maximum ankle dorsiflexion velocity (74.8 ± 19.7°/s for ITBS vs. 54.9 ± 27.9°/s for control, p = 0.02). During the fatigued trials, there were significant differences in minimum thigh flexion velocity (-29.5 ± 16.9°/s for ITBS vs. -16.7 ± 8.2°/s for control, p = 0.01) and maximum knee internal rotation velocity (16.4 ± 9.3°/s for ITBS vs. 10.3 ± 4.0°/s for control, p = 0.02). When comparing the changes that occurred between the rested and fatigued trials, there were no significant differences in angular velocities.

The timing of minimum and maximum kinematic angles during stance were unaffected by fatigue in either group. Between the two groups, there were large differences in timing for rested maximum hip internal rotation (5% of stance for ITBS vs. 60% of stance for control, Fig. 3.1), fatigued minimum hip adduction (85% of stance for ITBS vs. 100% of stance for control, Fig. 3.2), and fatigued maximum ankle adduction (30%
of stance for ITBS vs. 92% of stance for control, Fig. 3.3). When averaged over all subjects, the differences in timing were statistically significant ($p < 0.05$).

### 3.3 Musculoskeletal Model

Figures 3.4 and 3.5 show the average knee flexion angle at heel-strike and the number of knees impinged at heel-strike, respectively. During the rested trials, knee flexion angle at heel-strike was similar between the two groups ($17.2 \pm 7.0^\circ$ for ITBS vs. $15.2 \pm 3.9^\circ$ for control, $p = 0.16$). As the runners fatigued, the ITBS knees were significantly more flexed at heel-strike compared to the control knees ($20.6 \pm 7.8^\circ$ for ITBS vs. $15.3 \pm 5.2^\circ$ for control, $p = 0.01$). The number of impinged knees at heel-strike increased with fatigue for the ITBS group (from six to eleven) but decreased slightly for the control group (from six to four). In all of the unimpinged knees, the ITB was located anteriorly to the lateral femoral epicondyle (LFE) at heel-strike, and impingement had not yet occurred.
Figure 3.2  Hip adduction during stance (fatigued trials).

Figure 3.3  Ankle adduction during stance (fatigued trials).
Figure 3.4 Knee flexion angle at heel-strike. * = significant difference from fatigued control ($p = 0.03$).

Figure 3.5 Number of knees with ITB-LFE impingement at heel-strike.
3.3.1 Iliotibial band strain

Figure 3.6 displays the strain in the ITB during stance. At heel-strike, both groups had approximately 6% strain under both rested and fatigued conditions. Strain in the ITBS runners rose quickly after heel-strike, and the ITBS runners demonstrated more ITB strain than the control runners for the final 90% of stance. Fatigue did not have a large effect on ITB strain; however, the ITBS runners showed slightly less strain during the middle 50% and the final 15% of stance while fatigued.

Figure 3.7 shows the peak ITB strain during stance. Peak strains for the ITBS runners were 8.5 ± 1.2% (rested) and 8.4 ± 1.3% (fatigued). Peak strains for the control runners were 7.5 ± 0.7% (rested) and 7.5 ± 0.8% (fatigued). The peak strains were significantly greater in the ITBS runners (p = 0.005 for rested, p = 0.03 for fatigued).
3.3.2 Tension in the iliotibial band

Figures 3.8 and 3.9 show the tension in the ITB during stance and the tension in the ITB at heel-strike, respectively. Differences between groups in the tension data were not significant due to the large variation in tension data between participants, although some important trends were present. The rested ITBS, rested control, and fatigued control trials showed similar tensions near heel-strike of approximately 50 N. As the ITBS runners fatigued, their tension at heel-strike increased from 49.0 N to 54.1 N, an increase of 10.3%. The control runners only demonstrated a small decline in tension at heel-strike (49.7 N rested vs. 49.6 N fatiged).

After the first 10% of stance, the tension in the ITBS runners began to rise above the tension in the control runners. For the final 90% of stance, the ITBS runners demonstrated greater tension than the control runners. While fatigued, the tension in the ITBS runners was greater than the tension in control runners throughout the entire stance phase. While rested, the peak tension was 95.9 N in the ITBS runners and 73.5
N in the control runners. When fatigued, the peak tension was 91.1 N in the ITBS runners and 74.7 N in the control runners. When looking at fatigue-induced changes in tension over the entire stance phase, the ITBS runners slightly increased their tension from 0–20% of stance and slightly decreased their tension from 20–70% of stance. The control runners demonstrated only minor changes in tension during stance.
Figure 3.8  ITB tension during stance.

Figure 3.9  ITB tension at heel-strike.
4.1 Conclusions

The major kinematic finding from the study was the increase in knee flexion and the corresponding increase in the number of impinged knees at heel-strike for the iliotibial band syndrome (ITBS) runners with the onset of fatigue. The increase in knee flexion at heel-strike is consistent with previous results on kinematic adjustments during a run to fatigue [37, 67], although only the ITBS runners demonstrated a significantly greater knee flexion angle at heel-strike in the present study. Greater knee flexion may also indicate a smaller calf angle at heel-strike, which has been associated with less efficient running economy [66]. In the present study, calf flexion at the beginning of the run was greater for the ITBS group than for control, but at the end of the run the ITBS group had a smaller calf flexion angle than control. Neither difference was significant between groups, however the group difference in changes in calf flexion that occured with fatigue was close to significance ($p = 0.07$).

Previous research has suggested that decreasing the angle of knee flexion at heel-strike predisposes a runner to developing ITBS by increasing the amount of time spent in the impingement range of knee flexion angles during stance [2, 12]. The present study offers an alternative explanation that increasing knee flexion at heel-strike places a runner’s iliotibial band (ITB) in a compromised orientation where impingement with the lateral femoral epicondyle (LFE) during heel-strike is more likely to occur. The implications of this mechanism are discussed in the following sections.
4.1.1 Strain and tension

The hypothesis that ITBS runners would demonstrate increased strain in the ITB with fatigue was not supported. There was only a small (less than 0.5%) increase in strain for the ITBS runners during the first 10% of stance (see Fig. 3.7); however, the ITBS runners did demonstrate higher strain than the control runners throughout the entire stance phase (see Fig. 3.6). Under the assumption that higher strain places more stress on the ITB and puts the ITBS runners closer to their injury threshold [68], the strain data suggest that the ITBS runners are more likely to develop an overuse injury related to the ITB.

The increased strain in the ITBS runners was likely a function of the discriminating kinematic variables from the dynamic trials. Previous research [9, 69] along with the present results have suggested a zone of impingement for the ITB rather than a defined point. The minimum knee flexion angle was significantly larger for the ITBS runners both at the beginning and end of the run, suggesting that their knees were closer to the impingement zone during stance than the control knees. This condition could potentially increase the stance-time spent in the impingement zone, which may lead to injury. Changes in strain from heel-strike to toe-off suggest that the ITB in the control runners released from impingement towards the end of stance while the ITB in the injured runners remained impinged. From Fig. 3.6, the control runners decreased their strain from 6% at heel-strike to 5% at toe-off, and the ITBS runners increased their strain from 6% at heel-strike to 7% at toe-off.

The present results supported the hypothesis that ITBS runners would increase the tension in the ITB as they fatigued, although only during the first 20% of stance (see Fig. 3.9). Based on these results, an injury mechanism for ITBS is proposed: with fatigue, as the knee becomes more flexed at heel-strike, the odds for and degree of impingement between the ITB and LFE increase. Impingement at heel-strike, when impact forces
are first transmitted through the soft tissues of the leg, acts to increase the immediate tension in the ITB, causing more irritation and pain. The ITBS runners increased their tension at heel-strike by 10.3% with the onset of fatigue, suggesting that the initial "shock" of impact and the immediate loading of the ITB play important roles in injury.

The ITBS group's limited flexibility (see Table 3.1) may have also contributed to their ITB tension. Under rested conditions, the ITBS runners had a noticeably higher loading rate following heel-strike and during the approach to maximum tension (see slopes of Fig. 3.8), lending further support to the idea that the condition of the ITB at heel-strike is an important factor in the injury mechanism.

Although the results suggest an important role for heel-strike impacts in the pathology of ITBS, the remainder of stance should not be neglected. It is notable that strain and tension in the ITB were higher in the ITBS runners throughout all of stance, indicating that the ITB does not get as much of a chance to recoil and release tension in the runners with a history of the injury. Novacheck [70] notes that the active impact forces that occur during the latter 3/4ths of stance are larger in magnitude and of longer duration than the passive impact forces that occur near heel-strike. Peak strain and tension in both groups occurred near mid-stance, regardless of fatigue. Additionally, the peak strain was significantly higher in the ITBS runners, although the peak tension was not significantly higher due to the large spread of the tension data. Group differences in strain and tension during the latter phases of stance were likely due to the kinematic differences discussed in the next section.

4.1.2 Kinematics of ITBS runners

With the onset of fatigue, tension and strain both decreased during the middle 50% of stance in the ITBS runners while tension and strain in the control runners was unaffected by fatigue. The faster maximum internal knee rotation exhibited by the ITBS runners when fatigued may indicate an effort to leave the impingement zone, since rotating the
knee internally should change the location of the ITB relative to the LFE. However, only one ITBS runner reported lateral knee pain during the dynamic trial, and it is unlikely that the runners placed conscious control over this movement.

The internal hip rotation during stance (see Fig. 3.1) offers another possible explanation for the elevated strain and tension in the ITBS runners. At heel-strike, the ITBS runners’ hips were internally rotated to 14.5° and then externally rotated during the rest of stance. The control runners’ hips were internally rotated to only 7° and continued to be internally rotated for the first 65% of stance. Externally rotating the hip should move the LFE away from the posterior edge of the ITB, while flexing the knee should move the posterior edge of the ITB closer to the LFE. Since these motions occur simultaneously during stance, the combined effects may cancel each other out and act to maintain impingement between the ITB and LFE.

The kinematics of the foot suggest some interesting explanations for the elevated ITB tension and strain in the ITBS runners. While rested, the ITBS runners had a greater peak foot adduction (internal rotation) angle during stance than the control runners. As a segment angle, foot internal rotation is measured relative to the global coordinate system, and could indicate a combined effect of net internal rotation at the hip, knee, and ankle. The ITBS feet were also significantly more internally rotated than the control feet at heel-strike, indicating more internal rotation of the lower extremity during swing as the leg prepares for impact. This aspect of movement is predominantly controlled by motor activity. Conversely, foot internal rotations during heel-strike are the result of rapid joint loading, which may be due to structural factors rather than motor control. Internal rotation of the lower extremity should load the ITB under tension. The significant difference in foot orientation in the transverse plane may indicate an overall greater internal rotation of the entire lower extremity, even though group differences in internal rotation of the thigh and calf were not significant. Once fatigued, the ITBS runners exhibited more foot inversion than control runners, indicating more
overall adduction of the lower extremity that may have compressed the ITB against the lateral knee. Fairclough et al. [71] demonstrated with MRI that the ITB compresses against LFE as the knee flexes from full extension and suggested that ITBS develops due to a compressive mechanism rather than a sliding friction mechanism.

4.2 Prevention of Injury

The long-term goal of overuse running injury research should be to reduce the frequency of overuse injuries. Nigg and Bobbert [72] noted that there is no indication that load analysis during running has contributed to a reduction in injury frequency. One possible explanation is that injury research tends to be limited to retrospective studies of runners who are already injured. The knowledge base of injury biomechanics would benefit greatly from prospective investigations that follow a large group of runners over time and compare the mechanics of those who develop overuse injuries to those who do not. In a prospective study of four hundred previously healthy university students, those who developed lower extremity injuries during the study exhibited higher maximum foot eversion, more laterally-oriented roll-off of the foot, and a more centralized heel-strike center of pressure (COP) [73]. A more centralized COP could be partially accounted for by increased knee flexion at heel-strike. Future work should consider examining COP variability from stride to stride. Low variability could indicate repeated stress to the same tissues and trauma accumulation.

The present research has suggested several variables that may predispose runners to developing ITBS, based on comparisons with healthy runners. Although the present research was retrospective, treatments of ITBS tend to focus on the symptoms of the injury rather than its causes. The ITBS runners may therefore have still exhibited the lower extremity kinematics that influenced their injury etiologies. Clinically, the results of this research could be used to adjust the kinematics of ITBS runners, possibly via
orthotic inserts or gait retraining, to relieve symptoms and guard against future injury. Foot orthotics are capable of inducing significant changes in muscle EMG activity during running [74] and could potentially be used to relieve tension in the ITB.

4.3 Limitations and Future Work

Future investigations into the etiology of ITBS would benefit from including a strength testing component. Some researchers have proposed that ITBS may be influenced by weakness in the gluteal muscles [9, 75]. Overall hip muscle weakness has also been associated with overuse running injuries [76]. The initial protocol of the present study included an isometric strength test of hip adduction. Data were collected on one subject before an equipment malfunction forced the testing to be excluded from subsequent data collections. Strength data would also be useful in a musculoskeletal model to investigate how muscular weakness affects the tension in the ITB. In a preliminary investigation, ground reaction forces (GRF) and lower extremity kinematics were collected for a healthy young participant running over a force plate. The MATLAB optimization function fmincon() [77] was used to solve the indeterminate system of equations and distribute joint moments into the major hip and knee flexion/extension muscles based on cubic stress minimization. The optimization function demonstrated an inverse relationship between the maximum isometric force in the gluteal muscles and the predicted tension in the ITB. As the maximum isometric force allowed in the gluteal muscles was lowered, the tension was increased in the other structures of the model, including the ITB [unpublished results].

The present study was limited by the lack of kinetic data. Numerous authors have suggested a relationship between impact forces and overuse running injuries [77–80]. Without GRF data, it was not possible to realistically estimate the tension in the ITB during stance. The tensions estimated in this study were determined solely from the
strain in the ITB and the passive force-length relationship shown in Fig. 2.6 and should not be interpreted as physiologically realistic tensions due to the lack of GRF contributions. However, even without GRFs, the ITB tension estimates shown in Figs. 3.8 and 3.9 provide a relative comparison of passive ITB tension between the ITBS runners and control runners and the rested and fatigued conditions. The inclusion of GRF data into the study would be expected to magnify the trends shown in Figs. 3.8 and 3.9. Previous research has shown that ITB forces are a factor in knee, tibia, and patellar kinematics [82] and further research is needed to determine the role of ITB forces in overuse injuries. If the ITB is truly a passive structure, then the present model should be adequate. Only changes in length should factor into force production in passive tissue, and the changes in ITB length due to the kinematics during stance are included in the data. The aspects that the model ignores are the active contractile forces of the ITB’s muscular insertions from the gluteus maximus and the tensor fascia lata, which would require GRFs to estimate.

The present study has demonstrated that the kinematics of the transverse plane, particularly at the knee and hip, are important considerations when investigating the biomechanics of an overuse injury such as ITBS. Motion in the transverse plane can be difficult to measure due to the subtlety of the rotations compared the sagittal and frontal planes, and previous research has often neglected transverse kinematics. Future investigations of the biomechanics of running injuries should take care to consider the transverse plane when seeking to explain the etiology of an injury.

The ITB is a complex structure with multiple bony attachment sites and insertions from muscles. Mechanically and functionally, it shares similar properties to both a tendon and a ligament, but is not an exact match for either. These combined factors make modeling the ITB a difficult task. Although its anatomy has been investigated in cadaver studies [27, 71], the functional anatomy of the ITB during running has yet to be well-described. The exact pathogenesis of ITBS is also a point of some debate.
Most investigators agree that inflammation of the ITB results from repetitive friction with the LFE [9, 12, 83]. However, Fairclough et al. [71] proposed the alternative theory that the lateral knee pain associated with ITBS is due to medial-lateral compression of a highly innervated fat pad between the ITB and LFE, and that the pain is actually due to proprioceptive pressure sensors in the fat pad rather than irritation of the ITB. Regardless of the true mechanism of injury, future models of ITBS would benefit from a more complete definition of ITB anatomy and its function during running.

