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# The Effect of Leg Muscle Activation State and Localized Muscle Fatigue on Tibial Response During Impact

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The purpose of this research was to examine the effects of voluntarily manipulating muscle activation and localized muscle fatigue on tibial response parameters, including peak tibial acceleration, time to peak tibial acceleration, and the acceleration slope, measured at the knee during unshod heel impacts. A human pendulum delivered consistent impacts to 15 female and 15 male subjects. The tibialis anterior and lateral gastrocnemius were examined using electromyography, thus allowing voluntary contraction to various activation states (baseline, 15%, 30%, 45%, and 60% of the maximum activation state) and assessing localized muscle fatigue. A skin-mounted uniaxial accelerometer, preloaded medial to the tibial tuberosity, allowed tibial response parameter determination. There were significant decreases in peak acceleration during tibialis anterior fatigue, compared to baseline and all other activation states. In females, increased time to peak acceleration and decreased acceleration slope occurred during fatigue compared to 30% and 45%, and compared to 15% through 60% of the maximum activation state, respectively. Slight peak acceleration and acceleration slope increases, and decreased time to peak acceleration as activation state increased during tibialis anterior testing, were noted. When examining the lateral gastrocnemius, the time to peak acceleration was significantly higher across gender in the middle activation states than at the baseline and fatigue states. The acceleration slope decreased at all activation states above baseline in females, and decreased at 60% of the maximum activation state in males compared to the baseline and fatigue states. Findings agree with localized muscle fatigue literature, suggesting that with fatigue there is decreased impact transmission, which may protect the leg. The relative effects of leg stiffness and ankle angle on tibial response need to be verified.

***Key Words:*** tibial acceleration, wobbling mass, voluntary activation

During running, initial foot-ground contact generates a shock wave that travels through the musculoskeletal system from the feet to the head (Lafortune, Lake, et al., 1996). The wobbling mass (e.g., muscles, soft tissues, and heel pad) passively aids in shock wave attenuation (Chu et al., 1986; Hamill et al., 1995; Lafortune, Lake, et al., 1996), whereas active attenuation is accomplished by manipulating body kinematics, joint positions (Hamill et al., 1995), and muscle activity. Owing to potentially injurious impact forces, a critical function of the human musculoskeletal system is attenuation and dissipation of shock waves (Voloshin et al., 1998).

Nigg & Liu (1999) stated that changes in joint stiffness and the coupling between wobbling and rigid masses was partially due to leg muscle activation. The term *muscle tuning* describes an alteration

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in muscle activation to minimize soft tissue vibrations experienced after impact (Boyer & Nigg, 2004; Nigg & Liu, 1999; Wakeling & Nigg, 2001a, 2001b; Wakeling et al., 2001, 2003). Shock wave vibrations may be decreased through muscle tuning within lower extremity segments. In past studies, the effect of muscle tuning has been determined by soft tissue vibrations and frequency characteristics measured by accelerometers placed on soft tissue packages (Boyer & Nigg, 2004, Wakeling & Nigg, 2001a, 2001b). However, the effect that systematically manipulating muscle activation levels voluntarily has on tibial response (e.g., peak tibial acceleration, time to peak acceleration, acceleration slope) following impact, has yet to be determined.

Muscle fatigue has been described as having the ability to reduce the dampening effect on shock waves (Mizrahi et al., 2000; Voloshin et al., 1998). Clear associations have been found between fatigue (determined by a decrease in pressure of end tidal carbon dioxide [$PETCO_2$]) and an increase in shock waves (measured as accelerations at the knee) (Verbitsky et al., 1998). However, the $PETCO_2$ method accounts only for cardiovascular or whole body fatigue (Mizrahi et al., 2000; Voloshin et al., 1998), and is not appropriate to represent the effect that specific leg muscles, such as lateral gastrocnemius (LG) and tibialis anterior (TA), have on impact attenuation.

