Leg Tissue Mass Composition Affects Tibial Acceleration Response Following Impact

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To date, there has not been a direct examination of the effect that tissue composition (lean mass/muscle, fat mass, bone mineral content) differences between males and females has on how the tibia responds to impacts similar to those seen during running. To evaluate this, controlled heel impacts were imparted to 36 participants (6 M and 6 F in each of low, medium and high percent body fat [BF] groups) using a human pendulum. A skin-mounted accelerometer medial to the tibial tuberosity was used to determine the tibial response parameters (peak acceleration, acceleration slope and time to peak acceleration). There were no consistent effects of BF or specific tissue masses on the un-normalized tibial response parameters. However, females experienced 25% greater peak acceleration than males. When normalized to lean mass, wobbling mass, and bone mineral content, females experienced 50%, 62% and 70% greater peak acceleration, respectively, per gram of tissue than males. Higher magnitudes of lean mass and bone mass significantly contributed to decreased acceleration responses in general.

Keywords: lower extremity, body composition, sex differences

Repetitive impact loading to the lower extremities during activities such as walking, running, and marching can result in a variety of musculoskeletal disorders. Each heel strike initiates a shock wave that travels proximally through the leg, increasing the risk of musculoskeletal maladies such as chronic joint pain (Light et al., 1980; Voloshin & Wosk, 1981, 1982), osteoarthritis, cartilage degeneration (Buckwalter & Lane, 1997), and stress fractures (Jones et al., 2002). Impact forces can be attenuated actively, through the modification of body kinematics, muscle activity (Cholewicki & McGill, 1995), and joint positions (Hamill et al., 1995), or passively, through the movement of fat mass and lean/muscle mass (collectively referred to as wobbling mass; Gruber et al., 1998) relative to the underlying bone (Chu et al., 1986; Hamill et al., 1995; Lafontue et al., 1996a). Previous work has shown that varying the levels of leg muscle activation at impact can affect shock wave attenuation through the leg (Flynn et al., 2004; Holmes & Andrews, 2006). It was proposed that higher levels of leg muscle activation increased muscle stiffness, which reduced the muscles’ ability to dampen the impact shock wave and increased transmission to the proximal tibia.

While Liu and Nigg (2000) demonstrated a positive relationship between the magnitude of rigid and soft tissue masses (bone and wobbling mass) of the lower extremity and simulated ground reaction force peaks, the precise effects of whole body composition and individual tissue masses of the leg (i.e., fat mass, lean mass, and bone mineral content) on impact attenuation is not known. Wakeling and Nigg (2001) also proposed that oscillations of the leg soft tissues were a significant contributor to the dampening of impact-induced shock waves. However, the tissues of the leg have often been treated as a singular mass and the individual contributions of the soft and rigid tissues to shock attenuation have not been established. Given the inherent differences in material properties between the tissues types, it is reasonable to assume that each may play a different role in attenuation, as measured in terms of their effect on tibial acceleration response parameters (e.g., peak acceleration, time to peak acceleration, acceleration slope) (Flynn et al., 2004; Holmes & Andrews, 2006; Duquette & Andrews, 2010a, 2010b). Furthermore, significant sex differences have been found in the magnitudes and proportions of these tissue masses (Lemieux et al., 1993; Ogle et al., 1995). While Milner et al. (2006) suggested that females diagnosed with a tibial stress fracture recorded greater tibial shock, and numerous authors have identified a greater risk of tibial stress...
fractures (up to 5 times) in females compared with males (Bell et al., 2000; Cline et al. 1998; Jones et al. 2002), no mechanical association has been made between sex and magnitudes of the individual tissue masses or the proportions of each between men and women.

Finally, numerous modeling techniques have been used to aide our understanding regarding the transfer of forces through the lower extremity following impact, but as Gruber et al. (1998) demonstrated, not accounting for the effect of soft tissues (or wobbling masses) in these models can lead to significant errors in the predicted forces and moments at more proximal joints. If individual tissue types have a differential effect on impact force attenuation as the shock wave propagates through the segment, then modeling these tissues as separate components in wobbling mass models, to provide insight into possible injury mechanisms, would appear warranted.

