Lower extremity mechanics of iliotibial band syndrome during an exhaustive run

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Abstract

Injury patterns in distance running may be related to kinematic adjustments induced by fatigue. The goal was to measure changes in lower extremity mechanics during an exhaustive run in individuals with and without a history of iliotibial band syndrome (ITBS). Sixteen recreational runners ran to voluntary exhaustion on a treadmill at a self-selected pace. Eight runners had a history of ITBS. Twenty-three reflective marker positions were recorded by an eight-camera 120 Hz motion capture system. Joint angles during stance phase were exported to a musculoskeletal model (SIMM) with the iliotibial band (ITB) modeled as a passive structure to estimate strain in the ITB. For ITBS runners, at the end of the run: (1) knee flexion at heel-strike was higher than control (20.6° versus 15.3°, p = 0.01); (2) the number of knees with predicted ITB impingement upon the lateral femoral epicondyle increased from 6 to 11. Strain in the ITB was higher in the ITBS runners throughout all of stance. Maximum foot adduction in the ITBS runners was higher versus 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 versus control at the end of the run. In conclusion, ITB mechanics appear to be related to changes in knee flexion at heel-strike and internal rotation of the leg. These observations may suggest kinematic discriminators for clinical assessment.

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1. Introduction

Iliotibial band syndrome (ITBS) is an overuse injury of the lateral knee that affects up to 12% of 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 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 during running. Treatment of ITBS can require months of conservative therapy, a reduction or cessation of running, or surgery [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 ITB tension, such as hip abductor weakness and a lack of ITB flexibility [3–5].

While clinical investigations of ITBS have been numerous, the biomechanics of ITBS have received comparatively little attention. Orchard et al. [6] performed a treadmill study of ITBS runners and found no differences in knee flexion angles between symptomatic and asymptomatic legs. Messier et al. [1] found that runners with a history of ITBS reduced rearfoot pronation during heel-strike compared to healthy runners. The ITBS runners also showed faster maximum supination velocities, smaller maximum braking 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 running injuries [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 was constructed to investigate strain in the ITB during the run. It was hypothesized that ITB strain would be significantly higher in ITBS runners, and that the difference would 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 the study. Table 1 shows participant 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’ ITB flexibilities were assessed by a modified Ober test [13] procedure. 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, 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.

2.2. Protocol

Participants first completed a questionnaire that reported their training program, injury history and recent 5-km race time.

An eight-camera 120 Hz Peak Motus motion capture system (Vicon Peak, Centennial, CO) captured 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.

Participants performed a static trial by standing in the center of the camera ring, facing the front of the room with their feet shoulder width apart while marker positions were recorded for 1 s. The four medial markers were removed after the static trial.

Dynamic trials were performed on an adjustable speed Quinton treadmill (Stairmaster, Kirkland, WA) at zero percent grade. Participants were instructed to select a running pace that would exhaust them within 20 min. The selected pace was maintained throughout the dynamic trial. Participants were allowed to warm up at a slower pace for 2 min. Marker positions were recorded for 10 s every 2 min until the participants reported voluntary exhaustion. Participants ran in their own shoes and wore a black spandex shirt and shorts.

2.3. Data analysis

Ankle and knee joint centers were estimated as 50% of the distance between the medial and lateral markers at the respective joint. Hip joint centers were estimated as 25% of the distance between the ipsilateral and contralateral trochanters. For the dynamic trials, the four removed medial markers were reconstructed 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.

A custom MATLAB program (MathWorks, Natick, MA) isolated the stance phase by locating pairs of successive minimum vertical heel and toe coordinates. The code defined local coordinate systems (LCS) at each segment during the static and dynamic trials and calculated three-dimensional Cardan angles during stance for each joint and segment consistent with the method of Robertson et al. [14]. Rotational transformation matrices (RTM) between the static LCS and the global (room) coordinate system (GCS) defined the zero orientation of the segments relative to the GCS. The dynamic segment LCS were