Although the group running times were similar, it could be argued that the study was limited by allowing the participants to select their own pace and decide on their own when they had reached fatigue. Under these conditions, it is possible (and likely) that the participants reached different levels of fatigue. The degree of fatigue could have been standardized by first assessing the participants’ cardiovascular fitness or running ability, possibly by $\dot{V}O_2,max$ testing or a time trial, then having the runners maintain a running speed associated with a particular fraction of their thresholds for a set amount of time. However, since ITBS is an overuse injury, it is more likely developed during training rather than during a performance setting. Allowing the participants to run at a self-selected pace and to decide on their own when to stop does a better job of mimicking their normal training habits. Enforcing a standardized exhausting pace may have been awkward or unnatural, and may have produced data that were not representative of the conditions during which the ITBS runners developed their injuries. A non-standardized running pace is not necessarily a large source of error. Queen et al. [84] investigated the repeatability of lower extremity kinetics and kinematics between trials at a self-selected pace and at a standardized pace, and found greater repeatability when participants ran at their self-selected paces. Running speed did not have a significant effect on the repeatability of the data. The goal of the protocol was to allow the ITBS runners to demonstrate their preferred running mechanics, which could have been compromised if speed had been controlled by enforcing a standardized pace.
CHAPTER 5 MANUSCRIPT

This chapter contains a manuscript for an original research article, entitled "The role of fatigue in the biomechanics of iliotibial band syndrome," to be submitted to the journal Gait and Posture for review. The manuscript is formatted according to Gait and Posture submission specifications.
Lower extremity mechanics of iliotibial band syndrome during an exhaustive run

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ABSTRACT

Purpose: To measure changes in lower extremity mechanics during an exhaustive run in individuals with and without a history of iliobibial band syndrome (ITBS). Methods: Sixteen recreational runners ran to voluntary exhaustion on a treadmill at a self-selected pace. Eight runners had a history of ITBS. Twenty-three retro-reflective marker positions were recorded by an eight-camera 120 Hz motion capture system. Joint angles during the stance phase were exported to a musculoskeletal model (SIMM) with the iliobibial band (ITB) modeled as a passive structure to estimate tension and strain in the ITB. Results: For the ITBS runners, at the end of the run: (1) knee flexion at heel-strike was higher than control (20.6° vs. 15.3°, p = 0.01); (2) the number of knees with predicted ITB impingment upon the lateral femoral epicondyle increased from six to eleven; (3) tension in the ITB at heel-strike rose by 10.3% from the beginning of the run. Strain and tension in the ITB were higher in the ITBS runners throughout all of stance. Maximum foot adduction in the ITBS runners was higher vs. control at the start of the run (p = 0.003). Maximum foot inversion (p = 0.03) and maximum knee internal rotation velocity (p = 0.02) were higher vs. control at the end of the run. Conclusions: ITB mechanics appear to be related to changes in knee flexion at heel-strike and internal rotation of the leg. These observations may help to suggest kinematic discriminators for clinical assessment.

Key Words: ILIOTIBIAL BAND SYNDROME, RUNNING, STRAIN, TRANSVERSE KINEMATICS, FATIGUE
1. Introduction

Iliotibial band syndrome (ITBS) is an overuse injury of the lateral knee that affects up to 12% of all runners per year [1]. The iliotibial band (ITB) is a thick sheet of connective tissue that originates at the iliac crest and terminates at Gerdy’s tubercle and the fibular head. The ITB runs down the lateral aspect of the thigh and knee and passes over the lateral femoral epicondyle (LFE). ITBS develops when the ITB becomes inflamed due to excessive friction with the LFE as the two structures impinge on one another during running. Treatment of ITBS can require several months of conservative therapy, a reduction or complete cessation of running, or surgical resectioning of the ITB in serious cases [2]. Friction between the ITB and LFE can be elevated by factors that cause additional impingements, such as sudden increases in pace and mileage. Friction can also be elevated by factors that increase the tension in the ITB, such as hip abductor weakness and a lack of ITB flexibility [3-5].

While clinical investigations of ITBS have been numerous in the past, the biomechanics of ITBS have received comparatively little attention. Orchard et al. [6] performed a dynamic treadmill study of ITBS runners and found no differences in knee flexion angles between the runners' symptomatic and asymptomatic legs. Messier et al. [1] demonstrated that runners with a history of ITBS reduced rearfoot pronation during heel-strike when compared to runners in a control group. This finding was contrary to the usual assumption of excessive rearfoot motion as a mechanism of overuse injuries. The ITBS runners also showed faster maximum supination velocities, smaller maximum breaking forces, and were shorter and lighter than the control group.
Unfortunately, studies such as Orchard et al. [6] and Messier et al. [1] are rare and have not been further pursued. The present research is motivated by the lack of current biomechanical studies on ITBS. While previous research has shown significant associations between kinematic and kinetic variables and ITBS, these studies are few in number, and many important injury mechanisms are in need of further investigation. While ITBS is recognized as an overuse injury, the effects of fatigue on the etiology of ITBS have not been reported. Fatigue has long been suggested as a component of abnormal lower extremity mechanics and injuries [7-9]. Additionally, some researchers have suggested an association between excessive internal tibial rotation and lower extremity injuries in runners [10-12].

The goal of the present research was to expand the base of knowledge of ITBS biomechanics using both optical motion capture and mathematical modeling. The motion capture component measured the kinematics of runners with and without a history of ITBS during a run to voluntary exhaustion. It was hypothesized that with the onset of fatigue, runners with a history of ITBS will exhibit larger changes in internal tibial rotation and rearfoot motion than an age-matched control group. A mathematical model of the ITB and its associated anatomy will also be constructed to further investigate the effects of a fatigued run on the strain and tension in the ITB. It was hypothesized that both ITB strain and tension will be significantly higher in ITBS runners, and that these differences will become more distinct with fatigue.

2. Methods

2.1. Participants
Sixteen recreational runners were recruited for the study. Eight runners had a history of ITBS in one or both legs and comprised the ITBS group. Eight runners with no history of ITBS were age-matched to within four years of an ITBS runner and comprised the control group. All participants were symptom-free of any lower extremity injury for at least three months prior to participating in the study. Table 1 shows pertinent subject characteristics for both groups. The study was approved by the Institutional Review Board at Iowa State University. Participants gave informed written consent prior to participating in the study.

Participants completed a questionnaire that addressed their training regimen and injury history. ITB flexibility was assessed by the Ober test [13] on the left and right leg at 0° and 90° of knee flexion. Leg lengths were measured with a Gullick tape measure as the distance between the anterior-superior iliac spine and the ipsilateral medial malleolus.

2.2 Protocol

An eight-camera 120 Hz Peak Motus motion capture system (Vikon Peak, Centennial, CO) was used to capture three-dimensional optical data. The cameras were mounted on an octagon frame attached to the lab ceiling. Twenty-three retro-reflective markers (diameter 2 cm) were attached on both sides of the body at the toe, lateral dorsifoot, heel, medial and lateral malleoli, anterior calf, medial and lateral knee, anterior thigh, greater trochanter, anterior-superior iliac spine, and L5/S1.

Once the markers were placed, participants performed a static trial by standing in the center of the camera ring, facing the front of the room with their arms across their chest and feet shoulder width apart while marker positions were recorded for one second. The four medial markers were removed after completing the static trial.
Dynamic trials were next performed on an adjustable speed Quinton treadmill (Stairmaster, Kirkland, WA) set at a zero percent grade. Participants were instructed to select a running pace that would exhaust them within fifteen to twenty minutes. The selected pace was typically near a recent 5-km race pace and was maintained throughout the dynamic trial. Participants were allowed to warm up at a slower pace for two minutes before beginning the dynamic trial. Marker positions were recorded for ten seconds every two minutes until the participants reported voluntary exhaustion. Participants ran in their own shoes and wore a black spandex shirt and shorts. Figure 1 shows a participant during a dynamic trial.

2.3. Data Analysis

Joint angles for the hip, knee, and ankle and segment angles for the thigh, calf, and foot were calculated from the static and dynamic marker positions with a custom MATLAB program (MathWorks, Natick, MA). The program estimated joint centers and calculated neutral position joint angles based on the temporal averages of the static marker coordinates. Neutral position joint angles were calculated as three successive rotations: flexion/extension, adduction/abduction, and internal/external rotation.

For the dynamic trials, the four removed medial markers were reconstructed as virtual markers based on their relative positions during the static trial. Dynamic marker positions were filtered using a fourth-order low pass symmetric Butterworth filter with a cutoff frequency of 10 Hz. Dynamic joint and segment angles were calculated using the same rotation sequence as the static angles, but were performed at each time step rather than with a temporal average. To account for differences in anthropometrics and marker placement, the
neutral position angles were subtracted from the dynamic angles at each time step. The stance phase of each step for both the left and right legs were isolated by finding pairs of successive minimum vertical coordinates of the heel and toe markers. Each dynamic stance angle curve was interpolated to a 100-point spline. Ensemble curves for the dynamic stance angles were generated by averaging the splines for every step in a 10-s dynamic data block. Joint and segment angular velocities were calculated using the first-order central difference method.

2.4. Modeling

A musculoskeletal model of the lower extremity was created using SIMM 4.0 (MusculoGraphics, Santa Rosa, CA). The model included the pelvis/sacrum, femur, tibia/fibula, patella, talus, calcaneus, and metatarsal bones, along with forty-three muscles of the lower extremity. The ITB was included in the model by defining three additional tissue structures that originated at the iliac crest and terminated at the femoral insertion of the gluteus maximus, the fibular head, and Gerdy’s tubercle. The maximum isometric force of the ITB was estimated to be 860 N from cadaveric data [14]. Since the ITB was assumed to consist of only passive tissue, active force-length and force-velocity relationships were not considered.

Each participant’s ensemble joint angles for both the left and right legs were imported into SIMM motion files. Bone and muscle geometries were scaled to the participant’s leg length. Maximum isometric muscle forces were scaled to the participant’s weight. Impingement between the ITB and the LFE was modeled by defining a wrapping sphere whose surface was flush against the outer surface of the LFE. Impingement was defined as
instances during stance when the ITB was in contact with the LFE wrapping sphere. Figure 2 shows a sagittal view of the SIMM model and a close-up view of the lateral knee during impingement.

Strain in the ITB was calculated by dividing the change in length of the ITB by the length of the ITB when the model was in the anatomical position. Tension in the ITB was calculated by applying the passive force-length relationship shown in Figure 3 to the strain data. The force-length relationship in Figure 3 was created based on an ultimate tensile strength of 860 N (based on cadaveric data [14]) and the general force-length response for elastic tissues [15].

2.5. Statistical Analysis

Discriminant analyses were performed on the questionnaire responses, kinematics from the beginning and end of the dynamic trials, and the timings of kinematic maximums and minimums. Post-hoc t-tests were performed on the variables from the discriminant analyses that accounted for the most variability between the ITBS and control groups. The ITB strains and tensions were also compared between groups using t-tests. A difference between groups was considered significant at the $\alpha = 0.05$ level. SAS 9.1.3 (SAS Institute, Cary, NC) was used for statistical analysis.

3. Results

3.1. Questionnaire Responses and Measurements

Six of the eight ITBS runners had suffered from symptoms in both legs. One runner had symptoms only in the left leg, and one runner had symptoms only in the right leg. All
ITBS runners had been diagnosed by either a physical therapist or a medical doctor. Table 2 shows the reported training data and the results of the Ober testing. While none of the variables in Table 2 were significantly different, the ITBS runners showed trends to train at a higher weekly mileage and run the dynamic trials at a slower pace than the control group. On average, the ITBS runners were slightly less flexible than the control runners in all four Ober test measurements.

3.2. Dynamic Trial Kinematics

Kinematic angles and angular velocities during stance were compared between the ITBS and control groups. At the beginning of the runs, there were significant differences during the stance phase in maximum foot adduction (2.6 ± 7.9° for ITBS vs. -6.1 ± 7.6° for control, \( p = 0.003 \)), minimum knee flexion (12.5 ± 3.6° for ITBS vs. 7.7 ± 3.8° for control, \( p = 0.01 \)), and maximum ankle dorsiflexion velocity (74.8 ± 19.7°/s for ITBS vs. 54.9 ± 27.9° for control, \( p = 0.03 \)). At the end of the runs, there were significant differences during the stance phase in maximum knee flexion (43.8 ± 7.8° for ITBS vs. 36.5 ± 9.2° for control, \( p = 0.02 \)), maximum foot inversion (3.3 ± 10.9° for ITBS vs. -9.5 ± 19.6° for control, \( p = 0.03 \)), minimum thigh flexion velocity (-29.5 ± 16.9°/s for ITBS vs. -16.7 ± 8.2°/s for control, \( p = 0.01 \)), and maximum knee internal rotation velocity (16.4 ± 9.3°/s for ITBS vs. 10.3 ± 4.0°/s for control, \( p = 0.02 \)). Table 3 summarizes the significant kinematic results.

Figures 4 and 5 show the average knee flexion angle at heel-strike and the number of impinged knees at heel-strike, respectively. At the beginning of the run, the knee flexion angles were similar between the groups (17.2 ± 7.0° for ITBS vs. 15.2 ± 3.9° for control, \( p = 0.16 \)). At the end of the run, the ITBS knees were significantly more flexed than the control
knees (20.6 ± 7.8° for ITBS vs. 15.3 ± 5.2° for control, p = 0.01). The number of knees impinged at heel-strike increased over the course of the run for the ITBS group (from six to eleven) but decreased for the control group (from six to four). In all of the unimpinged knees, the ITB was located anteriorly to the LFE at heel-strike, and impingement had not yet occurred.

3.3. Strain and Tension

Figure 6 shows the strain in the ITB during the stance phase. At heel-strike, both groups had approximately 6% ITB strain at both the start and the end of the run. Strain in the ITBS runners rose quickly after heel-strike, and the ITBS runners demonstrated more strain than the control runners for the final 90% of stance. Fatigue did not have a large effect on strain in either group. However, the ITBS runners showed slightly less strain during the middle 50% and the final 15% of stance when fatigued.

Figure 7 shows the peak ITB strain during the stance phase. Peak strains for the ITBS runners was significantly higher than control at both the beginning and end of the run (beginning: 8.5 ± 1.2% for ITBS vs. 7.5 ± 0.7% for control, p = 0.005; end: 8.4 ± 1.3% for ITBS vs. 7.5 ± 0.8% for control, p = 0.03).

Figures 8 and 9 show the tension in the ITB during stance and the tension in the ITB at heel-strike, respectively. Although differences between groups were not significant due to the large variation in the tension between participants, some important trends were present. The ITB tension at heel-strike was similar for the control runners at both the beginning and end of the run, with tensions of 49.7 N and 49.6 N, respectively. The ITBS runners had 49.0 N of tension at the beginning of the run, but by the end of the run their tension at heel-strike
rose to 54.1 N, an increase of 10.3%. After the first 10% of stance, tension in the ITBS runners began to rise above the tension in the control runners. For the final 90% of stance, the ITBS runners demonstrated greater tension than the control runners. At the beginning of the run, the peak tension was 95.9 N in the ITBS runners and 73.5 N in the control runners. At the end of the run, the peak tension was 91.1 N in the ITBS runners and 74.7 N in the control runners. When looking at fatigue-induced changes in tension over the entire stance phase, the ITBS runners slightly increased their tension from 0-20% of stance and slightly decreased their tension from 20-70% of stance. The control runners did not demonstrate major changes in tension during the run.

4. Discussion

The major kinematic finding from the study was the increase in knee flexion and the corresponding increase in the number of impinged knees at heel-strike for the iliotibial band syndrome (ITBS) runners with the onset of fatigue. The increase in knee flexion at heel-strike is consistent with previous results on kinematic adjustments during a run to fatigue [16,17]. Greater knee flexion may also indicate a smaller calf angle at heel-strike, which has been associated with less efficient running economy [18].

Previous research has suggested that decreasing the angle of knee flexion at heel-strike predisposes a runner to developing ITBS by increasing the amount of time spent in the impingement range of knee flexion angles during stance [1,6]. The impingement range is defined as the range of knee flexion angles at which the ITB encounters friction with the LFE. The present study offers an alternative explanation that increasing knee flexion at heel-strike places a runner's iliotibial band (ITB) in a compromised orientation where
impingement with the lateral femoral epicondyle (LFE) during heel-strike is more likely to occur.

The hypothesis that ITBS runners would demonstrate increased strain in the ITB with fatigue was not supported. There was only a small (less than 0.5%) increase in strain for the ITBS runners during the first 10% of stance. However, the ITBS runners did demonstrate higher strain than the control runners throughout the entire stance phase, which was consistent with the original hypothesis. Under the assumption that higher strain places more stress on the ITB and puts the ITBS runners closer to their injury threshold, the strain data suggest that the ITBS runners were more likely to develop an overuse injury related to the ITB.