In 2001, Christina and colleagues realized that local muscle fatigue must be examined during running to properly assess shock-absorbing capabilities. Localized muscle fatigue was found to reduce the ability to dissipate shock waves in a similar manner to whole body fatigue (Christina et al., 2001). However, their protocol called for subjects to run on a treadmill until localized muscle fatigue of the leg was determined by a decrease in isometric torque. In 2004, Flynn and coworkers used the human pendulum method, similar to Lafortune and Lake (1995), to control initial impact variables. Among other indicators, a decrease of 15% in the mean power frequency (MPF) of the electromyographic (EMG) power density function denoted localized muscle fatigue (Flynn et al., 2004). In contrast to whole body fatigue findings, localized muscle fatigue was said to have caused the muscle to become less stiff, enabling more impact force attenuation. This was measured as reduced peak accelerations and acceleration slopes at the tibial tuberosity (Flynn et al., 2004). This supports results of Pain and Challis (2001), who determined that a softer structure (i.e., wobbling mass) attenuates more force than a rigid structure (i.e., tibia). Findings suggested that examining localized muscle fatigue would enhance understanding of force attenuation properties of the shank.

Therefore, the purpose of this study was to quantify the effect of (1) a range of activation levels and (2) voluntarily induced localized fatigue, of tibialis anterior and lateral gastrocnemius muscles, on tibial response parameters during impact.

## Methods

Thirty right-leg dominant subjects (15 male, 15 female; mean age 22.6 ± 1.4 years) participated in this study. Subjects were excluded if they had medical conditions that would prevent protocol completion, or if they did not participate in weight-bearing activities at least twice per week (subjects refrained from activities within 24 hours of testing to ensure residual fatigue effects did not arise during the protocol). Prior to initiation of the study, subjects were informed of the procedures of the study, which were approved by the Research Ethics Board at the University of Windsor, and a consent form was signed.

The human pendulum apparatus implemented by Flynn et al. (2004) was used in this study. The impact apparatus was a rigid steel frame (152.5 cm × 122 cm × 4 cm) bolted to the ground and wall, consisting of four vertical steel bars affixed to four horizontal steel bars in a gridlike pattern. A force platform (Model OR6-5-1, AMTI, Watertown, MA, USA) was vertically mounted to the impact apparatus, ensuring the axial direction was normal to the surface.

A linear velocity/displacement transducer (Celesco DV301, Don Mills, ON, Canada) was attached to the trailing edge of the pendulum. A miniature uniaxial accelerometer (Model EGA-25-/C, Entran, Fairfield, NJ, USA) was mounted to a small piece of balsa wood (total mass 1.62g) and glued to the skin just medial to the tibial tuberosity on the dominant leg. The accelerometer was preloaded with a strap, using a force of approximately 45 N perpendicular to the tibia shaft (Andrews & Dowling, 2000; Flynn et al., 2004). The sensitive axis was aligned parallel to the longitudinal axis of the tibia. Accelerometer location was marked to ensure accuracy and consistency of placement. Acceleration signals were amplified close to the source (Model IMV-15/15/5VM-/W/L6F, Entran, Fairfield, NJ, USA). Three dependent measures were taken from the acceleration waveform (Figure 1) during impact: peak acceleration, time to peak acceleration, and acceleration slope between 30% and 70% of the rise in acceleration (modified from Flynn et al., 2004).

Data collection took place over two sessions. During the first session, six states of activation (baseline, 15%, 30%, 45%, and 60% of the maximum activation state; and fatigue) were voluntarily induced in the TA. The activation states were tested in the LG during the second session, which took place at least 1 week after the initial session, ensuring no lingering fatigue.

Subjects were asked to lie supine on the pendulum apparatus, with their leg extended and unshod heel just contacting the force platform when at rest. Each subject was strapped to the pendulum by two nylon straps (5 cm wide) across the pelvis and just proximal to the knee of the dominant leg. The nondominant leg was flexed to prevent contact with the force platform. The pendulum apparatus was pulled back and released such that the subject impacted the force platform with their leg in full extension, and their foot dorsiflexed to ensure full heel contact. Velocities at impact were similar to those during running (1.0–1.15 m/s), with corresponding impact forces of 1.8–2.8 times body weight (BW) (Flynn et al., 2004). Impacts were repeated three times for each activation state. The amount of myoelectric activity (EMG) occurring within the muscle during the contraction was examined relative to the EMG recorded during the subject's maximum activation state (100% activation).