Therefore, the purposes of the current study were to (i) investigate the relative effects of body composition and individual leg tissue masses on the acceleration response of the tibia following impact and (ii) determine if leg tissue mass differences between males and females account for tibial shock sex differences reported in the literature.

Methods

Participants

Eighteen healthy male students (mean [SD]), age 22 (3.0) years, body mass 82.3 (0.6) kg, height 1.8 (0.7) m, and BMI 25.8 (6.7) m/kg², and 18 healthy female students (mean [SD]), 20.6 (1.9) years, body mass 67.3 (19.0) kg, height 1.71 (0.07) m, and BMI 23.8 (6.4) m/kg², were recruited from the University of Windsor population to participate in this study. Participants (N = 36) were recruited such that the spectrum of BMIs and percent body fat (BF), from underweight to obese, were represented (BMI of 16 m/kg² to 47 m/kg² and BF of 4% to 37%). They were eligible only if they had not experienced injuries to the back or lower extremities within the last 12 months. All methods and procedures were approved by the University of Windsor’s Research Ethics Board.

Pendulum and Impact Apparatus

A human pendulum, consisting of a rectangular steel frame (190.5 cm × 52.5 cm) and canvas bed suspended from the ceiling by four steel cables (Flynn et al., 2004; Holmes & Andrews, 2006; Duquette & Andrews, 2010a, 2010b), was used to swing participants toward an impact apparatus, such that they impacted it with the unshod heel. This type of apparatus has been used previously in the literature (Flynn et al., 2004; Holmes & Andrews, 2006; Duquette & Andrews, 2010a, 2010b), was used to swing participants toward an impact apparatus, such that they impacted it with the unshod heel. This type of apparatus has been used previously in the literature (Flynn et al., 2004; Holmes & Andrews, 2006; Duquette & Andrews, 2010a, 2010b). Accelerometer signals were filtered close to the source with a fourth-order Bessel switched capacitor low-pass filter with a cut-off frequency of 1.5 Hz.

To ensure consistent ankle postures across conditions and between groups (see Procedures) at impact, ankle angle in the sagittal plane was monitored with a surface mounted accelerometer (AMTI-OR6–6-1000, A-Tech Instruments Ltd., Scarborough, ON, Canada, natural frequency of 1000 Hz) mounted vertically to the impact apparatus, and a linear velocity transducer (Celesco DV301, Don Mills, ON, Canada) attached to the following edge of the pendulum, respectively.

The acceleration response of the proximal tibia was monitored with a surface mounted accelerometer (MMA3201D, Freescale Semiconductor, Inc., Ottawa, ON, Canada; range of ± 50 g; resolution of 0.07 g), with the sensitive axis visually aligned parallel to the long axis of the tibia (Figure 1). The accelerometer was securely attached to the skin overlying the bony area medial to the fibular tuberosity using double-sided tape and a normal preload of approximately 45 N applied with an elastic Velcro strap (Andrews & Dowling, 2000). Peak acceleration, acceleration slope, and time to peak acceleration were determined from the tibial acceleration waveforms. Acceleration slope was measured between 30% and 70% of the peak acceleration (Flynn et al., 2004; Holmes & Andrews, 2006; Duquette & Andrews, 2010a, 2010b). Accelerometer signals were filtered close to the source with a fourth-order Bessel switched capacitor low-pass filter with a cut-off frequency of 1.5 Hz.

Finally, muscle activation of the tibialis anterior and lateral gastrocnemius muscles was collected using bipolar disposable Ag/Ag-Cl surface electrodes (ES40076 Kendall, Mansfield, MA) placed over the muscle bellies of tibialis anterior and lateral gastrocnemius, with an interelectrode distance of 2 cm (Figure 1). Skin was shaved (if necessary) and cleaned with rubbing alcohol.
before attaching the electrodes. The EMG data for each muscle were full-wave rectified, filtered with a second order dual pass Butterworth filter (cut-off frequency of 1.5 Hz) and normalized to each participant’s maximal voluntary exertion (MVE; see Procedures).