The increased strain in the ITBS runners was likely a function of the discriminating kinematic variables from the dynamic trials. Previous research [2], along with the present results, have suggested a zone or range of impingement for the ITB rather than a defined point. The minimum knee flexion angle was significantly larger for the ITBS runners under rested conditions, suggesting that their knees were closer to the impingement zone during stance than the control knees. This condition could potentially increase the stance-time spent in the impingement zone, which may lead to injury. Changes in strain from heel-strike to toe-off suggest that the ITB in the control runners released from impingement towards the end of stance while the ITB in the injured runners remained impinged. From Fig. 6, the control runners decreased their strain from 6% at heel-strike to 5% at toe-off, while the ITBS runners increased their strain from 6% at heel-strike to 7% at toe-off.

The present results supported the hypothesis that ITBS runners would increase the tension in the ITB as they fatigued, although tension only increased during the first 20% of
stance. Based on these results, an injury mechanism for ITBS is proposed: with fatigue, as the knee becomes more flexed at heel-strike, the odds for and degree of impingement between the ITB and LFE increase. Impingement at heel-strike, when impact forces are first transmitted through the soft tissues of the leg, acts to increase the immediate tension in the ITB, causing more irritation and pain. The ITBS runners increased their tension at heel-strike by 10.3% with the onset of fatigue, suggesting that the initial “shock” of impact and the immediate loading of the ITB play important roles in injury. The ITBS group’s limited flexibility (Table 2) may have also contributed to their ITB tension. At the beginning of the run, the ITBS runners had a noticeably higher loading rate following heel-strike and during the approach to maximum tension (see slopes of Fig. 8), lending further support to the idea that the condition of the ITB at heel-strike is an important factor in the injury mechanism.

Although the results suggest an important role for heel-strike impacts in the pathology of ITBS, the remainder of stance should not be neglected. Novacheck [19] notes that the active impact forces that occur during the latter three-fourths of stance are larger in magnitude and of longer duration than the passive impact forces that occur near heel-strike. Peak strain and tension in both groups occurred near mid-stance, regardless of fatigue. Additionally, the peak strain was significantly higher in the ITBS runners, although the peak tension was not significantly higher due to the large spread of the tension data. Group differences in strain and tension during the latter phases of stance were likely due to the kinematic differences discussed in the next section.

The long-term goal of overuse running injury research should be to reduce the frequency of overuse injuries. Nigg and Bobbert [20] noted that there is no indication that load analysis during running has contributed to a reduction in injury frequency. One possible
explanation is that injury research tends to be limited to retrospective studies of runners who are already injured, or were injured in the past but are now healthy. The knowledge base of injury biomechanics would benefit greatly from prospective investigations that follow a large group of runners over time and compare the mechanics of those who develop overuse injuries to those who do not. In a prospective study of four hundred previously healthy university students, those who developed lower extremity injuries during the study exhibited higher maximum foot eversion, more laterally-oriented roll-off of the foot, and a more centralized heel-strike center of pressure (COP) [21]. A more centralized COP could be partially accounted for by increased knee flexion at heel-strike.

The present research has suggested several variables that may predispose runners to developing ITBS, based on comparisons with healthy runners. Although the present research was retrospective, treatments of ITBS tend to focus on the symptoms of the injury rather than its causes. The ITBS runners may therefore have still exhibited the lower extremity kinematics that influenced their injury etiologies. Clinically, the results of this research could be used to adjust the kinematics of ITBS runners, possibly via orthotic inserts or gait retraining, to relieve symptoms and guard against future injury. Foot orthotics are capable of inducing significant changes in muscle EMG activity during running [22] and could potentially be used to relieve tension in the ITB.

The study was limited by the lack of kinetic data. Numerous authors have suggested a relationship between impact forces and overuse running injuries [23-26]. Without GRF data, it was not possible to realistically estimate the tension in the ITB during stance. The tensions estimated in this study were determined solely from the strain in the ITB and the passive force-length relationship shown in Fig. 3 and should not be interpreted as
physiologically realistic tensions due to the lack of GRF contributions. However, even without GRFs, the ITB tension estimates shown in Figs. 8 and 9 provide a good picture of the relative ITB tension between the ITBS runners and control runners, and the rested and fatigued conditions. The present model should be adequate if the ITB is a completely passive structure, since changes in force would be associated only with changes in ITB length that were identified kinematically. The inclusion of GRF data into the study would then be expected to magnify the trends shown in Figs. 8 and 9. However, the present model excluded the insertions of the gluteus maximus and tensor fascia lata into the ITB. The predicted force in these muscles would increase with the inclusion of GRFs and would be expected to increase tension in the ITB due to their insertions. These changes in muscle length were not included in the present model of the ITB. Previous research has shown that ITB forces are a factor in knee, tibia, and patellar kinematics [27] and further research is needed to determine the role of ITB forces in overuse injuries.

The present study has demonstrated that the kinematics of the transverse plane, particularly at the knee and hip, are important considerations when investigating the biomechanics of an overuse injury such as ITBS. Motion in the transverse plane can be difficult to measure due to the subtlety of the motions compared to motions in the sagittal and frontal planes, and previous research has often neglected transverse kinematics. Future investigations of the biomechanics of running injuries should take care to consider the transverse plane when seeking to explain the etiology of an injury.

Although the group running times were similar, it could be argued that the study was limited by allowing the participants to select their own pace and decide on their own when they had reached fatigue. Under these conditions, it is possible (and likely) that the
participants reached different levels of fatigue. The degree of fatigue could have been standardized by first assessing the participants' cardiovascular fitness or running ability, then having the runners maintain a running speed associated with a particular fraction of their threshold for a set amount of time. Fatigue could also be assessed post-run by measuring blood lactate. However, since ITBS is an overuse injury, it is more likely developed during training rather than during a performance setting. Allowing the participants to run at a self-selected pace and to decide on their own when to stop does a better job of mimicking their normal training habits. Enforcing a standardized exhausting pace may have been awkward or unnatural, and may have produced data that were not representative of the conditions during which the ITBS runners developed their injuries. A non-standardized running pace is not necessarily a large source of error. Queen [28] investigated the repeatability of lower extremity kinetics and kinematics between trials at a self-selected pace and at a standardized pace, and found greater repeatability when participants ran at their self-selected paces. Running speed did not have a significant effect on the repeatability of the data.

References


Table 1. Subject characteristics. Values are mean(SD). \( N = 8 \) for both groups

<table>
<thead>
<tr>
<th></th>
<th>Age (yrs)</th>
<th>Height (cm)</th>
<th>Weight (kg)</th>
</tr>
</thead>
<tbody>
<tr>
<td>ITBS</td>
<td>27.5(9.0)</td>
<td>170.1(6.9)</td>
<td>68.7(15.9)</td>
</tr>
<tr>
<td>Control</td>
<td>26.4(7.7)</td>
<td>172.8(8.9)</td>
<td>71.3(14.4)</td>
</tr>
</tbody>
</table>
Table 2. Training data responses, Ober test (OT) measurements, and dynamic trial pace and time. Values are mean(SD).

<table>
<thead>
<tr>
<th></th>
<th>ITBS</th>
<th>Control</th>
<th>p</th>
</tr>
</thead>
<tbody>
<tr>
<td>Weekly mileage</td>
<td>23.7(15.1)</td>
<td>11.8(5.9)</td>
<td>0.06</td>
</tr>
<tr>
<td>Months at mileage</td>
<td>6.0(3.2)</td>
<td>6.3(2.5)</td>
<td>0.87</td>
</tr>
<tr>
<td>5-km race time (min)</td>
<td>23.4(5.3)</td>
<td>22.9(2.6)</td>
<td>0.84</td>
</tr>
<tr>
<td>OT right leg, 0° (deg)</td>
<td>17.4(6.8)</td>
<td>19.9(11.5)</td>
<td>0.62</td>
</tr>
<tr>
<td>OT right leg, 90° (deg)</td>
<td>12.6(7.4)</td>
<td>13.9(11.7)</td>
<td>0.79</td>
</tr>
<tr>
<td>OT right leg, 0° (deg)</td>
<td>15.9(5.2)</td>
<td>17.0(8.4)</td>
<td>0.56</td>
</tr>
<tr>
<td>OT right leg, 90° (deg)</td>
<td>13.2(5.5)</td>
<td>15.2(7.1)</td>
<td>0.54</td>
</tr>
<tr>
<td>Trial time (min)</td>
<td>16.1(5.0)</td>
<td>16.3(6.0)</td>
<td>0.95</td>
</tr>
<tr>
<td>Trial pace (min/mi)</td>
<td>9.5(2.2)</td>
<td>7.8(1.0)</td>
<td>0.08</td>
</tr>
</tbody>
</table>
Table 3. Significant kinematic differences ($p < 0.05$) between the ITBS and control groups. Values are mean(SD). Angles in degrees. Velocities in degrees per second.

<table>
<thead>
<tr>
<th></th>
<th>ITBS</th>
<th>Control</th>
<th>$p$</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Beginning of run</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Maximum foot adduction</td>
<td>2.6(7.9)</td>
<td>-6.1(7.6)</td>
<td>0.003</td>
</tr>
<tr>
<td>Minimum knee flexion</td>
<td>12.5(3.6)</td>
<td>7.7(3.8)</td>
<td>0.01</td>
</tr>
<tr>
<td>Maximum ankle dorsiflexion velocity</td>
<td>74.8(19.7)</td>
<td>54.9(27.9)</td>
<td>0.03</td>
</tr>
<tr>
<td><strong>End of run</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Maximum knee flexion</td>
<td>43.8(7.5)</td>
<td>36.5(9.2)</td>
<td>0.02</td>
</tr>
<tr>
<td>Maximum foot inversion</td>
<td>3.3(10.9)</td>
<td>-9.5(19.6)</td>
<td>0.03</td>
</tr>
<tr>
<td>Minimum thigh flexion velocity</td>
<td>-29.5(16.9)</td>
<td>-16.7(8.2)</td>
<td>0.01</td>
</tr>
<tr>
<td>Maximum knee internal rotation velocity</td>
<td>16.4(9.3)</td>
<td>10.3(4.0)</td>
<td>0.02</td>
</tr>
</tbody>
</table>
Figure 1.  A participant during the dynamic trial data collection.

Figure 2.  (a) The musculoskeletal model and (b) a close-up of the model’s knee during impingement.

Figure 3.  Passive force-length curve used to calculate ITB tension.

Figure 4.  Knee flexion at heel-strike.  * = significantly greater than fatigued control (p = 0.03).

Figure 5.  Number of knees with impinged IT bands at heel-strike.

Figure 6.  Strain in the ITB during the stance phase.  0% represents heel-strike and 100% represents toe-off.
Red/black lines represent the ITBS/control groups; solid/dashes lines represent the beginning/end of the run.

Figure 7.  Peak ITB strain during stance.  # = significantly higher than rested control (p = 0.03).  * =
significantly higher than fatigued control (p = 0.005).

Figure 8.  Tension in the ITB during the stance phase.  0% represents heel-strike and 100% represents toe-off.
Red/black lines represent the ITBS/control groups; solid/dashes lines represent the beginning/end of the run.

Figure 9.  Tension in the ITB at heel-strike.
Figure 2b

Figure 3

Normalized ultimate force

Normalized length
Figure 6

Figure 7
Figure 8

Figure 9
APPENDIX A  EXTENDED RESULTS

Appendix A presents the joint angles during stance at the hip, knee, and ankle in all three planes of motion. Canadian Society of Biomechanics definitions are followed. In all of the following figures, gray lines represent the ITBS group, black lines represent the control group, solid lines represent rested trials, and dotted lines represent fatigued trials.

Figure A.1  Hip flexion during stance for rested and fatigued conditions.
Figure A.2  Hip adduction during stance for rested and fatigued conditions.

Figure A.3  Hip internal rotation during stance for rested and fatigued conditions.
Figure A.4  Knee flexion during stance for rested and fatigued conditions.

Figure A.5  Knee adduction during stance for rested and fatigued conditions.
Figure A.6  Knee internal rotation during stance for rested and fatigued conditions.

Figure A.7  Ankle dorsiflexion during stance for rested and fatigued conditions.
Figure A.8 Ankle inversion during stance for rested and fatigued conditions.

Figure A.9 Ankle adduction during stance for rested and fatigued conditions.
Appendix B contains the informed consent document that participants read and signed before participating in the study.
INFORMED CONSENT DOCUMENT

Title of Study: The relationship between tibial / femoral rotation and rear foot motion as it relates to the causes of iliotibial band friction syndrome.

Investigators: Jennifer Lowry and Jason Gillette, Ph.D.

This is a research study. Take time to decide whether or not you would like to participate. Please ask any questions at any time.

INTRODUCTION

The purpose of this research study is to determine whether tibial / femoral rotation is related to rear foot motion. Tibial / femoral rotation is directly related to the movement of the iliotibial band. The information in this study may provide us with a quantitative methodology for finding risk factors that are related to iliotibial band friction syndrome. The instruments that will be used to observe tibial / femoral rotation and rear foot motion include an eight-camera video tracking system and an electrogoniometer. The video tracking system will specifically measure rotation of the tibia relative to the femur. The electrogoniometer will specifically measure the rear foot angle.

This research may benefit runners, especially those that experience symptoms of iliotibial band friction syndrome. You are invited to participate in this study because you are eighteen years or older, and you are not currently suffering from injuries other than iliotibial band friction syndrome in the lower extremity.

DESCRIPTION OF PROCEDURES

If you agree to take part in this study, your participation will last approximately one hour. The study will take place in the Biomechanics Laboratory in room 178N of the Forker Building on the Iowa State University campus. The study will follow a specific order of procedures. First, you will complete a questionnaire asking for your age, injury history, and details of your running training. Your height, weight, leg length, hip width, and knee width will be measured and recorded. The tightness of your iliotibial band will be measured using a procedure called Ober’s Test. You will lie on your side while a tester flexes your hip and knee and measures the angle your thigh makes with the ground. Reflective video markers will be placed on your knee and leg using double-sided tape. The location of the reflective markers will then be recorded by the eight camera video system. An electrogoniometer will be attached behind your ankle. This will measure how your foot rotates during running. You will be asked to run on a treadmill at a level grade level similar to running on flat ground (not uphill or downhill). You will be asked to warm-up for four minutes of running, during which the speed of the treadmill will be adjusted until you feel that it matches your preferred running speed for a mile run. During the fatigue test, you will be asked to run until you feel that the selected treadmill pace can no longer be maintained or 30 minutes has passed.
RISKS

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

1. Minor discomfort due to skin irritation from the tape on the reflective video markers. The markers will be on the skin for as short of a period of time as possible in order to reduce the chances of possible discomfort.
2. Minor discomfort while your iliotibial band stretches during Ober’s Test.
3. Muscle soreness and/or tightness may result from muscle fatigue due to running. We advise stretching, and possibly icing specific muscles, before and after running to avoid this minor discomfort.

BENEFITS

If you decide to participate in this study, there will be no direct benefits given to you. You will hopefully gain a better understanding of iliotibial band friction syndrome (ITBS) and what motions cause the symptoms associated with ITBS.

ALTERNATIVES TO PARTICIPATION

The only alternative is to not participate in this study.

COSTS AND COMPENSATION

There are no costs nor any compensation for participating in this study.

PARTICIPATION RIGHTS

Your participation in this study is completely voluntary and you may refuse to participate or leave the study at any time. If you decide not to participate in the study, or leave the study early, you will not be penalized or experience any loss of benefits to which you are otherwise entitled.

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 approval forms to the State Appeals Board and are available from the Iowa State University Office of Risk Management and Insurance.

CONFIDENTIALITY
The records containing subject information will be kept confidential to the extent permitted by applicable laws and regulations and will not be made available to public. However, the federal government regulatory agencies, and the Institutional Review Board, a committee that reviews and approves human subject research studies, may inspect and/or copy your records for quality assurance and data analysis. These records may contain private information.

To ensure confidentiality to the extent permitted by this law, the following measures will be carried out. Using alphanumeric identifiers unrelated to subject names, subject data will be kept confidential. Subject name and information/data will be kept in separate and secure locations. The researchers will keep any recorded information private to the extent allowed by law. Once results of this study are reported, combined information will be included in a presentation. If results are published, subject identity will be kept confidential.

QUESTIONS OR PROBLEMS

Please ask any questions during any time of the study. For additional questions, contact Jason Gillette at (515) 294-8310. If you have any questions about the rights of research subjects or research-related injury, please contact the Human Subjects Research Office, 1138 Pearson Hall, (515) 294-4566; austingr@iastate.edu or the Research Compliance Officer, Office of Research Compliance, 1138 Pearson Hall, (515) 294-3115; dament@iastate.edu
SUBJECT SIGNATURE

Your signature indicates that voluntarily agree to participate in this study, you have been properly informed of the procedures involved in this study, you have taken time to read the informed consent document, and that you have had all questions adequately answered. You will receive a copy of the signed and dated consent form prior to your participation in the study.