The TA maximum activation state was elicited through voluntary contraction while the subject was lying supine on the pendulum apparatus. Manual resistance was applied to the top of the foot by the investigator, as the subject maximally dorsiflexed. During the second session, the LG maximum activation state was collected in a standing position, with a platform placed under the subject's feet. Two pieces of nylon webbing (5 cm) were fed through two platform-mounted eyelets, with one strap tightened over each shoulder of the subject. Verbal encouragement was used during each session to motivate the subject to contract maximally against resistance.

When fatiguing the TA, a resistive rubber band was attached at each side of the impact apparatus. The toes of each subject's impact foot were placed under the rubber band, and they were asked to dorsiflex isometrically against the resistance to a level

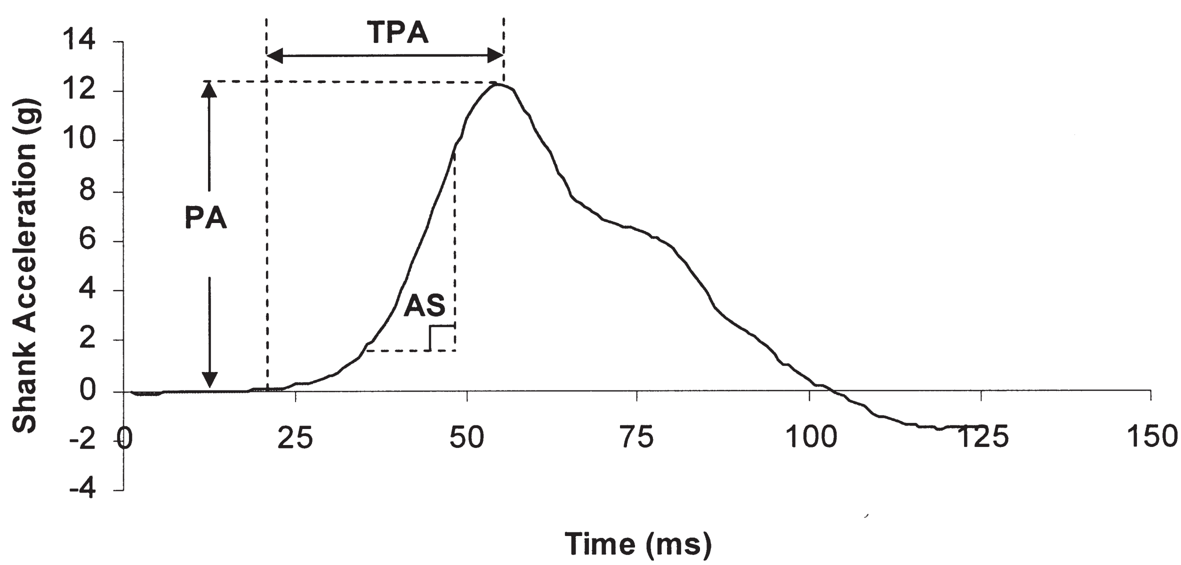


**Figure 1** — Acceleration-time graph with three dependent variables depicted: peak tibial acceleration (PA), time to peak tibial acceleration (TPA), and acceleration slope (AS) between 30% and 70% of the rise in acceleration. (Modified from Flynn et al., 2004.). The acceleration curve has been enlarged to clearly illustrate the dependent variables.

of 50% of their maximum activation state, until fatigued. Inducing localized muscle fatigue in the LG involved securing the pendulum apparatus to ensure minimal foot movement. Subjects performed a voluntary isometric contraction, held and maintained at 50% of their maximum activation state, by plantar flexing their foot against the force platform. Fatigue was determined when the subject could no longer maintain the contraction at 50% of their maximum activation state and when the MPF decreased by at least 15%, in conjunction with muscle trembling and joint angle changes.

Myoelectric signal recordings were taken from the TA and LG. Two Kendall bipolar disposable Ag/AgCl surface electrodes (23 mm × 33 mm: Tyco Healthcare, Mansfield, MA, USA) were placed over the muscle belly in question (intraelectrode spacing of 2 cm) in the direction of their lines of action, after the area was shaved and cleaned with an isopropyl alcohol pad.