<table>
<thead>
<tr>
<th>Table 1</th>
<th>Subject characteristics</th>
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<tbody>
<tr>
<td></td>
<td>Age (years)</td>
<td>Height (cm)</td>
</tr>
<tr>
<td>ITBS</td>
<td>27.5 (9.0)</td>
<td>170.1 (6.9)</td>
</tr>
<tr>
<td>Control</td>
<td>26.4 (7.7)</td>
<td>172.8 (8.9)</td>
</tr>
</tbody>
</table>

Values are mean (S.D.). N = 8 for both groups.
multiplied by the RTM to provide the segment coordinate systems (SCS) used to calculate dynamic joint and segment angles. Angles were calculated with a Cardan rotation sequence of three successive rotations (flexion extension, adduction/abduction, and internal/external rotation). Segment angles were calculated beginning at the GCS axes and ending at the SCS axes. Joint angles were calculated beginning at the distal SCS axes and ending at the proximal SCS axes. Kinematic curves were interpolated to 100-point splines. Ensemble curves for the kinematics were generated by averaging the splines for every stance phase in a 10-s dynamic data block. 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 structures that originated at the iliac crest and terminated at the femoral insertion of the gluteus maximus, the fibular head, and Gerdy’s tubercle.

Joint angles were imported into SIMM motion files. Bone and muscle geometries were scaled to the participant’s leg length. 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 contacted the LFE wrapping sphere. A wrapping cylinder along the femur prevented structures from passing through the femur. Fig. 1 shows a sagittal view of the SIMM model. Strain in the ITB was calculated by dividing the change in length of the ITB by the length of the ITB in the anatomical position.

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. Two-by-two factorial ANOVA with group (ITBS or control) and time (start or end) as the independent variables was performed on the dependent variables from the discriminant analyses that accounted for the most variability in the data. Post hoc t-tests were performed on variables that were significantly different between groups. The ITB strains were compared between groups using t-tests. A difference 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 been symptomatic in both legs. One runner had symptoms only in the left leg, and one had symptoms only in the right leg. All ITBS runners had been diagnosed by a physical therapist or a doctor. Table 2 shows the training data and Ober test results. While none of the variables in Table 2 were significantly different, the ITBS runners showed trends of higher weekly mileage and slower dynamic trial paces. On average, ITBS runners

| Training data responses, Ober test (OT) measurements and dynamic trial pace and time |
|---------------------------------|-------|-------|--------|
|                                 | ITBS  | Control| p-Value |
| Weekly mileage                  | 23.7 (15.1) | 11.8 (5.9) | 0.06   |
| Months at mile age              | 6.0 (3.2)  | 6.3 (2.5)  | 0.87   |
| 5-km Race time (min)            | 23.4 (5.3) | 22.9 (2.6) | 0.84   |
| OT right leg, 0° (°)            | 17.4 (6.8) | 19.9 (11.5) | 0.62   |
| OT right leg, 90° (°)           | 12.6 (7.4) | 13.9 (11.7) | 0.79   |
| OT left leg, 0° (°)             | 15.9 (5.2) | 17.0 (8.4)  | 0.56   |
| OT left leg, 90° (°)            | 13.2 (5.5) | 15.2 (7.1)  | 0.54   |
| Trial time (min)                | 16.1 (5.0)  | 16.3 (6.0)  | 0.95   |
| Trial pace (min/mile)           | 9.5 (2.2)   | 7.8 (1.0)   | 0.08   |

Values are mean (S.D.).
were slightly less flexible than control runners in all Ober test measurements.

3.2. Dynamic trial kinematics

Kinematics during stance were compared between the ITBS and control groups. At the beginning of the runs, there were significant differences during stance in maximum foot adduction (2.6 ± 7.9° for ITBS versus −6.1 ± 7.6° for control, p = 0.003), minimum knee flexion (12.5 ± 3.6° for ITBS versus 7.7 ± 3.8° for control, p = 0.01), and maximum ankle dorsiflexion velocity (74.8 ± 19.7°/s for ITBS versus 54.9 ± 27.9°/s for control, p = 0.03). At the end of the runs, there were significant differences during stance in maximum knee flexion (43.8 ± 7.8° for ITBS versus 36.5 ± 9.2° for control, p = 0.02), maximum foot inversion (3.3 ± 10.9° for ITBS versus −9.5 ± 19.6° for control, p = 0.03), minimum thigh flexion velocity (−29.5 ± 16.9°/s for ITBS versus −16.7 ± 8.2°/s for control, p = 0.01), and maximum knee internal rotation velocity (16.4 ± 9.3°/s for ITBS versus 10.3 ± 4.0°/s for control, p = 0.02). Tables 3 and 4 summarize the significant kinematic results.

