INTRODUCTION

Foot orthotic devices (FODs) are used to control excessive motion at the rearfoot, thereby alleviating overuse injuries such as posterior tibial tendonitis. In running some rearfoot eversion is necessary for shock attenuation (Oakley & Pratt, 1988), therefore, using FODs may result in an increase in tibial shock and subsequently increase the risk of stress-related bony injuries such as tibial stress fractures.

While most runners strike the ground with their rearfoot first, approximately 20% of distance runners are midfoot or forefoot strike (FFS) runners (Kerr, et al., 1983). FFS runners have been shown to have increased eversion excursions compared to rearfoot strike (RFS) runners (McClay & Manal, 1995a) and therefore, may benefit from orthotic intervention. The heel, however, does not make initial contact with the ground in a FFS running pattern. Therefore, it was hypothesized that the use of a FOD would increase shock in a RFS pattern but not in a FFS pattern.

A FFS pattern requires eccentric action of the gastrocnemius, soleus, and tibialis posterior as the heel lowers. These mechanisms are responsible for the lack, or reduction, of an initial impact peak and lower vertical ground reaction force load rates compared to RFS runners (McClay & Manal, 1995b). Hennig et al. (1991, 1993) reported strong positive correlations between vertical ground reaction force load rates and lower extremity shock. Therefore, it was hypothesized that a FFS pattern would exhibit lower shock due to the greater joint angular excursions and lower vertical load rates compared to a RFS pattern.

The purpose of this study was twofold: (1) to assess the effects of orthotic intervention and strike pattern upon tibial shock and (2) to compare the tibial shock between a FFS pattern and a RFS pattern.

METHODS

Fifteen RFS runners between the ages of 18 and 45 years with no previous FOD use participated in the study. Subjects with forefoot valgus or varus, limited rearfoot motion, or lower extremity misalignments in standing were excluded. Semi-rigid FODs with a 6° rearfoot varus post were fabricated from a non-weight bearing, neutral positioned plaster cast. Following a two week adjustment period to the FODs, the subjects had three-dimensional (3-D) kinematic, tibial acceleration, and ground reaction force data collected on their dominant leg.

Five trials of each condition were collected at a run speed of 3.7 m•s⁻¹ ± 5%. A uniaxial accelerometer (PCB Piezotronics, Depew, NY) was used to collect tibial acceleration at 960 Hz. The accelerometer was fastened to the distal, anteromedial aspect of the leg. Video data were collected at 120 Hz by 6 cameras (VICON, Oxford Metrics, UK). Ground reaction force data were collected at 960 Hz with a 60 X 90-cm. Bertec force plate (Bertec Corp., OH).

Dependent variables of interest were the peak positive acceleration value (PPA), the difference between the maximum and minimum acceleration values (PTP), and the median power frequency of the acceleration signal (MPFA). Vertical ground reaction force load rate (VLR) in the RFS pattern was determined from the slope of the ground reaction force to the first impact peak. The average percent stance to the first peak vertical ground reaction force for each subject’s RFS pattern was then used as the cut off to determine VLR in the FFS pattern. Kinematics were calculated using MOVE3D (NIH Biomechanics Laboratory, Bethesda, MD). All angular displacement data were resolved about a joint coordinate system (Grood & Suntay, 1983). Eversion excursion (EVEXC) and dorsiflexion excursion (DFEXC) were determined as the total angular range of motion in that plane during the first 60% of stance phase.
Ankle (Ka) and knee joint stiffness (Kk) were calculated as the slope of the torque-angle profile during the shock-absorbing phase of stance (Stefanyshyn & Nigg, 1998, Hamill et al. 2000). Leg stiffness (Kl) was estimated during running using a mass-spring model presented by McMahon and Cheng (1990). Two-way repeated measures ANOVAs were calculated for all variables.

RESULTS
No interactions were present in any of the variables tested. Therefore, only main effects were addressed (Table 1). Changes in EVEXC had no relationship to PPA (r=0.08). The FFS pattern exhibited an increase of 1.6 g in PPA compared to the RFS pattern. Representative tibial acceleration curves are presented in Figure 1. In addition, PTP and MPFA were significantly greater in the FFS pattern compared to the RFS pattern (p<0.05).

Table 1. Means, standard deviations, and P-values of kinematic, kinetic and tibial shock data.