Data collection began by manually triggering the software to record the acceleration, velocity, force, and EMG waveforms during the pendulum swing phase (approximately 2 s). Two computers and analog-to-digital (A/D) systems were operated at the same time to collect and process the raw EMG signals for different purposes. With one system, EMG signals were linear enveloped (2nd-order Butterworth filter with cutoff frequency = 1.5 Hz) and displayed on a monitor within the subject's view using custom LabVIEW Software (National Instruments, Austin, TX, USA). Using a line representing the required level of activation on the monitor, subjects were asked to maintain the activation (15%, 30%, 45%, or 60%) as they were impacted into the force platform. This program also calculated and plotted the MPF of the EMG so the investigators could see MPF decreases throughout testing. These EMG data were sampled at 2,048 Hz and filtered at 10–400 Hz. Raw EMG signals were also amplified (× 5,000) and filtered (13–1,000 Hz) with a 2nd-order Butterworth Filter (cutoff frequency = 1.5 Hz; Coulbourn Instruments, Allentown, PA, USA), A/D converted using a 16-bit Power 1401 A/D board at 2,000 Hz, and then stored using Spike 2 software (Cambridge Electronic Design, Cambridge, UK, Version 5, for Windows). The second system (Spike 2 software program and associated hardware) was also used to collect the force plate, accelerometer, and velocity transducer data at 4,000 Hz. The Spike 2 software and hardware were limited in terms of sampling rate and filtering choices so the values that were closest to the LabView system were selected (i.e., 2,000 Hz, and 13–1,000 Hz, respectively).

Raw data files from the Spike 2 software were exported to a spreadsheet for analysis. The peak of the three maximal exertions was used to represent the overall maximum activation state for each muscle. The average myoelectric signal (1 s prior to impact) for all trials was normalized to the subject's maximum activation state, and represented the muscle activation state prior to impact.

During the baseline and fatigue states, subjects were instructed to dorsiflex enough so that the heel impacted the force platform directly. In order to allow sufficient dorsiflexion while still enabling the subjects to generate plantar flexion to the required activation state during Session 2, a stiff nylon strap was used so that subjects could generate plantar flexor muscle force while in a dorsiflexed position. The strap was anchored to the pendulum and was adjusted so that it ran across the ball of the impact foot.

For the impact (velocity, force, EMG) and tibial response parameters (peak acceleration, acceleration slope, time to peak acceleration), mixed design (2 × 2 × 6: Gender × Muscle × Activation State) analyses of variance (ANOVA) were performed. A 2 × 2 × 2 (Gender × Muscle × Level) mixed design ANOVA for the MPF data (where "level" was the MPF prior to or after fatigue) was used. Alpha was set at 0.05, and Tukey HSD post hoc tests were completed on significant main effects and interactions. Omega-squared analyses were performed to determine the total amount of variance accounted for by each experimental treatment. To be included in further analyses, interactions had to account for at least 1% of the total variance (Keppel, 1982).

## Results

Male subjects had significantly greater masses, heights, and body mass indexes (BMIs) than did the female subjects. Not all subjects were able to successfully reach each activation state, with the greatest effect noted at 45% and 60% during Session 2 (LG) (Table 1).

The overall impact force means during each activation state in Session 1 fell within the target range (1.8–2.8 BW), as did impact forces during

**Table 1 Sample Size (*N*) and Number of Subjects Capable of Reaching Activation States During Session 1 (TA) and Session 2 (LG)**

| Subjects | Muscle | Activation state | | | | | |
|---|---|---|---|---|---|---|---|
| | | B | 15% | 30% | 45% | 60% | F |
| *N* Total (Female/Male) | TA | 30(15/15) | 30(15/15) | 30(15/15) | 30(15/15) | 29(15/14) | 30(15/15) |
| | LG | 30(15/15) | 30(15/15) | 29(14/15) | 17(5/12) | 5(1/4) | 30(15/15) |

*Note*. B = baseline, F = fatigue.

Session 2 at the baseline, 15%, and fatigue states. During Session 1, no significant differences were found in the impact velocities; however, all middle activation state velocities were significantly lower than those seen at baseline and fatigue during Session 2. The overall mean peak impact velocity values and the majority of values for each subject fell within the target range (1–1.15 m/s).

Electromyography displayed just prior to impact revealed that all activation states during Session 1 were significantly different from one another; except the baseline, fatigue, and 15% states (Figure 2). During Session 2, all activation states were significantly different from one another; except the baseline and fatigue states. Overall, when asked to contract their TA or LG muscles to a certain percentage of their maximum activation state, subjects were remarkably consistent in accomplishing this task. In addition, subjects were able to minimize the amount of coactivation of the LG and TA (5.4% and 16.1% on average during Sessions 1 and 2, respectively).