**Procedures**

Testing took place in two sessions separated by approximately two days. During the first session, lower extremity anthropometric measurements including lengths, circumferences, breadths, and skinfolds were taken, using standard anthropometric measurement equipment (a soft, flexible measuring tape anthropometer; Lafayette Instrument Company, Lafayette, IN), and skinfold calipers (Slimguide, Creative Health Products; Plymouth, MI C-120). These measurements were used to calculate body composition (i.e., percent body fat [BF]) as per the prediction equations reported by Jackson & Pollock (1978) and Golding et al. (1982) (mean \( r = .907 \); mean \( SE = .006 \) (Table 1). This facilitated the classification of participants into one of three groups (6 males and 6 females in each), based on their BF. Low, medium and high body fat percentages were classified as per Gallagher et al. (2000) (Table 2). Anthropometric measurements were also used as inputs to prediction equations used to calculate local leg tissue masses, including fat mass, lean mass, bone mineral content, and wobbling mass (fat mass + lean mass) of the right leg (Holmes et al., 2005) (Table 1). Participants were grouped according to their specific leg tissue masses to analyze the effects on tibial acceleration. The low, medium, and high groups for these individual tissues were established based on their correspondence with the whole body BF group classifications (Table 2).

During the second session, participants were first instrumented with EMG electrodes and were then asked to execute maximum voluntary exertions (MVEs) for the tibialis anterior and lateral gastrocnemius. The MVEs for the tibialis anterior were collected by asking participants to dorsiflex about the ankle against manual resistance through a range of motion while lying supine on the pendulum apparatus. For the lateral gastrocnemius, participants stood on a wooden platform, to which they were strapped via two 5 cm wide nylon straps that were snugly adjusted over their shoulders. Participants were asked to perform calf raises, with their arms crossed in front of their chest, such that they tried to elevate their bodies against the resistance of the shoulder straps. Three 5 s trials consisting of ramp up, hold, and ramp down phases were performed by each participant for the tibialis anterior and lateral gastrocnemius, and the maximum EMG amplitude achieved during the three trials was taken as the MVE for each muscle. Adequate rest was given between the trials to ensure that fatigue did not occur.

![Figure 1 — Example of a participant’s instrumented right leg while at rest against the vertically mounted force platform. Accelerometer straps and tape are not shown so the position of the instruments can be seen.](image-url)
Once secured to the pendulum, several practice trials were performed to determine the appropriate pull back distance for each participant, such that the velocity and force at impact fell within the ranges of 1.0–1.15 m/s and 1.8–2.8 times body weight, respectively (Flynn et al., 2004; Holmes & Andrews, 2006; Duquette & Andrews, 2010a, 2010b; Lafortune & Lake, 1995).

Data collection was triggered manually following an auditory queue. After the queue, the pendulum was released and all data were simultaneously recorded using a custom LabVIEW software program (National Instruments, Austin, TX, USA, version 7.1) and a 12-bit A/D card (ADI-32, A-Tech Instruments Ltd., Scarborough ON, Canada) at a rate of 4096 Hz.

### Statistical Analysis

Figure 2 outlines the statistical process that was used for data analysis. Separate two-way (sex by body fat) ANOVAs were used to compare the un-normalized mean acceleration responses (peak acceleration, acceleration slope, time to peak acceleration; Figure 2a) and the local tissue masses (fat mass, lean mass, bone mineral content, and wobbling mass; Figure 2b) across the low, medium, and high groups for males and females. Pearson correlations were performed to determine the relationship between the individual leg tissue masses and the acceleration responses. In addition, one-way ANOVAs were used to compare males and females in terms of their