<table>
<thead>
<tr>
<th></th>
<th>No Orthotic</th>
<th>Orthotic</th>
<th>FFS</th>
<th>RFS</th>
<th>p orth</th>
<th>p strike</th>
</tr>
</thead>
<tbody>
<tr>
<td>PPA (g)</td>
<td>7.88 (3.29)</td>
<td>7.52 (3.54)</td>
<td>8.49 (3.84)</td>
<td>6.91 (2.99)</td>
<td>.369</td>
<td>.093</td>
</tr>
<tr>
<td>MPFA (Hz)</td>
<td>47.96 (7.21)</td>
<td>49.82 (9.98)</td>
<td>51.79 (8.97)</td>
<td>45.99 (8.22)</td>
<td>.245</td>
<td>.023</td>
</tr>
<tr>
<td>VLR (bw/s)</td>
<td>50.00(12.09)</td>
<td>44.29(12.44)</td>
<td>39.66(13.16)</td>
<td>54.64(11.37)</td>
<td>.007</td>
<td>0.00</td>
</tr>
<tr>
<td>EVEXC (*)</td>
<td>15.59 (4.50)</td>
<td>14.45 (3.52)</td>
<td>16.38 (3.94)</td>
<td>13.66 (4.08)</td>
<td>.105</td>
<td>.005</td>
</tr>
<tr>
<td>DFEXC (*)</td>
<td>24.30 (4.87)</td>
<td>26.51 (4.21)</td>
<td>31.57 (4.31)</td>
<td>19.24 (4.77)</td>
<td>.017</td>
<td>0.00</td>
</tr>
<tr>
<td>Ka (N*m/deg.)</td>
<td>10.98 (3.02)</td>
<td>10.35 (2.54)</td>
<td>7.12 (2.03)</td>
<td>14.21 (3.53)</td>
<td>0.13</td>
<td>0.00</td>
</tr>
<tr>
<td>Kk (N*m/deg.)</td>
<td>5.00 (1.20)</td>
<td>5.24 (1.14)</td>
<td>5.47 (1.30)</td>
<td>4.77 (1.04)</td>
<td>0.01</td>
<td>0.00</td>
</tr>
<tr>
<td>Kl (kN/m)</td>
<td>7.61 (1.26)</td>
<td>7.81 (1.11)</td>
<td>8.49 (1.41)</td>
<td>6.93 (0.96)</td>
<td>0.34</td>
<td>0.00</td>
</tr>
</tbody>
</table>

DISCUSSION
Orthotic intervention did not alter tibial shock in either foot strike pattern. The hypothesis was based on the idea that restriction of rearfoot motion would result in an increase in tibial shock. EVEXC, however, was not significantly reduced with the use of the FOD. In addition there was no correlation
between changes in EVEXC and changes in PPA with orthotic intervention. As a result of the similarities in excursions between orthotic conditions, the ankle stiffness, and leg stiffness were similar.

It was interesting to note that VLR was significantly decreased with orthotic intervention. Because impulsive loading has been associated with stress fractures in bone studies (Burr et al., 1990), this may contribute to the reported reduction in both metatarsal and tibial stress fractures in military recruits with the use of FODs (Simkin et al., 1989).

Tibial shock was expected to be greater in a RFS compared to a FFS pattern. This hypothesis was based on a previous study by Hennig et al. (1991, 1993) that found a strong positive correlation between VLR and PPA. This relationship was assessed in RFS patterns only and may be different for a FFS pattern. The FFS pattern demonstrated a significantly lower VLR, which was surprisingly associated with greater PPA, PTP and MPFA compared to a RFS pattern.

To further explain the greater tibial shock in the FFS pattern, lower extremity stiffness was also assessed. Stiffness is inversely related to joint angular excursions or the linear excursion of the body’s center of mass. The FFS pattern resulted in increased dorsiflexion excursion and decreased ankle stiffness, compared to the RFS. Knee flexion excursion, however, was decreased in the FFS pattern resulting in increased knee stiffness. Overall leg stiffness also increased in the FFS pattern, suggesting that the knee may be a stronger modulator of overall lower extremity stiffness than the ankle. In a similar study of RFS and FFS patterns, Hamill et al. (2000) also noted the greater stiffness in a FFS pattern and greater contribution of the knee to overall leg stiffness.

In summary, footstrike pattern had a greater influence on tibial shock than did orthotic intervention. Further understanding of differences between FFS and RFS patterns may help in the identification of injury risks and development of preventative strategies for runners with varying strike patterns.

References