A significant decrease in peak acceleration was measured during the fatigued state, when compared to all other activation states during Session 1 (Figure 3a). It can also be noted that during Session 1, although not significant, a small gradual increase in magnitude was seen while increasing from the baseline state to 60% of the maximum activation state. Within Session 2, fairly consistent peak acceleration means were noted across all activation states.

Acceleration slope rose above baseline values at 30% and 45% of the maximum activation state within Session 1 (Figure 3b). This was followed by a significant drop in acceleration slope during the fatiguing state, as compared to 15% through 60% of the maximum activation in the female subjects. Findings were more prominent among females than males (no significant male differences during

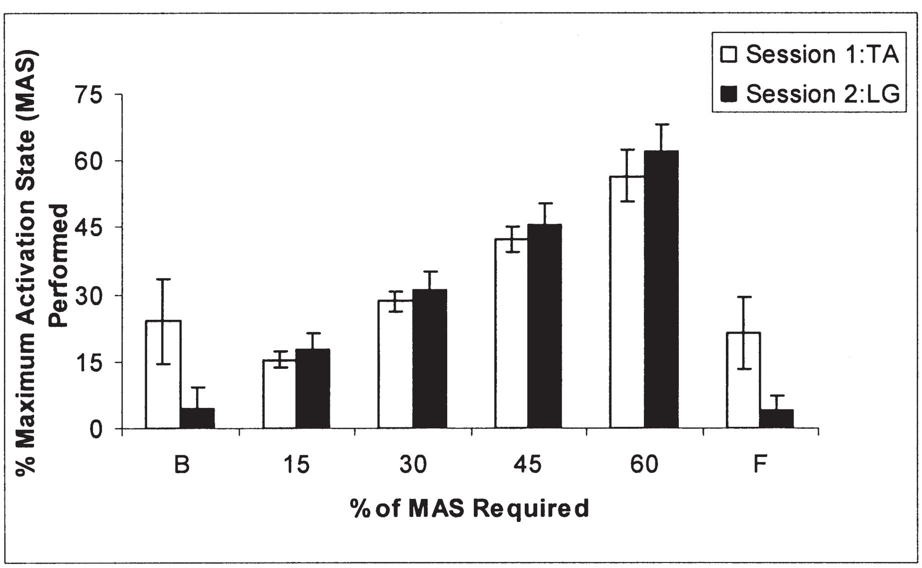


**Figure 2** — Electromyography interaction between muscle and activation state. Mean (± standard deviations)% activation levels for all activation states were significantly different ($p \leq 0.05$) from one another, except for the B, F, and 15% states during Session 1 and the B and F states during Session 2.