<table>
<thead>
<tr>
<th>Variable</th>
<th>Prediction Equations</th>
</tr>
</thead>
<tbody>
<tr>
<td>Tissue Masses</td>
<td></td>
</tr>
<tr>
<td>Fat mass</td>
<td>$= -927.818 - 140.279(x1) + 44.757(x9) + 29.592(x14)$</td>
</tr>
<tr>
<td>Lean mass</td>
<td>$= -395.1886 + 141.182(x1) + 105.746(x15) - 33.229(x9) + 762.337(x2) + 176.228(x10) +$</td>
</tr>
<tr>
<td></td>
<td>$160.907(x16) + 23.170(x17)$</td>
</tr>
<tr>
<td>Bone mineral content</td>
<td>$= -85.480 + 0.106(x1) + 30131(x7) + 4.155(x8)$</td>
</tr>
<tr>
<td>Wobbling mass</td>
<td>$= -5263.474 - 4.012(x1) + 37.256(x18) + 9.287(x9) + 11.045(x17) + 38.141(x19) +$</td>
</tr>
<tr>
<td></td>
<td>$230.608(x15) + 915.125(x2) + 42.199(x14)$</td>
</tr>
<tr>
<td>Percent Body Fat</td>
<td></td>
</tr>
<tr>
<td>7-site (male)</td>
<td>$= 1.112 - 0.00043499($Sum 7 Skinfolds$) + 0.00000055($Sum 7 Skinfolds$)^2 - 0.00028826($Age$)$</td>
</tr>
<tr>
<td>5-site (female)</td>
<td>$= 0.29731($Sum 5 Skinfolds$) - 0.0053($Sum 5 Skinfolds$)^2 + 0.03037($Age$) - 0.63054$</td>
</tr>
</tbody>
</table>

Note. Where $x1 =$ gender (0 for F, 1 for M); $x2 = $ height (m); $x3 = $ prox. midthigh length (cm); $x4 = $ lat. Thigh length (cm); $x5 = $ ant. midthigh skinfold (mm); $x6 = $ med/lat. midthigh breadth (cm); $x7 = $ participant mass (kg); $x8 = $ prox. midcalf length (cm); $x9 = $ med. midcalf skinfold (mm); $x10 = $ med/lat. midcalf breadth (cm); $x11 = $ prox. thigh circumference (cm); $x12 = $ mid. thigh circumference (cm); $x13 = $ ant/post. midthigh breadth (cm); $x14 = $ knee circumference (cm); $x15 = $ ant/post. midcalf breadth (cm); $x16 = $ malleoli breadth (cm); $x17 = $ lateral leg length (cm); $x18 = $ malleoli circumference (cm); and $x19 = $ ankle circumference.

Sum 7 skinfolds = sum of chest, axilla, triceps, subscapular, abdomen, suprailiac, and thigh skinfold thicknesses.

Sum 5 skinfolds = sum of thigh, suprailiac, abdomen, triceps, and scapula skinfold thicknesses.

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Sum 7 skinfolds = sum of chest, axilla, triceps, subscapular, abdomen, suprailiac, and thigh skinfold thicknesses.

Sum 5 skinfolds = sum of thigh, suprailiac, abdomen, triceps, and scapula skinfold thicknesses.

### Table 1 Summary of the tissue mass prediction equations—fat mass, lean mass, bone mass (bone mineral content) and wobbling mass (lean mass + fat mass) (Holmes et al., 2005)—and the 7-site (Jackson & Pollock, 1978) and 5-site (Golding et al., 1982) percent body fat equations