Subject’s Name (Printed) ____________________________________________

Subject’s Signature____________________________________ Date___________

INVESTIGATOR’S SIGNATURE

I signify that the participant has had adequate time to understand the procedures involved in this study, has adequately taken time to read the consent document, and has had all questions answered to satisfaction. I feel that the participant adequately understands the purpose, risk, benefit and procedures of this study and has voluntarily agreed to participate.

Signature of Person Obtaining Informed Consent Form _______________________

Date______________
Appendix C contains the questionnaire that participants completed as part of their involvement in the study.
# ITBS Participant Questionnaire

<table>
<thead>
<tr>
<th>Subject ID</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>Date</td>
<td></td>
</tr>
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</table>

## Subject Information

<table>
<thead>
<tr>
<th>Height (cm)</th>
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<tbody>
<tr>
<td>Weight (kg)</td>
<td></td>
</tr>
<tr>
<td>Age (years)</td>
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</table>

<table>
<thead>
<tr>
<th>Sex</th>
<th>Male</th>
<th>Female</th>
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<tbody>
<tr>
<td></td>
<td></td>
<td></td>
</tr>
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</table>

<table>
<thead>
<tr>
<th>Running intensity*</th>
<th>Recreational</th>
<th>Competitive</th>
<th>Rec. / Comp.</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
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</table>

<table>
<thead>
<tr>
<th>Mileage per week</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>Months at current mileage</td>
<td></td>
</tr>
<tr>
<td>5k time (min:sec)</td>
<td></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>History of ITBFS?</th>
<th>Yes</th>
<th>No</th>
</tr>
</thead>
<tbody>
<tr>
<td>Symptomatic side</td>
<td>Left</td>
<td>Right</td>
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<table>
<thead>
<tr>
<th>Diagnosed by</th>
<th>Dr.</th>
<th>PT</th>
<th>Other:</th>
<th>None</th>
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</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
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</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Other running injuries?</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>Pain pattern**</td>
<td>1</td>
</tr>
</tbody>
</table>

## Flexibility

<table>
<thead>
<tr>
<th>0° Ober’s Test (deg.)</th>
<th>Right:</th>
<th>Left:</th>
</tr>
</thead>
<tbody>
<tr>
<td>90° Ober’s Test (deg.)</td>
<td>Right:</td>
<td>Left:</td>
</tr>
</tbody>
</table>

## Anthropometrics

<table>
<thead>
<tr>
<th>Leg length (cm)</th>
<th>Right:</th>
<th>Left:</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hip width (cm)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Knee width (cm)</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

* Definitions for intensity:
  - Recreational – runs for fitness or enjoyment only; not concerned with performance
  - Competitive – regularly competes in timed races; trains specifically for performance
  - Rec. / Comp. – recreational but trains for a few races a year; monitors performance

** Definitions for pain pattern:
  - 1 – Both during and after running
  - 2 – Primarily after running
  - 3 – Primarily while running
  - 4 – Primarily while running when fatigued
Appendix D contains the MATLAB code used to calculate kinematic angles and velocities from the optical marker data. The code was written to run in MATLAB 7.0.4 [52] and also requires the MATLAB Signal Processing Toolbox [56]. The code requires three subprograms:

1. ITBS_markers – assigns marker names to the data

2. ITBS_output – outputs the kinematics to a Microsoft Excel file

3. lmin – finds the local minima of a curve

Variables and abbreviations in the program:

R          right leg
L          left leg
ang        angles
St         static
Dy         dynamic
rel        relative to
lat        lateral side
med        medial side
MST        marks heel-strike and toe-off of stance
EC         ensemble curve
Program reads in 3D marker coordinates for a runner on a treadmill and calculates angles of the hip, knee, ankle, thigh, calf, and foot. Velocities are calculated by central-differencing.

Authors: Ross Miller and Jason Gillette
Department of Health and Human Performance
Iowa State University

Created: October 11, 2005
Last updated: January 19, 2006

clear all;
fs = 120; % Sampling frequency (Hz)
dt = 1/fs; % Time step (s)
tDy = 10; % Dynamic trial collection time (s)
time = 0:1/fs:(tDy - 1/fs); % Time for dynamic trials (s)

for sub = 1:16
% Static trial
cond = 0;
trial = 1;
path = ['/Users/rosshm/ITBS Data/','j',num2str(sub),'/'];
file = [path,'j',num2str(sub),'c',num2str(cond),'
    num2str(trial),'.31d'];
data = dlmread(file);

% Filter data with a 4th-order, low-pass symmetric Butterworth filter
order = 4;
f = 10;
Wn = 2*f/fs;
[butter_b,butter_a] = butter(order,Wn);
data = filtfilt(butter_b,butter_a,data);

ITBS_markers; % Subprogram to assign marker names to data

% Coordinate axes for the static foot triads
R_foottri_origin = 1/3*(R_heel + R_lat_ankle + R_df);
R_foottri_St(1,:) = R_df - R_lat_ankle;
R_foottri_St(1,:) = R_foottri_St(1,:) / norm(R_foottri_St(1,:));
R_foottri_St(3,:) = cross(R_df - R_heel,R_foottri_St(1,:));
R_foottri_St(3,:) = R_foottri_St(3,:) / norm(R_foottri_St(3,:));
R_foottri_St(2,:) = cross(R_foottri_St(3,:),R_foottri_St(1,:));
L_foottri_origin = 1/3*(L_heel + L_lat_ankle + L_df);
L_foottri_St(1,:) = L_df - L_lat_ankle;
L_foottri_St(1,:) = L_foottri_St(1,:) / norm(L_foottri_St(1,:));
L_foottri_St(3,:) = cross(L_df - L_heel,L_foottri_St(1,:));
L_foottri_St(3,:) = L_foottri_St(3,:) / norm(L_foottri_St(3,:));
% Coordinate axes for the static calf triads
R_calftri_origin = 1/3*(R_lat_ankle + R_ant_leg + R_lat_knee);
R_calftri_St(3,:) = R_lat_knee - R_lat_ankle;
R_calftri_St(3,:) = R_calftri_St(3,:) / norm(R_calftri_St(3,:));
R_calftri_St(1,:) = cross(R_ant_leg - R_lat_ankle, R_calftri_St(3,:));
R_calftri_St(1,:) = R_calftri_St(1,:) / norm(R_calftri_St(1,:));
R_calftri_St(2,:) = cross(R_calftri_St(3,:),R_calftri_St(1,:));
L_calftri_origin = 1/3*(L_lat_ankle + L_ant_leg + L_lat_knee);
L_calftri_St(3,:) = L_lat_knee - L_lat_ankle;
L_calftri_St(3,:) = L_calftri_St(3,:) / norm(L_calftri_St(3,:));
L_calftri_St(1,:) = cross(L_ant_leg - L_lat_ankle, L_calftri_St(3,:));
L_calftri_St(1,:) = L_calftri_St(1,:) / norm(L_calftri_St(1,:));
L_calftri_St(2,:) = cross(L_calftri_St(3,:),L_calftri_St(1,:));

% Coordinate axes for the static torso
asis_c = 0.5*(R_asis + L_asis);
torsotri_origin = 1/3*(R_asis + L_asis + L5S1);
torsotri_St(2,:) = L_asis - R_asis;
torsotri_St(2,:) = torsotri_St(2,:) / norm(torsotri_St(2,:));
torsotri_St(3,:) = cross(asis_c - L5S1,torsotri_St(2,:));
torsotri_St(3,:) = torsotri_St(3,:) / norm(torsotri_St(3,:));
torsotri_St(1,:) = cross(torsotri_St(2,:),torsotri_St(3,:));

% Static marker positions relative to segment triads
R_med_ankte_ret = R_med_ankte - R_calftri_origin;
L_med_ankte_ret = L_med_ankte - L_calftri_origin;
R_med_knee_ret = R_med_knee - R_calftri_origin;
L_med_knee_ret = L_med_knee - L_calftri_origin;
R_df_d_ret = R_df_d - R_foottri_origin;
L_df_d_ret = L_df_d - L_foottri_origin;

% Hip joint locations relative to ASIS markers
Width_asis = sqrt((R_asis(1) - L_asis(1))^2 + (R_asis(2) - L_asis(2))^2 + (R_asis(3) - L_asis(3))^2);
R_hip_rel_asis = [-0.19*Width_asis -0.14*Width_asis -0.30*Width_asis];
L_hip_rel_asis = [-0.19*Width_asis 0.14*Width_asis -0.30*Width_asis];

% Calculate joint centers
R_ankle = 0.5*(R_med_ankle + R_lat_ankle);
L_ankle = 0.5*(L_med_ankle + L_lat_ankle);
R_knee = 0.5*(R_med_knee + R_lat_knee);
L_knee = 0.5*(L_med_knee + L_lat_knee);
R_hip_c = 0.75*R_hip + 0.25*L_hip;
L_hip_c = 0.75*L_hip + 0.25*R_hip;
% Coordinate axes for the static torso
torso_origin = (R_asis + L_asis + L5S1) / 3;
torso_St(2,:) = L_asis - R_asis;
torso_St(2,:) = torso_St(2,:) / norm(torso_St(2,:));
torso_St(3,:) = cross(0.5*(R_asis + L_asis) - L5S1,torso_St(2,:));
torso_St(3,:) = torso_St(3,:) / norm(torso_St(3,:));
torso_St(1,:) = cross(torso_St(2,:),torso_St(3,:));

% Coordinate axes for the static foot segments
R_foot_St(1,:) = R_df_d - R_ankle;
R_foot_St(1,:) = R_foot_St(1,:) / norm(R_foot_St(1,:));
R_foot_St(3,:) = cross(R_foot_St(1,:),R_med_ankle - R_lat_ankle);
R_foot_St(3,:) = R_foot_St(3,:) / norm(R_foot_St(3,:));
R_foot_St(2,:) = cross(R_foot_St(3,:),R_foot_St(1,:));
L_foot_St(1,:) = L_df_d - L_ankle;
L_foot_St(1,:) = L_foot_St(1,:) / norm(L_foot_St(1,:));
L_foot_St(3,:) = cross(L_foot_St(1,:),L_lat_ankle - L_med_ankle);
L_foot_St(3,:) = L_foot_St(3,:) / norm(L_foot_St(3,:));
L_foot_St(2,:) = cross(L_foot_St(3,:),L_foot_St(1,:));

% Coordinate axes for the static calf segments
R_calf_St(3,:) = R_knee - R_ankle;
R_calf_St(3,:) = R_calf_St(3,:) / norm(R_calf_St(3,:));
R_calf_St(1,:) = cross(R_med_knee - R_lat_knee,R_calf_St(3,:));
R_calf_St(1,:) = R_calf_St(1,:) / norm(R_calf_St(1,:));
R_calf_St(2,:) = cross(R_calf_St(3,:),R_calf_St(1,:));
L_calf_St(3,:) = L_knee - L_ankle;
L_calf_St(3,:) = L_calf_St(3,:) / norm(L_calf_St(3,:));
L_calf_St(1,:) = cross(L_lat_knee - L_med_knee,L_calf_St(3,:));
L_calf_St(1,:) = L_calf_St(1,:) / norm(L_calf_St(1,:));
L_calf_St(2,:) = cross(L_calf_St(3,:),L_calf_St(1,:));

% Coordinate axes for the static thigh segments
R_thigh_St(3,:) = R_hip_c - R_knee;
R_thigh_St(3,:) = R_thigh_St(3,:) / norm(R_thigh_St(3,:));
R_thigh_St(1,:) = cross(L_hip_c - R_hip_c,R_thigh_St(3,:));
R_thigh_St(1,:) = R_thigh_St(1,:) / norm(R_thigh_St(1,:));
R_thigh_St(2,:) = cross(R_thigh_St(3,:),R_thigh_St(1,:));
L_thigh_St(3,:) = L_hip_c - L_knee;
L_thigh_St(3,:) = L_thigh_St(3,:) / norm(L_thigh_St(3,:));
L_thigh_St(1,:) = cross(L_hip_c - R_hip_c,L_thigh_St(3,:));
L_thigh_St(1,:) = L_thigh_St(1,:) / norm(L_thigh_St(1,:));
L_thigh_St(2,:) = cross(L_thigh_St(3,:),L_thigh_St(1,:));

% Calculate static hip angles
S = torso_St*R_thigh_St';
kp = 1/(S(3,2) - 1);
\[ \text{S} = \text{torso}_\text{St} \times \text{L} \text{thigh}_\text{St}'; \]
\[ \text{kp} = 1/(\text{-S}(3,2) - 1); \]
\[ \text{kn} = 1/(\text{-S}(3,2) + 1); \]
\[ \text{R} \text{hip} \_ \text{angle}_\text{St}(::,1) = 0.5 \times \text{atan2}((\text{kp} \times (\text{S}(2,1) + \text{S}(1,3))), \ldots \]
\[ \text{R} \text{hip} \_ \text{angle}_\text{St}(::,2) = \text{atan2}(\text{-S}(3,2), \ldots \]
\[ \text{R} \text{hip} \_ \text{angle}_\text{St}(::,3) = 0.5 \times \text{atan2}(\text{kn} \times (\text{S}(2,1) - \text{S}(1,3)), \ldots \]
\[ \text{S} = \text{R} \text{thigh}_\text{St} \times \text{R} \text{ calf}_\text{St}'; \]
\[ \text{kp} = 1/(\text{-S}(3,2) - 1); \]
\[ \text{kn} = 1/(\text{-S}(3,2) + 1); \]
\[ \text{R} \text{knee} \_ \text{angle}_\text{St}(::,1) = 0.5 \times \text{atan2}((\text{kp} \times (\text{S}(2,1) + \text{S}(1,3))), \ldots \]
\[ \text{R} \text{knee} \_ \text{angle}_\text{St}(::,2) = \text{atan2}(\text{-S}(3,2), \ldots \]
\[ \text{R} \text{knee} \_ \text{angle}_\text{St}(::,3) = 0.5 \times \text{atan2}(\text{kn} \times (\text{S}(2,1) - \text{S}(1,3)), \ldots \]
\[ \text{S} = \text{L} \text{thigh}_\text{St} \times \text{L} \text{ calf}_\text{St}'; \]
\[ \text{kp} = 1/(\text{-S}(3,2) - 1); \]
\[ \text{kn} = 1/(\text{-S}(3,2) + 1); \]
\[ \text{L} \text{knee} \_ \text{angle}_\text{St}(::,1) = 0.5 \times \text{atan2}((\text{kp} \times (\text{S}(2,1) + \text{S}(1,3))), \ldots \]
\[ \text{L} \text{knee} \_ \text{angle}_\text{St}(::,2) = \text{atan2}(\text{-S}(3,2), \ldots \]
\[ \text{L} \text{knee} \_ \text{angle}_\text{St}(::,3) = 0.5 \times \text{atan2}(\text{kn} \times (\text{S}(2,1) - \text{S}(1,3)), \ldots \]
\[ L_{\text{knee\_angle\_St}(;3)} = 0.5 \times \text{atan2}(\text{kp} \times (S(2,1) + S(1,3)), (kp \times (S(2,3) - S(1,1))))\] 
\[ - 0.5 \times \text{atan2}(\text{kn} \times (S(2,1) - S(1,3)), (kn \times (S(2,3) + S(1,1))))\]

\% Calculate static ankle angles
\[ S = \text{R\_calf\_St} \times \text{R\_foot\_St'}; \]
\[ \text{kp} = 1/(-S(3,2) - 1); \]
\[ \text{kn} = 1/(-S(3,2) + 1); \]
\[ R_{\text{ankle\_angle\_St}(;1)} = 0.5 \times \text{atan2}(\text{kp} \times (S(2,1) + S(1,3)), \]
\[ (kp \times (S(2,3) - S(1,1)))) + \]
\[ 0.5 \times \text{atan2}(\text{kn} \times (S(2,1) - S(1,3)), \]
\[ (kn \times (S(2,3) + S(1,1))))\]
\[ R_{\text{ankle\_angle\_St}(;2)} = \text{atan2}(-S(3,2), \]
\[ (S(3,3)/\cos(R_{\text{ankle\_angle\_St}(;1)})))); \]
\[ R_{\text{ankle\_angle\_St}(;3)} = 0.5 \times \text{atan2}(\text{kp} \times (S(2,1) + S(1,3)), \]
\[ (kp \times (S(2,3) - S(1,1)))) - \]
\[ 0.5 \times \text{atan2}(\text{kn} \times (S(2,1) - S(1,3)), \]
\[ (kn \times (S(2,3) + S(1,1))))\]