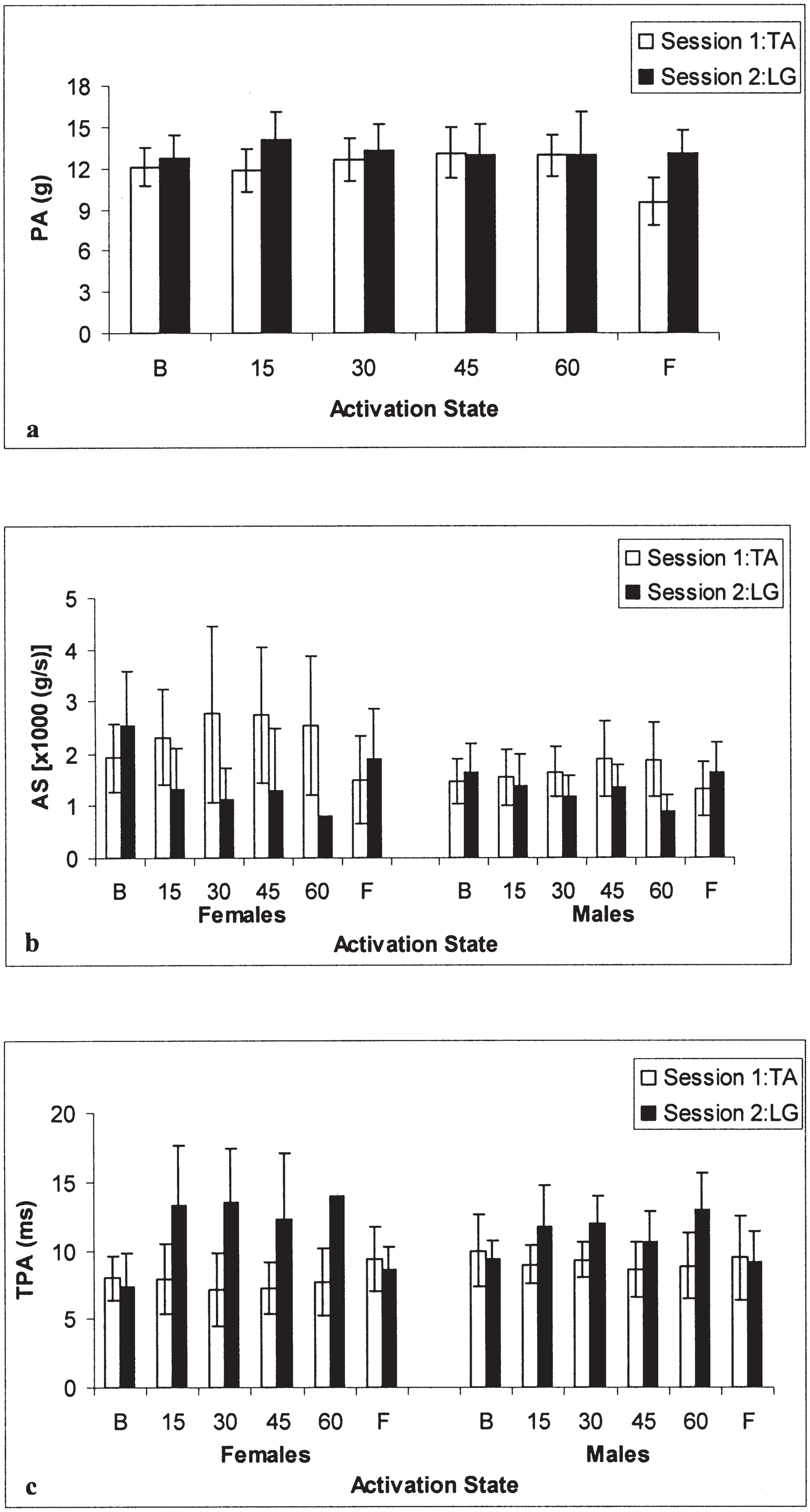


**Figure 3** — Mean (± standard deviations) tibial response parameters. The following pairs or groups of activation states were found to be significantly different from one another ($p \leq 0.05$); please use the notation (e.g., F-B, or F-B/15/30) for each graph: (a) Mean peak tibial acceleration interaction between muscle and activation state. TA: F-B/15/30/45/60. (b) Acceleration slope interaction between gender, muscle, and activation state. Female TA: B-30/45, F-15/30/45/60. Female LG: B-15/30/45/60/F, F-30/60. Male LG: 60-B/F. (c) Time to peak tibial acceleration interaction between gender, muscle, and activation state. Female TA: F-30/45. Female LG: B-15/30/45/60, F-15/30/45/60. Male LG: B-15/30/60, 45-60, F-15/30/60.

Session 1). The second session depicts almost the exact opposite finding: the acceleration slope decreased as activation state increased, with a substantial acceleration slope increase during fatigue. In both the male and female subjects, the baseline and fatigued states were significantly higher than the 60% activation state.

When examining time to peak acceleration, there was an increase at the fatigued state, compared to 30% and 45% in the females during Session 1 (Figure 3c). During Session 2, across gender, the baseline and fatigued states were statistically similar, but were different from all other activation states. In addition, the average time to peak accelerations were much higher during Session 2 (11.3 ms) than in Session 1 (8.6 ms).

Post-hoc tests for level indicated that MPF prior to fatigue (143 Hz) was significantly higher than MPF after fatigue was induced (105 Hz). These data show consistent MPF decreases of approximately 27% on average across each muscle and gender.