<table>
<thead>
<tr>
<th>Percent Body Fat</th>
<th>Fat mass (g)</th>
<th>Lean mass (g)</th>
<th>Bone mineral content (g)</th>
<th>Wobbling mass (g)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Male Low</td>
<td>&lt;8</td>
<td>&lt;400</td>
<td>&lt;2500</td>
<td>&lt;220</td>
</tr>
<tr>
<td>Med Low</td>
<td>8–21</td>
<td>400–750</td>
<td>2500–2650</td>
<td>220–230</td>
</tr>
<tr>
<td>High Low</td>
<td>&gt;21</td>
<td>&gt;750</td>
<td>&gt;2650</td>
<td>&lt;240</td>
</tr>
<tr>
<td>Male Med</td>
<td>&lt;21</td>
<td>&lt;750</td>
<td>&lt;1600</td>
<td>&lt;150</td>
</tr>
<tr>
<td>High Med</td>
<td>&gt;33</td>
<td>&gt;1200</td>
<td>&gt;2000</td>
<td>&gt;240</td>
</tr>
<tr>
<td>Female Low</td>
<td>&lt;21</td>
<td>&lt;750</td>
<td>&lt;1600</td>
<td>&lt;150</td>
</tr>
<tr>
<td>High Low</td>
<td>&gt;33</td>
<td>&gt;1200</td>
<td>&gt;2000</td>
<td>&gt;240</td>
</tr>
</tbody>
</table>

<ref>Table 2 Summary of the body composition and individual leg tissue mass group classification parameters. Percent body fat classifications were from Gallagher et al. (2000).</ref>
un-normalized acceleration responses (3 ANOVAs) and local tissue masses (4 ANOVAs) (Figure 2c). To determine the relative contribution of each of the individual tissue masses to shock wave behavior and to ensure an equivalent comparison of acceleration responses between the sexes, the acceleration values were then normalized to the tissue masses by dividing the acceleration responses by each leg tissue mass. Normalization such as this is commonly performed to compare quantities in relative terms, as supported by Bruce et al. (1989). Sex differences in each of the normalized acceleration responses were analyzed with 12 (3 acceleration responses × 4 tissue masses) one-way ANOVAs (Figure 2d).

Data were collapsed across ankle angle and muscle activation levels and the mean acceleration responses of the three impacts were used in all analyses. Alpha was set at .05 for all comparisons and Tukey HSD post hoc tests were performed for any significant main effects and interactions. Finally, post hoc power analyses were performed in the situations where no significant differences were found (G-power, v. 3.1.2, Kiel University, Germany; Faul et al., 2007)

Results

Peak acceleration ($p = .96, 1 – \beta = 0.95$), time to peak acceleration ($p = .66; 1 – \beta = 0.99$), and acceleration slope ($p = .09; 1 – \beta = 0.99$) were not significantly different between the low, medium, and high BF groups (Figure 3). This suggests that whole body composition was not indicative of shock wave behavior.

However, the acceleration responses were found to differ significantly across the localized leg tissue mass groups. Peak acceleration and acceleration slope decreased significantly by 16% ($p = .015$) and 82% ($p = .007$), respectively, from the low to the high BMC groups (Figure 4). Similarly, the acceleration slope was also significantly affected by leg fat mass ($p = .003$), and wobbling mass ($p = .04$). From the low to the high fat mass and wobbling mass groups, acceleration slope decreased from 2415 g/s to 1261 g/s and from 2380 g/s to 1426 g/s, respectively. Time to peak acceleration was affected by wobbling mass ($p = .012$), in that time to peak acceleration significantly increased from 12 ms to 16 ms between the low and high groups, respectively (Figure 4).

(a)  
Low, Med, High  
DV: Percent body fat  
(whole body, unnormalized)  
1 two-way ANOVA (factors: sex, % body fat groups)

(b)  
Low, Med, High  
DVs: Fat mass, lean mass, bone mineral content, wobbling  
(local, unnormalized)  
4 two-way ANOVAs (factors: sex, tissue mass groups)

(c)  
Males vs. Females  
DV: Acceleration responses (unnormalized)  
3 one-way ANOVAs (factor: sex)  
DV: Local tissue masses (local, unnormalized)  
4 one-way ANOVAs (factor: sex)

(d)  
Males vs. Females  
DV: Acceleration response normalized to local tissue masses (local, normalized)  
12 one-way ANOVAs (factor: sex)

Figure 2 — Schematic of the data analysis procedure used showing the progression of determining differences between the body composition groups (a), leg tissue mass groups (b), un-normalized sex differences (c) and normalized sex differences (d). The order of the results also follows this outline. DV = dependent variable.
Figure 3 — Un-normalized acceleration responses (peak acceleration, time to peak acceleration, and acceleration slope) across the low, medium, and high percent body fat groups. Peak acceleration and time to peak acceleration are read from the left axis; acceleration slope is read from the right axis (* represents significant differences; \( p \leq .05 \)).