\% Calculate static thigh segment angles
\[ S = \text{R\_thigh\_St}; \]
\[ \text{kp} = 1/(-S(3,2) - 1); \]
\[ \text{kn} = 1/(-S(3,2) + 1); \]
\[ R_{\text{thigh\_angle\_St}(;1)} = 0.5 \times \text{atan2}(\text{kp} \times (S(2,1) + S(1,3)), \]
\[ (kp \times (S(2,3) - S(1,1)))) + \]
\[ 0.5 \times \text{atan2}(\text{kn} \times (S(2,1) - S(1,3)), \]
\[ (kn \times (S(2,3) + S(1,1))))\]
\[ R_{\text{thigh\_angle\_St}(;2)} = \text{atan2}(-S(3,2), \]
\[ (S(3,3)/\cos(R_{\text{thigh\_angle\_St}(;1)})))); \]
\[ R_{\text{thigh\_angle\_St}(;3)} = 0.5 \times \text{atan2}(\text{kp} \times (S(2,1) + S(1,3)), \]
\[ (kp \times (S(2,3) - S(1,1)))) - \]
\[ 0.5 \times \text{atan2}(\text{kn} \times (S(2,1) - S(1,3)), \]
\[ ... \]
\[ S = \text{L\_thigh\_St}; \]
\[ \text{kp} = 1/(-\text{S}(3,2) - 1); \]
\[ \text{kn} = 1/(-\text{S}(3,2) + 1); \]
\[ \text{L\_thigh\_angle\_St(:,1)} = 0.5\text{atan2}(\text{kp}*(\text{S}(2,1) + \text{S}(1,3)), \ldots \]
\[ (\text{kp}*(\text{S}(2,3) - \text{S}(1,1)))) + \ldots \]
\[ 0.5\text{atan2}(\text{kn}*(\text{S}(2,1) - \text{S}(1,3))), \ldots \]
\[ (\text{kn}*(\text{S}(2,3) + \text{S}(1,1)))); \]
\[ \text{L\_thigh\_angle\_St(:,2)} = \text{atan2}(-\text{S}(3,2), \ldots \]
\[ (\text{S}(3,3)/\cos(\text{L\_thigh\_angle\_St(:,1))}); \]
\[ \text{L\_thigh\_angle\_St(:,3)} = 0.5\text{atan2}(\text{kp}*(\text{S}(2,1) + \text{S}(1,3)), \ldots \]
\[ (\text{kp}*(\text{S}(2,3) - \text{S}(1,1)))) - \ldots \]
\[ 0.5\text{atan2}(\text{kn}*(\text{S}(2,1) - \text{S}(1,3))), \ldots \]
\[ (\text{kn}*(\text{S}(2,3) + \text{S}(1,1)))); \]

\% Calculate static calf segment angles
\[ S = \text{R\_calf\_St}; \]
\[ \text{kp} = 1/(-\text{S}(3,2) - 1); \]
\[ \text{kn} = 1/(-\text{S}(3,2) + 1); \]
\[ \text{R\_calf\_angle\_St(:,1)} = 0.5\text{atan2}(\text{kp}*(\text{S}(2,1) + \text{S}(1,3)), \ldots \]
\[ (\text{kp}*(\text{S}(2,3) - \text{S}(1,1)))) + \ldots \]
\[ 0.5\text{atan2}(\text{kn}*(\text{S}(2,1) - \text{S}(1,3))), \ldots \]
\[ (\text{kn}*(\text{S}(2,3) + \text{S}(1,1)))); \]
\[ \text{R\_calf\_angle\_St(:,2)} = \text{atan2}(-\text{S}(3,2), \ldots \]
\[ (\text{S}(3,3)/\cos(\text{R\_calf\_angle\_St(:,1))}); \]
\[ \text{R\_calf\_angle\_St(:,3)} = 0.5\text{atan2}(\text{kp}*(\text{S}(2,1) + \text{S}(1,3)), \ldots \]
\[ (\text{kp}*(\text{S}(2,3) - \text{S}(1,1)))) - \ldots \]
\[ 0.5\text{atan2}(\text{kn}*(\text{S}(2,1) - \text{S}(1,3))), \ldots \]
\[ (\text{kn}*(\text{S}(2,3) + \text{S}(1,1)))); \]

\[ S = \text{L\_calf\_St}; \]
\[ \text{kp} = 1/(-\text{S}(3,2) - 1); \]
\[ \text{kn} = 1/(-\text{S}(3,2) + 1); \]
\[ \text{L\_calf\_angle\_St(:,1)} = 0.5\text{atan2}(\text{kp}*(\text{S}(2,1) + \text{S}(1,3)), \ldots \]
\[ (\text{kp}*(\text{S}(2,3) - \text{S}(1,1)))) + \ldots \]
\[ 0.5\text{atan2}(\text{kn}*(\text{S}(2,1) - \text{S}(1,3))), \ldots \]
\[ (\text{kn}*(\text{S}(2,3) + \text{S}(1,1)))); \]
\[ \text{L\_calf\_angle\_St(:,2)} = \text{atan2}(-\text{S}(3,2), \ldots \]
\[ (\text{S}(3,3)/\cos(\text{L\_calf\_angle\_St(:,1))}); \]
\[ \text{L\_calf\_angle\_St(:,3)} = 0.5\text{atan2}(\text{kp}*(\text{S}(2,1) + \text{S}(1,3)), \ldots \]
\[ (\text{kp}*(\text{S}(2,3) - \text{S}(1,1)))) - \ldots \]
\[ 0.5\text{atan2}(\text{kn}*(\text{S}(2,1) - \text{S}(1,3))), \ldots \]
\[ (\text{kn}*(\text{S}(2,3) + \text{S}(1,1)))); \]

\% Calculate static foot segment angles
\[ S = \text{R\_foot\_St}; \]
\[ \text{kp} = 1/(-\text{S}(3,2) - 1); \]
\[ \text{kn} = 1/(-\text{S}(3,2) + 1); \]
R_foot_angle_St(:,1) = 0.5*atan2(kp*(S(2,1) + S(1,3)), ... 
    (kp*(S(2,3) - S(1,1))))) + ... 
    0.5*atan2(kn*(S(2,1) - S(1,3)), ... 
    (kn*(S(2,3) + S(1,1))));
R_foot_angle_St(:,2) = atan2(-S(3,2), ... 
    (S(3,3)/cos(R_foot_angle_St(:,1))));
R_foot_angle_St(:,3) = 0.5*atan2(kp*(S(2,1) + S(1,3)), ... 
    (kp*(S(2,3) - S(1,1))))) - ... 
    0.5*atan2(kn*(S(2,1) - S(1,3)), ... 
    (kn*(S(2,3) + S(1,1))));
S = L_foot_St;
kp = 1/(-S(3,2) - 1);
kn = 1/(-S(3,2) + 1);
L_foot_angle_St(:,1) = 0.5*atan2(kp*(S(2,1) + S(1,3)), ... 
    (kp*(S(2,3) - S(1,1))))) + ... 
    0.5*atan2(kn*(S(2,1) - S(1,3)), ... 
    (kn*(S(2,3) + S(1,1))));
L_foot_angle_St(:,2) = atan2(-S(3,2), ... 
    (S(3,3)/cos(L_foot_angle_St(:,1))));
L_foot_angle_St(:,3) = 0.5*atan2(kp*(S(2,1) + S(1,3)), ... 
    (kp*(S(2,3) - S(1,1))))) - ... 
    0.5*atan2(kn*(S(2,1) - S(1,3)), ... 
    (kn*(S(2,3) + S(1,1))));

% Dynamic trials
cond = 1;
for trial = 1:2 % Trial 1 = rested, Trial 2 = fatigued
    clear R_toe R_df R_heel R_lat_anke L_ant_leg R_lat_knee;
    clear R_ant_thigh R_hip R_asis L_toe L_df L_heel L_lat_anke;
    clear L_ant_leg L_lat_knee L_ant_thigh L_hip L_asis L5S1;
    clear R_df_d L_df_d asis_c;
    file = [path,'j',num2str(sub),'c',num2str(cond),'t', ... 
        num2str(trial),'31d'];
    data = dlmread(file);
    % Filter with a 4th-order, low-pass symmetric Butterworth filter
    order = 4;
    fc = 10;
    Wn = 2*fc/fs;
    [butter_b,butter_a] = butter(order,Wn);
    data = filtfilt(butter_b,butter_a,data);
    ITBS_markers;
    for i = 1:fs*tDy
% Coordinate axes for the dynamic foot triads
R_foottri_origin = 1/3*(R_heel(i,:) + R_lat_ankle(i,:) + ...
  R_df(i,:));
R_foottri_Dy(1,:) = R_df(i,:) - R_lat_ankle(i,:);
R_foottri_Dy(1,:) = R_foottri_Dy(1,:) / ...
  norm(R_foottri_Dy(1,:));
R_foottri_Dy(3,:) = cross(R_df(i,:),R_foottri_Dy(1,:));
R_foottri_Dy(3,:) = R_foottri_Dy(3,:) / ...
  norm(R_foottri_Dy(3,:));
R_foottri_Dy(2,:) = cross(R_foottri_Dy(3,:), ...
  R_foottri_Dy(1,:));
L_foottri_origin = 1/3*(L_heel(i,:) + L_lat_ankle(i,:) + ...
  L_df(i,:));
L_foottri_Dy(1,:) = L_df(i,:) - L_lat_ankle(i,:);
L_foottri_Dy(1,:) = L_foottri_Dy(1,:) / ...
  norm(L_foottri_Dy(1,:));
L_foottri_Dy(3,:) = cross(L_df(i,:),L_foottri_Dy(1,:));
L_foottri_Dy(3,:) = L_foottri_Dy(3,:) / ...
  norm(L_foottri_Dy(3,:));
L_foottri_Dy(2,:) = cross(L_foottri_Dy(3,:), ...
  L_foottri_Dy(1,:));
%
% Coordinate axes for the dynamic calf triads
R_calftri_origin = 1/3*(R_lat_ankle(i,:) + ...
  R_ant_leg(i,:) + R_lat_knee(i,:));
R_calftri_Dy(3,:) = R_lat_knee(i,:) - R_lat_ankle(i,:);
R_calftri_Dy(3,:) = R_calftri_Dy(3,:) / ...
  norm(R_calftri_Dy(3,:));
R_calftri_Dy(1,:) = cross(R_ant_leg(i,:),R_calftri_Dy(3,:));
R_calftri_Dy(1,:) = R_calftri_Dy(1,:) / ...
  norm(R_calftri_Dy(1,:));
R_calftri_Dy(2,:) = cross(R_calftri_Dy(3,:), ...
  R_calftri_Dy(1,:));
L_calftri_origin = 1/3*(L_lat_ankle(i,:) + ...
  L_ant_leg(i,:) + L_lat_knee(i,:));
L_calftri_Dy(3,:) = L_lat_knee(i,:) - L_lat_ankle(i,:);
L_calftri_Dy(3,:) = L_calftri_Dy(3,:) / ...
  norm(L_calftri_Dy(3,:));
L_calftri_Dy(1,:) = cross(L_ant_leg(i,:),L_calftri_Dy(3,:));
L_calftri_Dy(1,:) = L_calftri_Dy(1,:) / ...
  norm(L_calftri_Dy(1,:));
L_calftri_Dy(2,:) = cross(L_calftri_Dy(3,:), ...
  L_calftri_Dy(1,:));
% Coordinate axes for the dynamic torso
asis_c(i,:) = 0.5*(R_asis(i,:) + L_asis(i,:));
torso_origin = (R_asis(i,:) + L_asis(i,:) + L5S1(i,:)) / 3;
torso_Dy(2,:) = L_asis(i,:) - R_asis(i,:);
torso_Dy(2,:) = torso_Dy(2,:) / norm(torso_Dy(2,:));
torso_Dy(3,:) = cross(asis_c(i,:),torso_Dy(2,:));
torso_Dy(3,:) = torso_Dy(3,:) / norm(torso_Dy(3,:));
torso_Dy(1,:) = cross(torso_Dy(2,:),torso_Dy(3,:));

% Re-construct pulled markers
R_med_ankle(i,:) = R_med_ankle_origtn + ...
R_med_ankle_rel*R_med_ankle_St'*R_med_ankle_Dy;
L_med_ankle(i,:) = L_med_ankle_origtn + ...
L_med_ankle_rel*L_med_ankle_St'*L_med_ankle_Dy;
R_med_knee(i,:) = R_med_knee_origtn + ...
R_med_knee_rel*R_med_ankle_St'*R_med_ankle_Dy;
L_med_knee(i,:) = L_med_knee_origtn + ...
L_med_knee_rel*L_med_ankle_St'*L_med_ankle_Dy;
R_df_d(i,:) = R_foottri_origtn + ...
R_df_d_rel*R_foottri_St'*R_foottri_Dy;
L_df_d(i,:) = L_foottri_origtn + ...
L_df_d_rel*L_foottri_St'*L_foottri_Dy;

% Calculate joint centers
R_ankle(i,:) = 0.5*(R_lat_ankle(i,:) + R_med_ankle(i,:));
L_ankle(i,:) = 0.5*(L_lat_ankle(i,:) + L_med_ankle(i,:));
R_knee(i,:) = 0.5*(R_lat_knee(i,:) + R_med_knee(i,:));
L_knee(i,:) = 0.5*(L_lat_knee(i,:) + L_med_knee(i,:));
R_hip_c(i,:) = 0.75*R_hip(i,:) + 0.25*L_hip(i,:);
L_hip_c(i,:) = 0.25*R_hip(i,:) + 0.75*L_hip(i,:);

% Coordinate axes for the dynamic foot segments
R_foot_Dy(1,:) = R_df_d(i,:) - R_ankle(i,:);
R_foot_Dy(1,:) = R_foot_Dy(1,:) / norm(R_foot_Dy(1,:));
R_foot_Dy(3,:) = cross(R_foot_Dy(1,:),R_med_ankle(i,:)) - ...
R_lat_ankle(i,:));
R_foot_Dy(3,:) = R_foot_Dy(3,:) / norm(R_foot_Dy(3,:));
R_foot_Dy(2,:) = cross(R_foot_Dy(3,:),R_foot_Dy(1,:));
L_foot_Dy(1,:) = L_df_d(i,:) - L_ankle(i,:);
L_foot_Dy(1,:) = L_foot_Dy(1,:) / norm(L_foot_Dy(1,:));
L_foot_Dy(3,:) = cross(L_foot_Dy(1,:),L_lat_ankle(i,:)) - ...
L_med_ankle(i,:));
L_foot_Dy(3,:) = L_foot_Dy(3,:) / norm(L_foot_Dy(3,:));
L_foot_Dy(2,:) = cross(L_foot_Dy(3,:),L_foot_Dy(1,:));

% Coordinate axes for the dynamic calf segments
% Coordinate axes for the dynamic thigh segments
R_thigh_Dy(3,:) = R_thigh(i,:) - R_knee(i,:);
R_thigh_Dy(3,:) = R_thigh_Dy(3,:) / norm(R_thigh_Dy(3,:));
R_thigh_Dy(1,:) = cross(R_med_knee(i,:), R_thigh_Dy(3,:));
R_thigh_Dy(1,:) = cross(R_thigh_Dy(3,:), R_thigh_Dy(1,:));
L_thigh_Dy(3,:) = R_knee(i,:) - L_ankle(i,:);
L_thigh_Dy(3,:) = L_thigh_Dy(3,:) / norm(L_thigh_Dy(3,:));
L_thigh_Dy(1,:) = cross(L_lat_knee(i,:), L_thigh_Dy(3,:));
L_thigh_Dy(1,:) = cross(L_thigh_Dy(3,:), L_thigh_Dy(1,:));

% Calculate hip angles
kp = 1/(-S(3,2) - 1);
kn = 1/(-S(3,2) + 1);
R_hip_angle(i,1) = 0.5*atan2(kp*(S(2,1) + S(1,3)), ...
                (kp*(S(2,3) - S(1,1)))) + ...
                0.5*atan2(kn*(S(2,1) - S(1,3)), ...
                (kn*(S(2,3) + S(1,1))));