Fig. 2 shows knee flexion at heel-strike. At the beginning of the run, the knee flexion at heel-strike were similar between the groups (17.2 ± 7.0° for ITBS versus 15.2 ± 3.9° for control, p = 0.16). At the end of the run, ITBS knees were significantly more flexed at heel-strike than control knees (20.6 ± 7.8° for ITBS versus 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 (six to eleven) but decreased for the control group (six to four). For all unimpinged knees, the ITB was anterior to the LFE at heel-strike, and impingement had not yet occurred.

3.3. Strain

Fig. 3 shows the strain in the ITB during stance phase. At heel-strike, both groups had approximately 6% ITB strain at 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. Peak strain during stance was higher for the ITBS runners at both the beginning (8.5 ± 1.2% for ITBS versus 7.5 ± 0.6% for control) and end (8.4 ± 1.3% for ITBS versus 7.5 ± 0.7% for control) of 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 iliobibial 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 extended runs [15,16]. Greater knee flexion could indicate a smaller calf angle at heel-strike, which has been associated with less efficient running economy [17]. It should be noted that the rotation sequence used to calculate joint and segment angles could have introduced crosstalk errors in the frontal and transverse kinematics [18].

Previous research has suggested that decreasing the knee flexion at heel-strike predisposes a runner to ITBS by increasing the time spent in the impingement range of knee flexion angles [1,6]. The impingement range is defined as the range of knee flexion angles when 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 ITB in a compromised orientation where impingement with the LFE during heel-strike is more likely to occur.
The hypothesis that ITBS runners would increase strain in the ITB with fatigue was not supported. There was only a small (<0.5%) increase in strain for ITBS runners during early stance. However, ITBS runners demonstrated higher strain than control runners throughout the entire stance phase. 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 ITB-related injury.

The increased strain in the ITBS runners was likely a function of the kinematic variables. Previous research [2] and the present results suggest a zone 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. This condition could increase the stance-time spent in the impingement zone. Changes in strain during stance suggest that control runners’ ITB released from impingement towards the end of stance while ITBS runners’ ITB remained impinged. From Fig. 3, control runners decreased strain from 6% at heel-strike to 5% at toe-off, while ITBS runners increased strain from 6% at heel-strike to 7% at toe-off.

Based on the kinematics and ITB strain results, an injury mechanism for ITBS is proposed: with fatigue, as the knee becomes more flexed at heel-strike, the 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 stress in the ITB, causing more irritation and pain and suggesting that the initial “shock” of impact and immediate loading of the ITB play important roles in injury. At the beginning of the run, the ITBS runners had a noticeably higher strain loading rate following heel-strike (see slopes of Fig. 3), lending further support to the idea that the orientation 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] noted that the active impact forces during the latter three-fourths of stance are larger in magnitude and of longer duration than the passive impact forces near heel-strike. Peak strain in both groups occurred near mid-stance, regardless of fatigue. Additionally, peak strain was significantly higher in the ITBS runners. Group differences in strain during the latter phases of stance were likely due to the kinematic differences.

The long-term goal of injury research should be to reduce the frequency of 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 is typically limited to retrospective studies. The knowledge base of injury biomechanics would benefit greatly from prospective investigations of large groups of runners over time that compare the mechanics of those who develop injuries to those who do not. In a prospective study of 400 previously healthy runners, those who developed lower extremity injuries 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.