Figure 4 — Un-normalized acceleration responses (peak acceleration, time to peak acceleration, and acceleration slope) across the low, medium, and high tissue mass (fat mass, lean mass, bone mass (bone mineral content) and wobbling mass) groups. Peak acceleration and time to peak acceleration are read from the left axis; acceleration slope is read from the right axis (* represents significant differences; \( p \leq .05 \)).
The peak acceleration and acceleration slopes were negatively correlated to leg lean mass ($r = -0.511$ to $-0.456$), wobbling mass ($r = -0.465$ to $-0.563$), and bone mineral content ($r = -0.514$ to $-0.585$) (Figure 5). Conversely, time to peak acceleration was significantly correlated with wobbling mass ($r = 0.342$) in the positive direction ($p$-values ranged from $0.001$ to $0.02$). These results suggest that as absolute tissue masses increase, the respective acceleration responses decrease, becoming smaller in magnitude and longer in duration.

Sex differences were found when the overall acceleration responses and tissue masses were compared.
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(Figure 6a). In general, females experienced greater peak acceleration, acceleration slope, and time to peak acceleration compared with males, although a significant difference was only found for the peak acceleration \( (p = .004) \) (Figure 6a). Females had significantly more fat mass than males within the leg segment \( (p = .002) \), while males had significantly more lean mass \( (p = .001; 42\% \text{ and } 26\%) \), respectively (Figure 6b).

When the acceleration responses were normalized to local leg tissue masses, it was found that males experienced significantly less peak acceleration per gram of lean mass \( (p = .001) \) than females \( (0.003 \text{ vs. } 0.006 \text{ g/gram}) \), while females \( (0.01 \text{ g/gram}) \) had significantly lower peak acceleration per gram of fat mass \( (p = .001) \) compared with males \( (0.02 \text{ g/gram}) \). Furthermore, males recorded significantly lower peak acceleration when normalized to bone mineral content \( (p = .002) \) and wobbling mass \( (p = .008) \) \( (38\% \text{ and } 30\%) \) less than females, respectively) (Figure 7).

Males experienced significantly lower normalized acceleration slope with respect to lean mass \( (p = .005) \), bone mineral content \( (p = .002) \), and wobbling mass

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**Figure 6** — Overall sex differences with respect to the three tibial acceleration responses (a) and the absolute leg tissue masses (b). Peak acceleration and time to peak acceleration are read from the left axis; acceleration slope is read from the right axis (* represents a significant sex difference; \( p \leq .05 \)).
Tibial Acceleration Response Following Impact

Normalized to lean mass, acceleration slope for males was approximately 49% less than for females ($p = .005$), while the differences between the sexes were slightly less when normalized to bone mineral content (44%) and wobbling mass (36%) (Figure 7).

Normalized time to peak acceleration was significantly less in males with respect to lean mass ($p = .001$) and bone mineral content ($p = .018$) by approximately 32% and 24%, respectively. Time to peak acceleration normalized to fat mass was significantly greater for males (0.032 ms/g) compared with females (0.016 ms/g) ($p = .011$) (Figure 7).

No significant differences in mean tibialis anterior (37% MVE; $p = .92$; $1 – \beta = 0.99$) or lateral gastrocnemius (17% MVE; $p = .35$; $1 – \beta = 0.99$) muscle activation were found across any of the whole body (BF) groups. Similarly, no significant differences in ankle angle were found across the BF groups ($p = .20$; $1 – \beta = 1.00$).