## Discussion

The tibial response parameters measured at the knee during heel impact responded to the changing activation states and localized muscle fatigue induced in the TA and LG. Significant decreases in peak acceleration and acceleration slope were noted at fatigue during Session 1. Slight increases in peak acceleration and acceleration slope, and decreased time to peak acceleration with increased TA activation state were also indicated. Subjects were extremely successful in their ability to voluntarily maintain the level of activation required of them during impact.

Altering the activation states within the muscles in question, and then measuring the associated tibial response parameters allowed a closer examination of the transmissibility of impact forces through the leg. Previous authors have examined muscle tuning, and the ability to minimize soft tissue vibrations experienced after impact (Boyer & Nigg, 2004; Nigg & Liu, 1999; Wakeling & Nigg, 2001a, 2001b; Wakeling et al., 2001, 2003). However, in general, the role that the soft tissues, or wobbling masses, play on tibial responses is largely unknown.

Overall, there were significant interactions between muscle and activation state for each tibial response parameter. Peak acceleration values showed slight increases as activation state within each muscle increased; however, the variability within this parameter likely contributed to the lack of significant findings. There was a small, yet steady decrease in time to peak acceleration from the baseline state to 60% of the maximum activation state within Session 1, agreeing with the findings of Pain and Challis (2002) that an increased segment stiffness decreases the time taken to reach peak force (in this case, the time taken to reach peak acceleration). As leg muscles became stiffer through increased activation state, the TA muscle was less equipped to attenuate impact shock, resulting in the decreased time to peak acceleration. A steady rise in acceleration slope as activation state increased from baseline through 60% supports the premise that increasing the stiffness of the leg muscles may create a disadvantage when trying to attenuate impact shock waves. During Session 1, the 30% and 45% activation states displayed significantly greater acceleration slope values than those seen at baseline in the female subjects. The male acceleration slope values were not significant, but the same pattern was seen during Session 1. Session 2 showed opposite findings to those noted in Session 1: for each gender, the acceleration slope decreased and time to peak acceleration increased from baseline to 60% of the maximum activation state. The opposite findings are believed to have occurred due to the strap that was required in Session 2, and not due to any major differences in muscle stiffness between the two muscles (TA in Session 1 and LG in Session 2).

Findings for peak acceleration from this study agree with those of Flynn et al. (2004), in that they decreased significantly between the baseline and fatigue states during Session 1 (from 12.1 g [± 1.4 g] to 9.6 g [± 1.7 g]). This highlights the ability of the fatigued wobbling mass (the TA) to attenuate the impact force traveling toward the knee. Both of these studies localized the fatigue effect to muscle between the point of impact (heel) and the point of segment response recording (proximal tibia), thereby controlling for other variables that change during whole body fatigue protocols while running on treadmills. A decrease of 2.5 g (21%) was seen in this study, an even larger difference than that found in the younger subjects of Flynn et al. (2004) (1.7 g, or 13%).

Acceleration slope showed marked decreases during fatigue (compared to baseline) of approximately 16% and 15% during Sessions 1 and 2, respectively, however, not significant for either gender. The large amount of variability (~30% CV) may have contributed to the lack of significance despite appreciable decreases in acceleration slope. The female acceleration slope values significantly decreased at fatigue, when compared to 15% through 60% of the maximum activation state. Time to peak acceleration increased on average from baseline by approximately 14% during Session 1 in female subjects during localized muscle fatigue, indicating that the leg muscles became less stiff.

There was a significant muscle main effect for peak impact force. On average, the impact force during Session 2 was 2.56 BW, compared to 2.17 BW during Session 1, a difference of approximately 15%. This difference likely occurred when the strap was added during the middle activation states (15% through 60%) in Session 2. This possibly created a more rigid leg structure, thereby increasing impact force. In addition, muscle main effects were noted in each of the tibial response parameters, but the findings are contradictory. The peak acceleration values during Session 2 were on average 10% higher than Session 1. The same trend in acceleration slope would be expected; however, acceleration slope values during Session 1 were on average 22% higher than Session 2. Time to peak acceleration values during Session 2 were on average 24% longer than those observed during Session 1. Time to peak acceleration and acceleration slopes were probably indicative of the experimental setup. The fact that significant peak acceleration results were seen suggests further investigation is warranted. In order to examine the different activation states used in this study, a new strap or restraining device would have to be developed. To examine localized muscle fatigue effects between the two muscles, the experiment could be repeated using only the baseline and fatigued states.

