INTRODUCTION

Iliotibial band syndrome (ITBS) is the second most commonly reported overuse running injury [1]. The iliotibial band is a thickened lateral component of the tensor fasciae latae [2]. This band of dense fibrous connective tissue traverses down the lateral thigh and passes over the lateral femoral epicondyle before inserting onto the anterolateral aspect of the tibia [2]. Until recently, it was believed that the iliotibial band rolled over the lateral femoral epicondyle during activities involving repeated knee joint flexion and extension which caused friction [3]. This friction force was thought to lead to inflammation of the iliotibial band [3]. However, based on cadaveric experiments, anterior-posterior glide of the iliotibial band over the lateral femoral epicondyle may not be possible [2]. Instead, it has been proposed that medio-lateral motion of the iliotibial band compresses adipose tissue containing free nerve endings lying beneath the band, thus, causing pain during activity [2]. An increase in hip adduction angle may be one mechanism that increases medio-lateral motion of the iliotibial band. Results from previous studies examining running kinematics supports the premise that increased hip adduction but not knee flexion angle differences are exhibited in runners with ITBS compared to controls [4, 5]. In addition to increased hip adduction angle, the results from modeling investigations suggest that increased strain magnitude [6] and strain rate [7] may also be biomechanical risk factors associated with ITBS. However, it has yet to be determined if runners with increased iliotibial band strain also exhibit increased iliotibial band force. Therefore, the purpose of this investigation was to determine if differences in iliotibial band force exist between female runners with and without a history of ITBS. We also sought to determine if iliotibial band strain and strain rate differences exist between groups during running. It was hypothesized that iliotibial band force, strain, and strain rate would be greater in runners with a history of ITBS compared to controls.

METHODS

Ten currently healthy female runners aged between 18 and 35 years old were recruited. All procedures were approved by the Institutional Review Board prior to the commencement of the study. All participants provided written informed consent to participate. Runners with a history of ITBS (n = 5; age: 25.2 ± 4.9 years; height: 1.67 ± 0.05 m; mass: 55.2 ± 2.7 kg; weekly mileage: 25.4 ± 10.1 mi·wk⁻¹) were matched for height and weight with a control group (n = 5; age: 20.6 ± 2.1 years; height: 1.6 ± 0.01 m; mass: 58.0 ± 5.3 kg; weekly mileage: 25.2 ± 9.8 mi·wk⁻¹). Running data were collected using standard three-dimensional motion capture techniques. Participants’ lower extremities and trunk were instrumented with passive reflective markers. Standard laboratory footwear was worn by participants. Marker trajectories were collected using a 9 camera optoelectric motion capture system sampling at 120 Hz. Participants ran at a velocity of 3.5 m·s⁻¹ (± 5%) over a 17 m runway for 5 acceptable trials. Two force plates sampling at 1200 Hz were used to determine the stance phase. Data were processed using Visual 3D (C-Motion, Rockville, MD). Five motion trials from Visual 3D for each participant was exported to OpenSim [8]. The SimBody gait model was scaled to match each participant’s anthropometrics based on markers placed over anatomical landmarks during data collection. The iliotibial band was then added to the scaled model following the tensor fasciae latae’s path. The resting length of the iliotibial band was determined as the tensor fasciae latae’s length during the static calibration trial. A wrap sphere object was defined at the height of the lateral femoral epicondyle. The sphere’s surface was flush with the surface of the lateral femoral epicondyle to prevent the iliotibial band from penetrating through the femur during running. The iliotibial band was modeled as a muscle with only a passive contractile component and an optimal muscle fiber length of approximately zero. A generic tendon force-length
curve was scaled using the iliotibial band’s resting length and a stiffness value of 97 N/mm [9]. To track participant’s running motion, joint moments were calculated using a residual reduction algorithm (RRA) [8]. Using the kinematic output from the RRA iliotibial band length during the stance phase of running was computed using the Muscle Analysis tool in OpenSim. Iliotibial band strain was calculated by dividing the change in length of the band during stance and dividing by its resting length at each time frame. Strain rate was computed using the finite difference method. Lastly, iliotibial band force was computed by multiplying the iliotibial band’s change in length by the stiffness value of 97 N/mm. Dependent variables are reported at peak knee flexion [7]. Additionally, peak knee flexion and hip adduction angles were computed to aid in interpretation of the simulation results. Data were analyzed using descriptive statistics and effect size. Moderate effects were considered clinically relevant (≥ 0.5).

RESULTS AND DISCUSSION

Large effect sizes were associated with greater iliotibial band strain and force and peak hip adduction angle in the ITBS group compared to controls (Table 1). Moderate effects were associated with greater iliotibial band strain rate and and decreased knee flexion angle in the ITBS group than controls.

The purpose of this investigation was to determine if iliotibial band strain and force differences exist between female runners with and without a history of ITBS. Our findings support the hypotheses. Previous investigations have also found iliotibial band strain [6] and strain rate [7] to be higher in females runners currently with ITBS and in a prospective investigation, respectively. Although the direction of the current findings is similar with previous studies, strain magnitude was different. This is likely due to the different definitions of the path of the iliotibial band between studies even though resting length was computed the same. A discrepancy between in vivo iliotibial band resting length and the current methods of determining resting length are further compounded when determining iliotibial band force. An inaccurate resting length of even a few millimeters can mean the difference between iliotibial band force values being within or outside’s its maximal tensile force value of 582 N [9].

CONCLUSIONS

Our current method of determining iliotibial band strain in OpenSim produces results that are in agreement with previous ITBS modeling investigations. Runners with a history of ITBS exhibit increased strain and strain rates compared to controls. However, defining the iliotibial band’s path is indeed crucial to producing accurate simulation results for which comparisons can be made between studies. We suggest that providing a 0.5 cm range of iliotibial band resting length is a reasonable approach to determine if strain or force fall within these bounds to decide which runners may be at risk for developing ITBS.

REFERENCES

### Table 1: Mean(sd) of the variables of interest at peak knee flexion during stance

<table>
<thead>
<tr>
<th>Parameters</th>
<th>ITBS</th>
<th>Control</th>
<th>Effect Size</th>
</tr>
</thead>
<tbody>
<tr>
<td>Strain (%)</td>
<td>2.26(3.98)</td>
<td>-0.39(1.59)</td>
<td>0.84</td>
</tr>
<tr>
<td>Strain rate (%/s)</td>
<td>40.08(7.08)</td>
<td>37.05(3.55)</td>
<td>0.55</td>
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<tr>
<td>Force (N)</td>
<td>1187.31(2033.31)</td>
<td>-112.81(685.87)</td>
<td>0.82</td>
</tr>
<tr>
<td>Peak knee flexion (°)</td>
<td>46.44(5.88)</td>
<td>49.73(3.44)</td>
<td>0.68</td>
</tr>
<tr>
<td>Peak hip adduction (°)</td>
<td>15.0(2.4)</td>
<td>11.53(2.7)</td>
<td>1.14</td>
</tr>
</tbody>
</table>
**Figure 1.** Representative iliotibial band strain curves during stance for a participant with a history of ITBS.

**Figure 2.** A representative iliotibial band strain curves during stance for a control participant.
Control Participant

IT-band Strain (%) vs. Stance (%)

-6 -4 -2 0 2 4 6

Stance (%)