Table 3
Kinematic differences between the ITBS and control groups at the beginning of the run

<table>
<thead>
<tr>
<th></th>
<th>ITBS</th>
<th>Control</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Maximum foot inversion</td>
<td>−2.5 (9.3)</td>
<td>3.3 (19.0)</td>
<td>n.s.</td>
</tr>
<tr>
<td>Maximum foot adduction</td>
<td>2.6 (7.6)</td>
<td>−6.1 (7.9)</td>
<td>0.003</td>
</tr>
<tr>
<td>Maximum knee flexion</td>
<td>42.6 (5.9)</td>
<td>38.0 (8.9)</td>
<td>n.s.</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>Minimum thigh flexion velocity</td>
<td>−26.4 (19.3)</td>
<td>−17.3 (8.9)</td>
<td>n.s.</td>
</tr>
<tr>
<td>Maximum knee internal rotation velocity</td>
<td>13.9 (9.4)</td>
<td>10.2 (4.5)</td>
<td>n.s.</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>
</tbody>
</table>

Angles are measured in degrees. Velocities are measured in degrees per second. Values are mean (S.D.). Significance was defined at p < 0.05.

Table 4
Kinematic differences between the ITBS and control groups at the end of the run

<table>
<thead>
<tr>
<th></th>
<th>ITBS</th>
<th>Control</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<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>Maximum foot adduction</td>
<td>0.0 (8.8)</td>
<td>−4.5 (9.6)</td>
<td>n.s.</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>Minimum knee flexion</td>
<td>11.1 (4.2)</td>
<td>7.4 (6.2)</td>
<td>n.s.</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>
<tr>
<td>Maximum ankle dorsiflexion velocity</td>
<td>71.0 (21.2)</td>
<td>55.1 (25.7)</td>
<td>n.s.</td>
</tr>
</tbody>
</table>

Angles are measured in degrees. Velocities are measured in degrees per second. Values are mean (S.D.). Significance was defined at p < 0.05.
The present research has suggested variables that may predispose runners to ITBS. Although the present research was retrospective, treatments of ITBS tend to focus on symptoms rather than causes. The ITBS runners may therefore have still exhibited the kinematics that influenced their injuries. Clinically, these results could be used to adjust the kinematics of ITBS runners, via orthotics 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 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 running injuries [23–26]. Without GRF data, it was not possible to realistically estimate the tension in the ITB. 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 would be expected to magnify the trends shown in Figs. 3 and 4. However, the present model excluded the insertions of the gluteus maximus and tensor fascia lata into the ITB. Tension in these muscles should have a direct effect on ITB tension. 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 foot, 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, 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 injury mechanisms.

A recent prospective study by Noehren et al. [29] suggested a role for increased hip adduction in ITB strain. While the present study did not find any significant main effects at the hip, the method for calculating the hip joint center was an estimate and may have influenced the results. Future studies should closely examine the role of hip kinematics in ITBS.

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 exhaustion criteria. Under these conditions, it is possible (and likely) that the participants reached different levels of fatigue. The degree of fatigue could be standardized by having participants maintain a running speed associated with fraction of their aerobic 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. 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 pace may have been kinematically awkward, and may have produced data that were not representative of the conditions during which the runners developed their injuries. A non-standardized running pace is not necessarily a large source of error. Queen et al. [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.

In conclusion, the runners with a history of ITBS demonstrated more internal rotation of the leg, and larger increases in knee flexion angle at heel-strike, near voluntary exhaustion. Internal rotation of the leg should increase strain in the ITB, as was demonstrated by a simple model of the ITB. The model also predicted an increase in cases of ITB impingement at heel-strike in the ITBS runners as they neared exhaustion. The results may suggest kinematic discriminators for clinical assessments, and possible targets for treating and preventing ITBS with orthotics or gait retraining. Future ITBS research should further consider lower extremity mechanics at heel-strike, as well as swing phase kinematics immediately before heel-strike. It is important to note that the ITB model was simplistic and lacked the important anatomical features of muscular insertions and the broad, sleeve-like structure of the ITB. Consequently, although it is notable that the ITBS runners demonstrated more ITB strain than control runners, the modeling results should be interpreted more as a difference between groups under these simplifying assumptions, rather than true physiological ITB strain. Future work will address these limitations with a more complex model that includes kinetics and muscular insertions that may influence ITB mechanics.

Fig. 4. Peak ITB strain during stance for the ITBS and control groups at the beginning and end of the run. The symbol (#) denotes significantly higher than rested control (p = 0.03) and (*) denotes significantly higher than fatigued control (p = 0.005).
References