**Discussion**

Whole body tissue composition, measured as BF, did not affect the tibial acceleration response in this study. However, the composition of rigid and soft tissues measured locally in the leg was found to influence the attenuation of the shock wave through the tibia. Significant tissue mass group differences suggest that as fat mass, wobbling mass, and bone mineral content increase, the acceleration response at the tibia decreases. Finally, females were found to experience a greater peak tibial acceleration per gram of lean mass, wobbling mass and bone mineral content than males, further supporting the need to consider individual roles of the soft tissues in shock wave attenuation.

Tissue masses of the leg were calculated using the validated prediction equations reported by Holmes et al. (2005), and BF was determined using the 7-site (Jackson & Pollock, 1978) and 5-site (Golding et al., 1982) equations for males and females, respectively. These equations provide a quick, affordable, and noninvasive means for quantifying whole body composition and segment tissue mass magnitudes. Support for the use of these specific equations is provided by Burkhart et al. (2008), who reported excellent between- and within-measurer reliability for many of the anthropometric measures used as prediction equation inputs in the current study. Finally, although there are numerous BF prediction equations in the literature, the 7-site (Jackson & Pollack, 1978) and 5-site (Golding et al., 1982) equations were chosen as they incorporate the greatest number of input skinfolds, and are therefore most representative for full body assessments.

Muscle activation levels were monitored in the tibialis anterior and lateral gastrocnemius to ensure that there were no muscle activation differences across the tissue mass groups, as a positive relationship exists between the level of muscle activation and the stiffness of the muscle (Cholewicki & McGill, 1995). In turn, muscle stiffness has a positive relationship with the amount of shock wave transmission through the leg (Nigg & Liu, 1999). However, one of the objectives of this study was to measure the passive effects of the soft tissues while controlling...

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**Figure 7** — Sex differences between the mean peak acceleration, time to peak acceleration, and acceleration slope normalized to tissue mass (fat mass, lean mass, bone mass (bone mineral content) and wobbling mass). Peak acceleration and time to peak acceleration are read from the left axis; acceleration slope is read from the right axis (* represents significant sex differences; $p \leq .05$).
for the active component of force attenuation. The results suggest that tibial acceleration response differences are, to some extent, a function of the composition of the passive soft tissues, as no significant differences in muscle activation levels were found between any of the groups in this study. Similarly, the ankle angle at impact was not statistically different between body fat groups. It has been shown repeatedly (Hamill et al., 1995; Zhang et al., 2000) that joint kinematic strategies can alter the transmission of impact shock waves. Therefore, to focus on the passive tissue effects, it was necessary to control these confounding factors. Finally, while applying impacts with a human pendulum deviates from the normal postures and circumstances of an actual heel impact during gait, it allows for consistent application of impact forces while keeping knee angle constant. In addition, the mean force (2.22 [0.57] times body weight) and velocity (1.10 [0.15] m/s) data collected here compared favorably with past studies that have used a human pendulum as an impact apparatus (Flynn et al., 2004; Holmes & Andrews, 2006; Duquette & Andrews, 2010a, 2010b; Lafortune & Lake, 1995; Lafortune et al., 1996a, 1996b).

Although the magnitudes of individual leg tissue masses were found to be positively correlated with overall body composition, this study highlights the importance of examining structures locally (i.e., at the site of interest), as overall BF was unsuccessful in explaining differences in the magnitudes of the acceleration responses. The majority of studies examining stress fracture risk factors have taken a peripheral (whole body) approach, and have failed to investigate local biological factors. For example, Cline et al. (1998) measured bone mineral density at the wrist and spine while attempting to make inferences regarding the risk of fracture at the tibia and femur. Similarly, Beck et al. (2000) measured thigh muscle mass and distal tibia bone mineral density with respect to stress fractures along the entire tibia. This approach has led to general inconsistencies in the reported risk factors for tibial stress fractures, and makes it difficult to identify the mechanical characteristics leading to injury. In the current study, all anthropometric measurements were taken locally and the findings of decreased acceleration responses are therefore site-specific (i.e., for the leg).