R_hip_angle(i,2) = atan2(-S(3,2), ...
                (S(3,3)/cos(R_hip_angle(i,1))));
R_hip_angle(i,3) = 0.5*atan2(kp*(S(2,1) + S(1,3)), ...
                (kp*(S(2,3) - S(1,1)))) - ...
                0.5*atan2(kn*(S(2,1) - S(1,3)), ...
                (kn*(S(2,3) + S(1,1))));

S = torso_Dy*R_thigh_Dy';
kp = 1/(-S(3,2) - 1);
kn = 1/(-S(3,2) + 1);
L_hip_angle(i,1) = 0.5*atan2(kp*(S(2,1) + S(1,3)), ...
                (kp*(S(2,3) - S(1,1)))) + ...
                0.5*atan2(kn*(S(2,1) - S(1,3)), ...)
\[
L_{\text{hip angle}}(i,2) = \text{atan2}(-S(3,2), (S(3,3)/\cos(L_{\text{hip angle}}(i,1))));
\]
\[
L_{\text{hip angle}}(i,3) = 0.5\times\text{atan2}(kp*(S(2,1) + S(1,3)), (kp*(S(2,3) - S(1,1)))) - 0.5\times\text{atan2}(kn*(S(2,1) - S(1,3)), (kn*(S(2,3) + S(1,1))));
\]

\% Calculate knee angles
\[
S = R_{\text{thigh Dy}}R_{\text{calf Dy}}';
\]
kp = 1/(S(3,2) - 1);
kn = 1/(S(3,2) + 1);
\[
R_{\text{knee angle}}(i,1) = 0.5\times\text{atan2}(kp*(S(2,1) + S(1,3)), (kp*(S(2,3) - S(1,1)))) + 0.5\times\text{atan2}(kn*(S(2,1) - S(1,3)), (kn*(S(2,3) + S(1,1))));
\]
\[
R_{\text{knee angle}}(i,2) = \text{atan2}(-S(3,2), (S(3,3)/\cos(R_{\text{knee angle}}(i,1))));
\]
\[
R_{\text{knee angle}}(i,3) = 0.5\times\text{atan2}(kp*(S(2,1) + S(1,3)), (kp*(S(2,3) - S(1,1)))) - 0.5\times\text{atan2}(kn*(S(2,1) - S(1,3)), (kn*(S(2,3) + S(1,1))));
\]

\% Calculate ankle angles
\[
S = L_{\text{thigh Dy}}L_{\text{calf Dy}}';
\]
kp = 1/(S(3,2) - 1);
kn = 1/(S(3,2) + 1);
\[
L_{\text{knee angle}}(i,1) = 0.5\times\text{atan2}(kp*(S(2,1) + S(1,3)), (kp*(S(2,3) - S(1,1)))) + 0.5\times\text{atan2}(kn*(S(2,1) - S(1,3)), (kn*(S(2,3) + S(1,1))));
\]
\[
L_{\text{knee angle}}(i,2) = \text{atan2}(-S(3,2), (S(3,3)/\cos(L_{\text{knee angle}}(i,1))));
\]
\[
L_{\text{knee angle}}(i,3) = 0.5\times\text{atan2}(kp*(S(2,1) + S(1,3)), (kp*(S(2,3) - S(1,1)))) - 0.5\times\text{atan2}(kn*(S(2,1) - S(1,3)), (kn*(S(2,3) + S(1,1))));
\]

\% Calculate ankle angles
\[
S = R_{\text{calf Dy}}R_{\text{foot Dy}}';
\]
kp = 1/(S(3,2) - 1);
kn = 1/(S(3,2) + 1);
\[
R_{\text{ankle angle}}(i,1) = 0.5\times\text{atan2}(kp*(S(2,1) + S(1,3)), (kp*(S(2,3) - S(1,1)))) + 0.5\times\text{atan2}(kn*(S(2,1) - S(1,3)), (kn*(S(2,3) + S(1,1))));
\]
\[
R_{\text{ankle angle}}(i,2) = \text{atan2}(-S(3,2), (S(3,3)/\cos(R_{\text{ankle angle}}(i,1))));
\]
\[
R_{\text{ankle angle}}(i,3) = 0.5\times\text{atan2}(kp*(S(2,1) + S(1,3)), (kp*(S(2,3) - S(1,1)))) - 0.5\times\text{atan2}(kn*(S(2,1) - S(1,3)), (kn*(S(2,3) + S(1,1))));
\]
\( S = L_{\text{calf}_{-}\text{Dy}}*L_{\text{foot}_{-}\text{Dy}'}; \)
\( kp = 1/(-5(3,2) - 1); \)
\( kn = 1/(-5(3,2) + 1); \)
\[
L_{\text{ankle}_{-}\text{angle}}(i,1) = 0.5*\text{atan2}(kp*(S(2,3) - S(1,3)), ... \)
\( (kp*(S(2,3) - S(1,3)))) + ... \)
\( 0.5*\text{atan2}(kn*(S(2,1) - 1), ... \)
\( (kn*(S(2,3) + 1))); \)
\[
L_{\text{ankle}_{-}\text{angle}}(i,2) = \text{atan2}(-S(3,2), ... \)
\( (S(3,3)/\cos(L_{\text{ankle}_{-}\text{angle}}(i,1))))); \)
\[
L_{\text{ankle}_{-}\text{angle}}(i,3) = 0.5*\text{atan2}(kp*(S(2,3) - S(1,3)), ... \)
\( (kp*(S(2,3) - S(1,3)))) - ... \)
\( 0.5*\text{atan2}(kn*(S(2,1) - S(1,3)), ... \)
\( (kn*(S(2,3) + S(1,3))))); \)
\[
\% \text{Calculate thigh segment angles} \)
\( S = R_{\text{thigh}_{-}\text{Dy}}; \)
\( kp = 1/(-5(3,2) - 1); \)
\( kn = 1/(-5(3,2) + 1); \)
\[
R_{\text{thigh}_{-}\text{angle}}(i,1) = 0.5*\text{atan2}(kp*(S(2,3) - S(1,3)), ... \)
\( (kp*(S(2,3) - S(1,3)))) + ... \)
\( 0.5*\text{atan2}(kn*(S(2,1) - S(1,3)), ... \)
\( (kn*(S(2,3) + S(1,3))))); \)
\[
R_{\text{thigh}_{-}\text{angle}}(i,2) = \text{atan2}(-S(3,2), ... \)
\( (S(3,3)/\cos(R_{\text{thigh}_{-}\text{angle}}(i,1))))); \)
\[
R_{\text{thigh}_{-}\text{angle}}(i,3) = 0.5*\text{atan2}(kp*(S(2,3) - S(1,3)), ... \)
\( (kp*(S(2,3) - S(1,1)))) - ... \)
\( 0.5*\text{atan2}(kn*(S(2,1) - S(1,3)), ... \)
\( (kn*(S(2,3) + S(1,3))))); \)
\[
S = L_{\text{thigh}_{-}\text{Dy}}; \)
\( kp = 1/(-5(3,2) - 1); \)
\( kn = 1/(-5(3,2) + 1); \)
\[
L_{\text{thigh}_{-}\text{angle}}(i,1) = 0.5*\text{atan2}(kp*(S(2,3) - S(1,3)), ... \)
\( (kp*(S(2,3) - S(1,1)))) + ... \)
\( 0.5*\text{atan2}(kn*(S(2,1) - S(1,3)), ... \)
\( (kn*(S(2,3) + S(1,1))))); \)
\[
L_{\text{thigh}_{-}\text{angle}}(i,2) = \text{atan2}(-S(3,2), ... \)
\( (S(3,3)/\cos(L_{\text{thigh}_{-}\text{angle}}(i,1))))); \)
\[
L_{\text{thigh}_{-}\text{angle}}(i,3) = 0.5*\text{atan2}(kp*(S(2,3) - S(1,3)), ... \)
\( (kp*(S(2,3) - S(1,1)))) - ... \)
\( 0.5*\text{atan2}(kn*(S(2,1) - S(1,3)), ... \)
\( (kn*(S(2,3) + S(1,3))))); \)
\[
\% \text{Calculate calf segment angles} \)
\( S = R_{\text{calf}_{-}\text{Dy}}; \)
\[ R_{\text{calf\_angle}}(i,1) = 0.5\text{atan2}(kp*(S(2,1) + S(1,3)), (kp*(S(2,3) - S(1,1)))) + ... \\
\]
\[ 0.5\text{atan2}(kn*(S(2,1) - S(1,3)), (kn*(S(2,3) + S(1,1)))); \]

\[ R_{\text{calf\_angle}}(i,2) = \text{atan2}(-S(3,2), ... (S(3,3)/cos(R_{\text{calf\_angle}}(i,1))))); \]

\[ R_{\text{calf\_angle}}(i,3) = 0.5\text{atan2}(kp*(S(2,1) + S(1,3)), ... (kp*(S(2,3) - S(1,1)))) - ... \\
\]
\[ 0.5\text{atan2}(kn*(S(2,1) - S(1,3)), (kn*(S(2,3) + S(1,1))))); \]

\[ S = L_{\text{calf\_Dy}}; \]

\[ kp = 1/(-S(3,2) - 1); \]
\[ kn = 1/(-S(3,2) + 1); \]

\[ L_{\text{calf\_angle}}(i,1) = 0.5\text{atan2}(kp*(S(2,1) + S(1,3)), ... (kp*(S(2,3) - S(1,1)))) + ... \\
\]
\[ 0.5\text{atan2}(kn*(S(2,1) - S(1,3)), (kn*(S(2,3) + S(1,1))))); \]

\[ L_{\text{calf\_angle}}(i,2) = \text{atan2}(-S(3,2), ... (S(3,3)/cos(L_{\text{calf\_angle}}(i,1))))); \]

\[ L_{\text{calf\_angle}}(i,3) = 0.5\text{atan2}(kp*(S(2,1) + S(1,3)), ... (kp*(S(2,3) - S(1,1)))) - ... \\
\]
\[ 0.5\text{atan2}(kn*(S(2,1) - S(1,3)), (kn*(S(2,3) + S(1,1))))); \]

% Calculate foot segment angles
\[ S = R_{\text{foot\_Dy}}; \]

\[ kp = 1/(-S(3,2) - 1); \]
\[ kn = 1/(-S(3,2) + 1); \]

\[ R_{\text{foot\_angle}}(i,1) = 0.5\text{atan2}(kp*(S(2,1) + S(1,3)), ... (kp*(S(2,3) - S(1,1)))) + ... \\
\]
\[ 0.5\text{atan2}(kn*(S(2,1) - S(1,3)), (kn*(S(2,3) + S(1,1))))); \]

\[ R_{\text{foot\_angle}}(i,2) = \text{atan2}(-S(3,2), ... (S(3,3)/cos(R_{\text{foot\_angle}}(i,1))))); \]

\[ R_{\text{foot\_angle}}(i,3) = 0.5\text{atan2}(kp*(S(2,1) + S(1,3)), ... (kp*(S(2,3) - S(1,1)))) - ... \\
\]
\[ 0.5\text{atan2}(kn*(S(2,1) - S(1,3)), (kn*(S(2,3) + S(1,1))))); \]

\[ S = L_{\text{foot\_Dy}}; \]

\[ kp = 1/(-S(3,2) - 1); \]
\[ kn = 1/(-S(3,2) + 1); \]

\[ L_{\text{foot\_angle}}(i,1) = 0.5\text{atan2}(kp*(S(2,1) + S(1,3)), ... (kp*(S(2,3) - S(1,1)))) + ... \\
\]
\[ 0.5\text{atan2}(kn*(S(2,1) - S(1,3)), (kn*(S(2,3) + S(1,1))))); \]
L_foot_angle(i,2) = atan2(-S(3,2), ... 
(S(3,3)/cos(L_foot_angle(i,1)))
)
L_foot_angle(i,3) = 0.5*atan2(kp*(S(2,1) + S(1,3)), ... 
(kp*(S(2,3) - S(1,1))) - ... 
0.5*atan2(kn*(S(2,1) - S(1,3)), ... 
(kn*(S(2,3) + S(1,1)))
);
end % for i

% Smooth out hip flexion angles to find stride lengths
order = 4;
fc ~= 2;
Wn = 2*fc/fs;
[butter_b,butter_a] = butter(order,Wn);
R_hip_angle_filt = filtfilt(butter_b,butter_a,R_hip_angle(:,1));
L_hip_angle_filt = filtfilt(butter_b,butter_a,L_hip_angle(:,1));

% Find hip flexion minimums and average index lengths b/w them
[junk1,hipflexmins_R] = lmin(R_hip_angle_filt,3);
[junk2,hipflexmins_L] = lmin(L_hip_angle_filt,3);
for i = 1:length(hipflexmins_R)-1
    stride_length_R(i) = hipflexmins_R(i+1) - hipflexmins_R(i);
end
for i = 1:length(hipflexmins_L)-1
    stride_length_L(i) = hipflexmins_L(i+1) - hipflexmins_L(i);
end

SL_R = mean(stride_length_R); % Time steps b/w right strides
SL_L = mean(stride_length_L); % Time steps b/w left strides

% Find stance phases for the right leg
stc_R = zeros(int16(1200/SL_R),2);
flag = 0;
[junk3(1,1),stc_R(1,1)] = min(R_heel(1:SL_R+10,3));
[junk3(1,2),stc_R(1,2)] = min(R_toe(1:SL_R+10,3));
i = 2;
while i < int16(1200/SL_R) & flag == 0
    i = int16(i);
    SL_R = int16(SL_R);
    jmin_HS_R = stc_R(i-1,1) + SL_R/8;
    jmax_HS_R = jmin_HS_R + SL_R;
    jmin_TO_R = stc_R(i-1,2) + SL_R/8;
    jmax_TO_R = jmin_TO_R + SL_R;
    if jmax_HS_R > 1200 | jmax_TO_R > 1200
        break;
    end
    [junk3(i,1),stc_R(i,1)] = min(R_heel(jmin_HS_R:jmax_HS_R,3));
end
[junk3(i,2), stc_R(i,2)] = min(R_toe(jmin_TO_R:jmax_TO_R,3));
stc_R(i,1) = stc_R(i,1) + jmin_HS_R;
stc_R(i,2) = stc_R(i,2) + jmin_TO_R;
i = i + 1;
end

% Find stance phases for the left leg
stc_L = zeros(int16(1200/SL_L), 2);
flag = 0;
[junk4(1,1), stc_L(1,1)] = min(L_heel(1:SL_L+10, 3));
[junk4(1,2), stc_L(1,2)] = min(L_toe(1:SL_L+10, 3));
i = 2;
while i < int16(1200/SL_L) & flag == 0
    i = int16(i);
    SL_L = int16(SL_L);
    jmin_HS_L = stc_L(i-1,1) + SL_L/8;
    jmax_HS_L = jmin_HS_L + SL_L;
    jmin_TO_L = stc_L(i-1,2) + SL_L/8;
    jmax_TO_L = jmin_TO_L + SL_L;
    if jmax_HS_L > 1200 || jmax_TO_L > 1200
        break;
    end
    [junk4(i,1), stc_L(i,1)] = ...
        min(L_heel(jmin_HS_L:jmax_HS_L, 3));
    [junk4(i,2), stc_L(i,2)] = ...
        min(L_toe(jmin_TO_L:jmax_TO_L, 3));
    stc_L(i,1) = stc_L(i,1) + jmin_HS_L;
    stc_L(i,2) = stc_L(i,2) + jmin_TO_L;
i = i + 1;
end

% Account for a toe-off occurring before a heel-strike at the
% front
% of the data. Shifts toe-off indices up one row and chops off
% the last row.
if stc_R(1,1) > stc_R(1,2)
    for i = 1:(length(stc_R)-1)
        stc_R(i,2) = stc_R(i+1,2);
    end
    stc_R = stc_R(1:(length(stc_R)-1),:);
end

if stc_L(1,1) > stc_L(1,2)
    for i = 1:(length(stc_L)-1)
        stc_L(i,2) = stc_L(i+1,2);
    end
    stc_L = stc_L(1:(length(stc_L)-1),:);
% Account for multiple local minima in the stride search window
count_R = 0;
for j = 1:2
   for i = 1:(length(stc_R)-1)
      if stc_R(i+1,j) - stc_R(i,j) < 5
         for k = i:(length(stc_R)-1)
            stc_R(k,j) = stc_R(k+1,j);
         end
         count_R = count_R + 1;
      end
   end
end
stc_R = stc_R(1:(length(stc_R) - count_R),:);

count_L = 0;
for j = 1:2
   for i = 1:(length(stc_L)-1)
      if stc_L(i+1,j) - stc_L(i,j) < 5
         for k = i:(length(stc_L)-1)
            stc_L(k,j) = stc_L(k+1,j);
         end
         count_L = count_L + 1;
      end
   end
end
stc_L = stc_L(1:(length(stc_L) - count_L),:);

% Check for arrays that are not full
if stc_R(length(stc_R),1) == 0 || stc_R(length(stc_R),2) == 0
   stc_R = stc_R(1:(length(stc_R)-1),:);
end
if stc_L(length(stc_L),1) == 0 || stc_L(length(stc_L),2) == 0
   stc_L = stc_L(1:(length(stc_L)-1),:);
end
if subj == 2
   if trial == last_trial
      stc_L(4,2) = stc_L(5,2);
      for i = 5:10
         stc_L(i,2) = stc_L(i+1,2);
      end
      stc_L = stc_L(1:10,:);
   end
end
Fit a 100-point spline to each kinematic curve
for i = 1:length(stc_R)
    HS = stc_R(i,1);
    TO = stc_R(i,2);
    stc_length = TO - HS;
    x = 0:stc_length;
    xx = 0:stc_length/99:stc_length;

    y = R_hip_angle(HS:TO,1)';
    R_hip_flex(1:100,i) = spline(x,y,xx);
    y = R_hip_angle(HS:TO,2)';
    R_hip_add(1:100,i) = spline(x,y,xx);
    y = R_hip_angle(HS:TO,3)';
    R_hip_inrot(1:100,i) = spline(x,y,xx);

    y = R_knee_angle(HS:TO,1)';
    R_knee_flex(1:100,i) = spline(x,y,xx);
    y = R_knee_angle(HS:TO,2)';
    R_knee_add(1:100,i) = spline(x,y,xx);
    y = R_knee_angle(HS:TO,3)';
    R_knee_inrot(1:100,i) = spline(x,y,xx);

    y = R_ankle_angle(HS:TO,1)';
    R_ankle_DF(1:100,i) = spline(x,y,xx);
    y = R_ankle_angle(HS:TO,2)';
    R_ankle_add(1:100,i) = spline(x,y,xx);
    y = R_ankle_angle(HS:TO,3)';
    R_ankle_inrot(1:100,i) = spline(x,y,xx);

    y = R_thigh_angle(HS:TO,1)';
    R_thigh_flex(1:100,i) = spline(x,y,xx);
    y = R_thigh_angle(HS:TO,2)';
    R_thigh_add(1:100,i) = spline(x,y,xx);
    y = R_thigh_angle(HS:TO,3)';
    R_thigh_inrot(1:100,i) = spline(x,y,xx);

    y = R_calf_angle(HS:TO,1)';
    R_calf_flex(1:100,i) = spline(x,y,xx);
    y = R_calf_angle(HS:TO,2)';
    R_calf_add(1:100,i) = spline(x,y,xx);
    y = R_calf_angle(HS:TO,3)';
    R_calf_inrot(1:100,i) = spline(x,y,xx);

    y = R_foot_angle(HS:TO,1)';
    R_foot_flex(1:100,i) = spline(x,y,xx);
    y = R_foot_angle(HS:TO,2)';
R_foot_add(1:100,i) = spline(x,y,xx);
y = R_foot_angle(HS:T0,3)';
R_foot_inrot(1:100,i) = spline(x,y,xx);
end

for i = 1:length(stc_L)
    HS = stc_L(i,1);
    TO = stc_L(i,2);
    stc_length = TO - HS;
    x = 0:stc_length;
    xx = 0:stc_length/99:stc_length;

    y = L_hip_angle(HS:T0,1)';
    L_hip_flex(1:100,i) = spline(x,y,xx);
    y = L_hip_angle(HS:T0,2)';
    L_hip_add(1:100,i) = spline(x,y,xx);
    y = L_hip_angle(HS:T0,3)';
    L_hip_inrot(1:100,i) = spline(x,y,xx);

    y = L_knee_angle(HS:T0,1)';
    L_knee_flex(1:100,i) = spline(x,y,xx);
    y = L_knee_angle(HS:T0,2)';
    L_knee_add(1:100,i) = spline(x,y,xx);
    y = L_knee_angle(HS:T0,3)';
    L_knee_inrot(1:100,i) = spline(x,y,xx);

    y = L_ankle_angle(HS:T0,1)';
    L_ankle_DF(1:100,i) = spline(x,y,xx);
    y = L_ankle_angle(HS:T0,2)';
    L_ankle_add(1:100,i) = spline(x,y,xx);
    y = L_ankle_angle(HS:T0,3)';
    L_ankle_inrot(1:100,i) = spline(x,y,xx);

    y = L_thigh_angle(HS:T0,1)';
    L_thigh_flex(1:100,i) = spline(x,y,xx);
    y = L_thigh_angle(HS:T0,2)';
    L_thigh_add(1:100,i) = spline(x,y,xx);
    y = L_thigh_angle(HS:T0,3)';
    L_thigh_inrot(1:100,i) = spline(x,y,xx);

    y = L_calf_angle(HS:T0,1)';
    L_calf_flex(1:100,i) = spline(x,y,xx);
    y = L_calf_angle(HS:T0,2)';
    L_calf_add(1:100,i) = spline(x,y,xx);
    y = L_calf_angle(HS:T0,3)';
    L_calf_inrot(1:100,i) = spline(x,y,xx);
\[ y = \text{L\_foot\_angle}(\text{HS}\colon T0,1); \]
\[ \text{L\_foot\_flex}(1:100,i) = \text{spline}(x,y,xx); \]
\[ y = \text{L\_foot\_angle}(\text{HS}\colon T0,2); \]
\[ \text{L\_foot\_add}(1:100,i) = \text{spline}(x,y,xx); \]
\[ y = \text{L\_foot\_angle}(\text{HS}\colon T0,3); \]
\[ \text{L\_foot\_inrot}(1:100,i) = \text{spline}(x,y,xx); \]

\% Subtract static trial angles
for \( i = 1:\text{length}(\text{stc\_R}) \)
\[
\begin{align*}
    \text{R\_hip\_ang}(i,1) &= \text{R\_hip\_flex}(i) - \text{R\_hip\_angle\_St}(1,1); \\
    \text{R\_hip\_ang}(i,2) &= \text{R\_hip\_add}(i) - \text{R\_hip\_angle\_St}(1,2); \\
    \text{R\_hip\_ang}(i,3) &= \text{R\_hip\_inrot}(i) - \text{R\_hip\_angle\_St}(1,3); \\
    \text{R\_knee\_ang}(i,1) &= \text{R\_knee\_flex}(i) - \text{R\_knee\_angle\_St}(1,1); \\
    \text{R\_knee\_ang}(i,2) &= \text{R\_knee\_add}(i) - \text{R\_knee\_angle\_St}(1,2); \\
    \text{R\_knee\_ang}(i,3) &= \text{R\_knee\_inrot}(i) - \text{R\_knee\_angle\_St}(1,3); \\
    \text{R\_ankle\_ang}(i,1) &= \text{R\_ankle\_DF}(i) - \text{R\_ankle\_angle\_St}(1,1); \\
    \text{R\_ankle\_ang}(i,2) &= \text{R\_ankle\_add}(i) - \text{R\_ankle\_angle\_St}(1,2); \\
    \text{R\_ankle\_ang}(i,3) &= \text{R\_ankle\_inrot}(i) - \text{R\_ankle\_angle\_St}(1,3); \\
\end{align*}
\]
end

for \( i = 1:\text{length}(\text{stc\_L}) \)
\[
\begin{align*}
    \text{L\_hip\_ang}(i,1) &= \text{L\_hip\_flex}(i) - \text{L\_hip\_angle\_St}(1,1); \\
    \text{L\_hip\_ang}(i,2) &= \text{L\_hip\_add}(i) - \text{L\_hip\_angle\_St}(1,2); \\
    \text{L\_hip\_ang}(i,3) &= \text{L\_hip\_inrot}(i) - \text{L\_hip\_angle\_St}(1,3); \\
    \text{L\_knee\_ang}(i,1) &= \text{L\_knee\_flex}(i) - \text{L\_knee\_angle\_St}(1,1); \\
    \text{L\_knee\_ang}(i,2) &= \text{L\_knee\_add}(i) - \text{L\_knee\_angle\_St}(1,2); \\
    \text{L\_knee\_ang}(i,3) &= \text{L\_knee\_inrot}(i) - \text{L\_knee\_angle\_St}(1,3); \\
    \text{L\_ankle\_ang}(i,1) &= \text{L\_ankle\_DF}(i) - \text{L\_ankle\_angle\_St}(1,1); \\
    \text{L\_ankle\_ang}(i,2) &= \text{L\_ankle\_add}(i) - \text{L\_ankle\_angle\_St}(1,2); \\
    \text{L\_ankle\_ang}(i,3) &= \text{L\_ankle\_inrot}(i) - \text{L\_ankle\_angle\_St}(1,3); \\
\end{align*}
\]
end

\% Calculate initial ensemble curves
for \( n = 1:3 \)
\[
\begin{align*}
    \text{R\_hip\_EC}(i,n) &= \text{mean}(\text{R\_hip\_ang}(i,1:\text{length}(\text{stc\_R}),n))*180/\pi; \\
    \text{R\_knee\_EC}(i,n) &= \text{mean}(\text{R\_knee\_ang}(i,1:\text{length}(\text{stc\_R}),n))*-180/\pi; \\
    \text{R\_ankle\_EC}(i,n) &= \ldots \\
    \text{R\_thigh\_EC}(i,n) &= \ldots \\
    \text{R\_calf\_EC}(i,n) &= \ldots \\
    \text{R\_foot\_EC}(i,n) &= \ldots \\
\end{align*}
\]
for \( i = 1:\text{length}(\text{L\_hip\_flex}) \)
\[
\begin{align*}
    \text{L\_hip\_EC}(i,n) &= \text{mean}(\text{L\_hip\_ang}(i,1:\text{length}(\text{stc\_L}),n))*180/\pi; \\
    \text{L\_knee\_EC}(i,n) &= \text{mean}(\text{L\_knee\_ang}(i,1:\text{length}(\text{stc\_L}),n))*-180/\pi; \\
\end{align*}
\]
L_ankle_EC(i,n) = ...
    mean(L_ankle_ang(i,1:length(stc_L),n))*180/pi;
L_thigh_EC(i,n) = ...
    mean(L_thigh_ang(i,1:length(stc_L),n))*180/pi;
L_calf_EC(i,n) = mean(L_calf_ang(i,1:length(stc_L),n))*-180/pi;
L_foot_EC(i,n) = mean(L_foot_ang(i,1:length(stc_L),n))*180/pi;
end
end

R_hip_EC(:,2:3) = -1*R_hip_EC(:,2:3);
R_ankle_EC(:,2:3) = -1*R_ankle_EC(:,2:3);
R_thigh_EC(:,2:3) = -1*R_thigh_EC(:,2:3);
R_foot_EC(:,2:3) = -1*R_foot_EC(:,2:3);
L_knee_EC(:,2:3) = -1*L_knee_EC(:,2:3);
L_calf_EC(:,2:3) = -1*L_calf_EC(:,2:3);

% Remove outlying stances, R_hip
for n = 1:3
    for j = 1:size(R_hip_ang,2)
        for i = 1:length(R_hip_ang)
            resid(i,n) = (abs(R_hip_ang(i,j,n)*180/pi
        end
    end
    res(j,n) = mean(resid(:,n));
end
res_mean(n) = mean(res(:,n));
for n = 1:3
    for j = 1:length(res)
        if res(j,n) / res_mean(n) > 4.5
            for i = 1:length(R_hip_ang)
                if j <= size(R_hip_ang,2)
                    for k = j:(size(R_hip_ang,2)-1)
                        R_hip_ang(i,k,n) = R_hip_ang(i,k+1,n);
                    end
                    R_hip_EC(i,n) = mean(R_hip_ang(i, ...
                        (1:length(stc_R)-1),n)*180/pi);
                elseif j == 1
                    R_hip_EC(i,n) = mean(R_hip_ang(i, ...
                        2:length(stc_R),n)*180/pi);
                elseif j == size(R_hip flex,2)
                    R_hip_EC(i,n) = mean(R_hip_ang(i, ...
                        (1:length(stc_R)-1),n)*180/pi);
                end
            end
        end
    end
end
% Remove outlying stances, R_knee
for n = 1:3
    for j = 1:size(R_knee_ang,2)
        for i = 1:length(R_knee_ang)
            resid(i,n) = (abs(R_knee_ang(i,j,n)*180/pi - ... 
                           R_knee_EC(i,n)))^2;
        end
    end
    res(j,n) = mean(resid(:,n));
end
res_mean(n) = mean(res(:,n));
for n = 1:3
    for j = 1:length(res)
        if res(j,n) / res_mean(n) > 4.5
            for i = 1:length(R_knee_ang)
                if j <= size(R_knee_ang,2)
                    for k = j:(size(R_knee_ang,2)-1)
                        R_knee_ang(i,k,n) = R_knee_ang(i,k+1,n);
                    end
                    R_knee_EC(i,n) = mean(R_knee_ang(i,... 
                                           1:length(stc_R)-1),n)*-180/pi);
                elseif j == 1
                    R_knee_EC(i,n) = mean(R_knee_ang(i,... 
                                           2:length(stc_R),n)*-180/pi);
                elseif j == size(R_knee_flex,2)
                    R_knee_EC(i,n) = mean(R_knee_ang(i,... 
                                           1:length(stc_R)-1),n)*-180/pi);
                end
            end
        end
    end
end

% Remove outlying stances, R_ankle
for n = 1:3
    for j = 1:size(R_ankle_ang,2)
        for i = 1:length(R_ankle_ang)
            resid(i,n) = (abs(R_ankle_ang(i,j,n)*180/pi - ... 
                             R_ankle_EC(i,n)))^2;
        end
    end
    res(j,n) = mean(resid(:,n));
end
res_mean(n) = mean(res(:,n));
for n = 1:3
for j = 1:length(res)
    if res(j,n) / res_mean(n) > 4.5
        for i = 1:length(R_ankle_ang)
            if j <= size(R_ankle_ang,2)
                for k = j:(size(R_ankle_ang,2)-1)
                    R_ankle_ang(i,k,n) = R_ankle_ang(i,k+1,n);
                end
                R_ankle_EC(i,n) = mean(R_ankle_ang(i, ...
                        (1:length(stc_R)-1),n)*180/pi);
            elseif j == 1
                R_ankle_EC(i,n) = mean(R_ankle_ang(i, ...
                        2:length(stc_R),n)*180/pi);
            elseif j == size(R_ankle_ang,2)
                R_ankle_EC(i,n) = mean(R_ankle_ang(i, ...
                        (1:length(stc_R)-1),n)*180/pi);
            end
        end
    end
end
end

% Remove outlying stances, R_thigh
for n = 1:3
    for j = 1:size(R_thigh_ang,2)
        for i = 1:length(R_thigh_ang)
            resid(i,n) = (abs(R_thigh_ang(i,j,n)*180/pi - ...
                        R_thigh_EC(i,n)))^2;
        end
    end
    res(j,n) = mean(resid(:,n));
end
res_mean(n) = mean(res(:,n));
for n = 1:3
    for j = 1:length(res)
        if res(j,n) / res_mean(n) > 4.5
            for i = 1:length(R_thigh_ang)
                if j <= size(R_thigh_ang,2)
                    for k = j:(size(R_thigh_ang,2)-1)
                        R_thigh_ang(i,k,n) = R_thigh_ang(i,k+1,n);
                    end
                    R_thigh_EC(i,n) = mean(R_thigh_ang(i, ...
                                (1:length(stc_R)-1),n)*180/pi);
                elseif j == 1
                    R_thigh_EC(i,n) = mean(R_thigh_ang(i, ...
                                2:length(stc_R),n)*180/pi);
                elseif j == size(R_thigh_ang,2)
                    R_thigh_EC(i,n) = mean(R_thigh_ang(i, ...
                                (1:length(stc_R)-1),n)*180/pi);
                end
            end
        end
    end
end
\begin{verbatim}
% Remove outlying stances, R_calf
for n = 1:3
    for j = 1:size(R_calf_ang,2)
        for i = 1:length(R_calf_ang)
            resid(i,n) = (abs(R_calf_ang(i,j,n)*180/pi - ...
                R_calf_EC(i,n)))^2;
        end
    end
    res(j,n) = mean(resid(:,n));
end
res_mean(n) = mean(res(:,n));
for n = 1:3
    for j = 1:length(res)
        if res(j,n) / res_mean(n) > 4.5
            for i = 1:length(R_calf_ang)
                if j <= size(R_calf_ang,2)
                    for k = j:(size(R_calf_ang,2)-1)
                        R_calf_ang(i,k,n) = R_calf_ang(i,k+1,n);
                    end
                    R_calf_EC(i,n) = mean(R_calf_ang(i, ...
                        (1:length(stc_R)-1),n)*180/pi);
                elseif j == 1
                    R_calf_EC(i,n) = mean(R_calf_ang(i, ...
                        2:length(stc_R),n)*180/pi);
                elseif j == size(R_calf_ang,2)
                    R_calf_EC(i,n) = mean(R_calf_ang(i, ...
                        (1:length(stc_R)-1),n)*180/pi);
                end
            end
        end
    end
end