The results presented here also have important modeling implications, as traditional biomechanical models used to analyze impact scenarios in the lower extremity do not consider the modifying effects of the surrounding soft tissues (Gruber et al., 1998; Pain & Challis, 2006). While numerous studies have shown the active force attenuating capabilities of muscle tissue through simulation of muscle stiffness (Chi & Schmitt, 2005), the results presented here indicate that the individual masses of the soft tissues, especially lean mass, also influence the attenuation of impact shock, and do so to a different extent for males and females. While it may be tempting to create biomechanical models using homogeneous soft tissue mass components to reduce model complexity, this study suggests that the roles that separate soft tissues play are different and should be considered when modeling the characteristics of the wobbling mass elements. The different ratios of soft and rigid tissue masses between males and females should also be taken into consideration, based on the results of this study.

With respect to specific tissue masses, lean mass and bone mineral content appear to be the significant contributors to decreased acceleration responses, based on the comparisons between the normalized values of males and females. Numerous studies have cited overall or peripheral decreases in bone mineral density as a risk factor for stress fractures (Beck et al., 2000; Cline et al., 1998), suggesting that bone becomes weaker and more susceptible to the bending forces that are implicated in these types of injuries. While this may be the case, the results presented here also suggest that bone may play a protective role; more bone mineral content may attenuate impact forces to a greater extent. However, more bone mineral content in a segment could be argued to result in higher acceleration response values due to the increased stiffness that greater bone mineral content might provide. Regardless, it appears that bone properties have a profound effect on tibial shock, highlighting the need for further research into this area.

Lean mass was also related to reduced peak acceleration and acceleration slope. Although not significant, the differences in lean mass between the low and high groups resulted in 16% less peak acceleration and 40% less acceleration slope. This provides a possible explanation for the findings of Beck et al. (2000), who reported a significant negative relationship between total calf girth and stress fracture injury risk. The combined effects of bone mineral content and lean mass may also explain why overall health and fitness appear as risk factors in many studies. Bell et al. (2000), Cline et al. (1998), and Shaffer et al. (2006) all found that healthier individuals were less likely to experience a tibial stress fracture, but with little explanation as to why. It is likely that more active or physically fit individuals would have greater leg lean mass and bone mineral content compared with less healthy individuals, which would have a protective effect in terms of stress fracture risk.

Females are up to five times more likely to sustain a lower extremity impact-related stress fracture compared with their male counterparts (Jones et al., 2002). The data presented here also suggest that females are less protected against the effects of impact forces, as a result of having less leg lean mass and bone mineral content compared with males. The normalized lean mass and bone mineral content data generally support these findings, with significantly lower peak acceleration and acceleration slope values (per gram of tissue) in males than females. As tissue masses increased, both males and females overall experienced a decrease in the peak acceleration and acceleration slope, although females experienced a greater rate of decrease across all tissue masses. Furthermore, when the trends in the normalized data are examined together with the correlation coefficients between the acceleration responses and tissue masses, the relationships show that females generally
experienced a greater acceleration response for a similar mass of tissue at low tissue masses. When greater tissue masses are observed, however, sex differences disappear; when comparing the larger tissue masses, the response of females may actually be less than that of males. This may provide insight into the prevention and treatment of impact-related injuries, and has significant fitness and training applications. However, more direct experimental and epidemiological evidence is required to determine the exact effect of specific tissue masses and the possibility of tissue mass ratio (e.g., bone mineral content vs. lean mass) effects on the potential for injury between sexes.

To the authors' knowledge, this is the first study to analyze the effects of the individual leg tissue masses on the acceleration response of the proximal tibia following impact, while controlling for leg muscle activation and ankle angle. The results suggest that lean mass and bone mineral content significantly contribute to decreased acceleration responses. Furthermore, this study provides a possible biological and mechanical explanation for the differences in proximal tibia accelerations between males and females, and highlights the importance of examining the composition of tissues locally, as opposed to relying on more global measures of body composition, when developing biomechanical models for evaluating impacts.

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References