% Remove outlying stances, R_foot
for n = 1:3
    for j = 1:size(R_foot_ang,2)
        for i = 1:length(R_foot_ang)
            resid(i,n) = (abs(R_foot_ang(i,j,n)*180/pi - ...
                R_foot_EC(i,n)))^2;
        end
    end
end
end
\end{verbatim}
end
res(j,n) = mean(resid(:,n));
end
res_mean(n) = mean(res(:,n));
for n = 1:3
    for j = 1:length(res)
        if res(j,n) / res_mean(n) > 4.5
            for i = 1:length(R_foot_ang)
                if j <= size(R_foot_ang,2)
                    for k = j:(size(R_foot_ang,2)-1)
                        R_foot_ang(i,k,n) = R_foot_ang(i,k+1,n);
                    end
                    R_foot_EC(i,n) = mean(R_foot_ang(i, ... 
                        (1:length(stc_R)-1),n)*180/pi);
                elseif j == 1
                    R_foot_EC(i,n) = mean(R_foot_ang(i, ... 
                        (2:length(stc_R),n)*180/pi);
                elseif j == size(R_foot_flex,2)
                    R_foot_EC(i,n) = mean(R_foot_ang(i, ... 
                        (1:length(stc_R)-1),n)*180/pi);
                end
            end
        end
    end
end
% Remove outlying stances, L_hip
for n = 1:3
    for j = 1:size(L_hip_ang,2)
        for i = 1:length(L_hip_ang)
            resid(i,n) = (abs(L_hip_ang(i,j,n)*180/pi - ...
                L_hip_EC(i,n)))^2;
        end
    end
    res(j,n) = mean(resid(:,n));
end
res_mean(n) = mean(res(:,n));
for n = 1:3
    for j = 1:length(res)
        if res(j,n) / res_mean(n) > 4.5
            for i = 1:length(L_hip_ang)
                if j <= size(L_hip_ang,2)
                    for k = j:(size(L_hip_ang,2)-1)
                        L_hip_ang(i,k,n) = L_hip_ang(i,k+1,n);
                    end
                    L_hip_EC(i,n) = mean(L_hip_ang(i, ... 
                        (1:length(stc_L)-1),n)*180/pi);
                elseif j == 1
                    L_hip_EC(i,n) = mean(L_hip_ang(i, ... 
                        (2:length(stc_L),n)*180/pi);
                elseif j == size(L_hip_flex,2)
                    L_hip_EC(i,n) = mean(L_hip_ang(i, ... 
                        (1:length(stc_L)-1),n)*180/pi);
                end
            end
        end
    end
end
elseif j == 1
  L_hip_EC(i,n) = mean(L_hip_ang(i, ...
  2:length(stc_L),n)*180/pi);
elseif j == size(L_hip_flex,2)
  L_hip_EC(i,n) = mean(L_hip_ang(i, ...
    (1:length(stc_L)-1),n)*180/pi);
end
end
end
end

% Remove outlying stances, L_knee
for n = 1:3
  for j = 1:size(L_knee_ang,2)
    for i = 1:length(L_knee_ang)
      resid(i,n) = (abs(L_knee_ang(i,j,n)*180/pi - L_knee_EC(i,n)))^2;
    end
    res(j,n) = mean(resid(:,n));
  end
  res_mean(n) = mean(res(:,n));
  for n = 1:3
    for j = 1:length(res)
      if res(j,n) / res_mean(n) > 4.5
        for i = 1:length(L_knee_ang)
          if j <= size(L_knee_ang,2)
            for k = j:(size(L_knee_ang,2)-1)
              L_knee_ang(i,k,n) = L_knee_ang(i,k+1,n);
            end
            L_knee_EC(i,n) = mean(L_knee_ang(i, ...
              (1:length(stc_L)-1),n)*-180/pi);
          elseif j == 1
            L_knee_EC(i,n) = mean(L_knee_ang(i, ...
              2:length(stc_L),n)*-180/pi);
          elseif j == size(L_knee_flex,2)
            L_knee_EC(i,n) = mean(L_knee_ang(i, ...
              (1:length(stc_L)-1),n)*-180/pi);
          end
        end
      end
    end
  end
end

% Remove outlying stances, L_ankle
for n = 1:3
for j = 1:size(L_ankle_ang,2)
    for i = 1:length(L_ankle_ang)
        resid(i,n) = (abs(L_ankle_ang(i,j,n)*180/pi - 
                      L_ankle_EC(i,n)))^2;
    end
    end
res(j,n) = mean(resid(:,n));
end
res_mean(n) = mean(res(:,n));
for n = 1:3
    for j = 1:length(res)
        if res(j,n) / res_mean(n) > 4.5
            for i = 1:length(L_ankle_ang)
                if j <= size(L_ankle_ang,2)
                    for k = j:(size(L_ankle_ang,2)-1)
                        L_ankle_ang(i,k,n) = L_ankle_ang(i,k+1,n);
                    end
                    L_ankle_EC(i,n) = mean(L_ankle_ang(i, ... 
                                            (1:length(stc_L)-1),n)*180/pi);
                elseif j == 1
                    L_ankle_EC(i,n) = mean(L_ankle_ang(i, ... 
                                            2:length(stc_L),n)*180/pi);
                elseif j == size(L_ankle_ang,2)
                    L_ankle_EC(i,n) = mean(L_ankle_ang(i, ... 
                                            (1:length(stc_L)-1),n)*180/pi);
                end
            end
        end
    end
end

% Remove outlying stances, L_thigh
for n = 1:3
    for j = 1:size(L_thigh_ang,2)
        for i = 1:length(L_thigh_ang)
            resid(i,n) = (abs(L_thigh_ang(i,j,n)*180/pi - 
                             L_thigh_EC(i,n)))^2;
        end
        end
    res(j,n) = mean(resid(:,n));
end
res_mean(n) = mean(res(:,n));
for n = 1:3
    for j = 1:length(res)
        if res(j,n) / res_mean(n) > 4.5
            for i = 1:length(L_thigh_ang)
                if j <= size(L_thigh_ang,2)

```
for k = j:(size(L_thigh_ang,2)-1)
    L_thigh_ang(i,k,n) = L_thigh_ang(i,k+1,n);
end
L_thigh_EC(i,n) = mean(L_thigh_ang(i, ...)
    (1:length(stc_L)-1),n)*180/pi);
elseif j == 1
    L_thigh_EC(i,n) = mean(L_thigh_ang(i, ...)
    2:length(stc_L),n)*180/pi);
elseif j == size(L_thigh_flex,2)
    L_thigh_EC(i,n) = mean(L_thigh_ang(i, ...)
    (1:length(stc_L)-1),n)*180/pi);
end
end
end
end

% Remove outlying stances, L_calf
for n = 1:3
    for j = 1:size(L_calf_ang,2)
        for i = 1:length(L_calf_ang)
            resid(i,n) = (abs(L_calf_ang(i,j,n)*180/pi - ...
                L_calf_EC(i,n)))^2;
        end
        res(j,n) = mean(resid(:,n));
    end
    res_mean(n) = mean(res(:,n));
for n = 1:3
    for j = 1:length(res)
        if res(j,n) / res_mean(n) > 4.5
            for i = 1:length(L_calf_ang)
                if j <= size(L_calf_ang,2)
                    for k = j:(size(L_calf_ang,2)-1)
                        L_calf_ang(i,k,n) = L_calf_ang(i,k+1,n);
                    end
                    L_calf_EC(i,n) = mean(L_calf_ang(i, ...)
                        (1:length(stc_L)-1),n)*-180/pi);
                elseif j == 1
                    L_calf_EC(i,n) = mean(L_calf_ang(i, ...)
                        2:length(stc_L),n)*-180/pi);
                elseif j == size(L_calf_flex,2)
                    L_calf_EC(i,n) = mean(L_calf_ang(i, ...)
                        (1:length(stc_L)-1),n)*-180/pi);
                end
            end
        end
    end
% Remove outlying stances, L_foot
for n = 1:3
    for j = 1:size(L_foot_ang,2)
        for i = 1:length(L_foot_ang)
            resid(i,n) = (abs(L_foot_ang(i,j,n)*180/pi - ...
                          L_foot_EC(i,n)))^2;
        end
    end
    res(j,n) = mean(resid(:,n));
end
res_mean(n) = mean(res(:,n));
for n = 1:3
    for j = 1:length(res)
        if res(j,n) / res_mean(n) > 4.5
            for i = 1:length(L_foot_ang)
                if j <= size(L_foot_ang,2)
                    for k = j:(size(L_foot_ang,2)-1)
                        L_foot_ang(i,k,n) = L_foot_ang(i,k+1,n);
                    end
                    L_foot_EC(i,n) = mean(L_foot_ang(i, ...
                          (1:length(stc_L)-1),n)*180/pi);
                elseif j == 1
                    L_foot_EC(i,n) = mean(L_foot_ang(i, ...
                          2:length(stc_L),n)*180/pi);
                elseif j == size(L_foot_flex,2)
                    L_foot_EC(i,n) = mean(L_foot_ang(i, ...
                          (1:length(stc_L)-1),n)*180/pi);
                end
            end
        end
    end
end

R_hip_EC(:,2:3) = -1*R_hip_EC(:,2:3);
R_ankle_EC(:,2:3) = -1*R_ankle_EC(:,2:3);
R_thigh_EC(:,2:3) = -1*R_thigh_EC(:,2:3);
R_foot_EC(:,2:3) = -1*R_foot_EC(:,2:3);
L_knee_EC(:,2:3) = -1*L_knee_EC(:,2:3);
L_calf_EC(:,2:3) = -1*L_calf_EC(:,2:3);

% Calculate joint and segment angular velocities
for n = 1:3
    for i = 1:100
        if i == 1
R\_hip\_vel\_EC(i,n) = (R\_hip\_EC(i+1,n) - R\_hip\_EC(i,n)) / dt;
R\_knee\_vel\_EC(i,n) = (R\_knee\_EC(i+1,n) - ...
R\_knee\_EC(i,n)) / dt;
R\_ankle\_vel\_EC(i,n) = (R\_ankle\_EC(i+1,n) - ...
R\_ankle\_EC(i,n)) / dt;
R\_thigh\_vel\_EC(i,n) = (R\_thigh\_EC(i+1,n) - ...
R\_thigh\_EC(i,n)) / dt;
R\_calf\_vel\_EC(i,n) = (R\_calf\_EC(i+1,n) - ...
R\_calf\_EC(i,n)) / dt;
R\_foot\_vel\_EC(i,n) = (R\_foot\_EC(i+1,n) - ...
R\_foot\_EC(i,n)) / dt;
L\_hip\_vel\_EC(i,n) = (L\_hip\_EC(i+1,n) - L\_hip\_EC(i,n)) / dt;
L\_knee\_vel\_EC(i,n) = (L\_knee\_EC(i+1,n) - ...
L\_knee\_EC(i,n)) / dt;
L\_ankle\_vel\_EC(i,n) = (L\_ankle\_EC(i+1,n) - ...
L\_ankle\_EC(i,n)) / dt;
L\_thigh\_vel\_EC(i,n) = (L\_thigh\_EC(i+1,n) - ...
L\_thigh\_EC(i,n)) / dt;
L\_calf\_vel\_EC(i,n) = (L\_calf\_EC(i+1,n) - ...
L\_calf\_EC(i,n)) / dt;
L\_foot\_vel\_EC(i,n) = (L\_foot\_EC(i+1,n) - ...
L\_foot\_EC(i,n)) / dt;

elseif i == 100
R\_hip\_vel\_EC(i,n) = (R\_hip\_EC(i,n) - R\_hip\_EC(i-1,n)) / dt;
R\_knee\_vel\_EC(i,n) = (R\_knee\_EC(i,n) - ...
R\_knee\_EC(i-1,n)) / dt;
R\_ankle\_vel\_EC(i,n) = (R\_ankle\_EC(i,n) - ...
R\_ankle\_EC(i-1,n)) / dt;
R\_thigh\_vel\_EC(i,n) = (R\_thigh\_EC(i,n) - ...
R\_thigh\_EC(i-1,n)) / dt;
R\_calf\_vel\_EC(i,n) = (R\_calf\_EC(i,n) - ...
R\_calf\_EC(i-1,n)) / dt;
R\_foot\_vel\_EC(i,n) = (R\_foot\_EC(i,n) - ...
R\_foot\_EC(i-1,n)) / dt;
L\_hip\_vel\_EC(i,n) = (L\_hip\_EC(i,n) - ...
L\_hip\_EC(i-1,n)) / dt;
L\_knee\_vel\_EC(i,n) = (L\_knee\_EC(i,n) - ...
L\_knee\_EC(i-1,n)) / dt;
L\_ankle\_vel\_EC(i,n) = (L\_ankle\_EC(i,n) - ...
L\_ankle\_EC(i-1,n)) / dt;
L\_thigh\_vel\_EC(i,n) = (L\_thigh\_EC(i,n) - ...
L\_thigh\_EC(i-1,n)) / dt;
L\_calf\_vel\_EC(i,n) = (L\_calf\_EC(i,n) - ...
L\_calf\_EC(i-1,n)) / dt;
L\_foot\_vel\_EC(i,n) = (L\_foot\_EC(i,n) - ...
L\_foot\_EC(i-1,n)) / dt;

else

\[ R_{\text{hip vel}}_{EC}(i,n) = \frac{(R_{\text{hip EC}}(i+1,n) - R_{\text{hip EC}}(i-1,n))}{(2*dt)}; \]
\[ R_{\text{knee vel}}_{EC}(i,n) = \frac{(R_{\text{knee EC}}(i+1,n) - R_{\text{knee EC}}(i-1,n))}{(2*dt)}; \]
\[ R_{\text{ankle vel}}_{EC}(i,n) = \frac{(R_{\text{ankle EC}}(i+1,n) - R_{\text{ankle EC}}(i-1,n))}{(2*dt)}; \]
\[ R_{\text{thigh vel}}_{EC}(i,n) = \frac{(R_{\text{thigh EC}}(i+1,n) - R_{\text{thigh EC}}(i-1,n))}{(2*dt)}; \]
\[ R_{\text{calf vel}}_{EC}(i,n) = \frac{(R_{\text{calf EC}}(i+1,n) - R_{\text{calf EC}}(i-1,n))}{(2*dt)}; \]
\[ R_{\text{foot vel}}_{EC}(i,n) = \frac{(R_{\text{foot EC}}(i+1,n) - R_{\text{foot EC}}(i-1,n))}{(2*dt)}; \]

\[ L_{\text{hip vel}}_{EC}(i,n) = \frac{(L_{\text{hip EC}}(i+1,n) - L_{\text{hip EC}}(i-1,n))}{(2*dt)}; \]
\[ L_{\text{knee vel}}_{EC}(i,n) = \frac{(L_{\text{knee EC}}(i+1,n) - L_{\text{knee EC}}(i-1,n))}{(2*dt)}; \]
\[ L_{\text{ankle vel}}_{EC}(i,n) = \frac{(L_{\text{ankle EC}}(i+1,n) - L_{\text{ankle EC}}(i-1,n))}{(2*dt)}; \]
\[ L_{\text{thigh vel}}_{EC}(i,n) = \frac{(L_{\text{thigh EC}}(i+1,n) - L_{\text{thigh EC}}(i-1,n))}{(2*dt)}; \]
\[ L_{\text{calf vel}}_{EC}(i,n) = \frac{(L_{\text{calf EC}}(i+1,n) - L_{\text{calf EC}}(i-1,n))}{(2*dt)}; \]
\[ L_{\text{foot vel}}_{EC}(i,n) = \frac{(L_{\text{foot EC}}(i+1,n) - L_{\text{foot EC}}(i-1,n))}{(2*dt)}; \]

end
end
end
end

ITBS_output; % Subprogram outputs kinematics to MS Excel
end % for trial
end % for subj
REFERENCES