The impact forces and tibial response parameters measured in this study correspond to previous findings of various authors (Table 2). The pendulum impact forces during the baseline state are remarkably close to the values of Flynn et al. (2004) and Lafortune and Lake (1995), as the same impact forces were targeted during both of these studies (1.8–2.8 BW). Peak acceleration, time to peak acceleration, and the acceleration slope all fall within the normal ranges of those reported in the literature. The data from Session 1 in the current study most

**Table 2 Comparison of the Means (± Standard Deviation) of the Tibial Response Parameters in This Study With Previous Work**

| Reference | PF (× BW) | | PA (g) | | TPA (ms) | | AS (g/s) | |
|---|---|---|---|---|---|---|---|---|
| | NF | F | NF | F | NF | F | NF | F |
| Current study, Session 1 | 2.20 (0.25) | 2.15 (0.25) | 12.11 (1.4) | 9.56 (1.7) | 9.0 (2.2) | 9.4 (2.7) | 1703 (549) | 1423 (679) |
| Current study Session 2, | 2.21 (0.23) | 2.30 (0.23) | 12.73 (1.7) | 13.03 (1.8) | 8.4 (1.9) | 8.9 (2.0) | 2095 (801) | 1790 (757) |
| Flynn et al. (2004) Session 1 | 2.35 (0.3) | — | 13.28 (3.7) | 12.09 (3.1) | 10.1 (5.0) | 10.9 (6.0) | 3067 (1488) | 2416 (1363) |
| Flynn et al. (2004) Session 2 | 2.35 (0.3) | — | 13.21 (4.5) | 11.95 (3.5) | 9.7 (2.0) | 10.2 (4.0) | 2843 (1883) | 2589 (1759) |
| Lafortune & Lake (1995) | 2.00 (0.2) | — | 6.40 (0.7) | — | 16.1 (1.7) | — | 671 (220) | — |
| Lafortune et al. (1995) | — | — | 11.20 (3.1) | — | 31.0 (6.0) | — | — | — |
| Lafortune, Lake, et al. (1996) | — | — | 8.80 (2.6) | — | 23.7 (4.1) | — | — | — |
| Lafortune, Hennig, et al. (1996) | — | — | 9.40 (4.0) | — | 22.5 (5.5) | — | 1150 (930) | — |

*Note*. PF = peak impact force; PA = peak tibial acceleration; TPA = time to peak tibial acceleration; AS = acceleration slope; NF = nonfatigued state (baseline); F = fatigued state; Session 1 = TA; Session 2 = LG (for the current study and Flynn et al. [2004]).

closely align with that of Flynn et al. (2004), as a similar experimental apparatus and protocol was used. Time to peak acceleration and acceleration slope values correspond nicely to previous data, with expected increases and decreases, respectively, when fatigued. The values obtained during Session 2, especially for peak acceleration, do not show the same drop from baseline to fatigue, which may be indicative of the amount of dorsiflexion subjects employed during impact.

The human pendulum allowed delivery of consistent impact velocities and forces, and the maintenance of a constant knee angle (of approximately 0°) during impact. However, the dorsiflexion angle used by each subject was not accounted for during impacts. Consistent instruction was given prior to each impact for subjects to dorsiflex their foot just enough to allow direct heel impact with the force platform. However, differences in ankle flexibility, foot shape and size, and foot arch may have contributed to differing foot angles. Future research should address this issue to determine the degree to which ankle angle significantly affects tibial response parameters, relative to fatigue.

The most prominent limitation of this study was the apparatus used for the 15%–60% activation states during LG testing. The strap allowed subjects to generate the required plantar flexor activation, while maintaining enough dorsiflexion to impact the force platform with their heel. The goal of the strap was accomplished, but flatfoot impacts in some subjects resulted in longer times to peak acceleration. Despite a consistent method, the result was higher times to peak acceleration, thus altering the acceleration slopes during the second session. A more rigid device, light and flexible enough to allow subjects to contract to required activations, might ensure more direct heel impacts, and deserves consideration in the future.

The experimental setup in Session 2 (i.e., the strap that was used during the middle activation states [15–60%]) likely contributed to the opposite findings for the TA and LG activation states, and differences may not only be a result of stiffness changes. Future studies should quantify kinematics in depth (specifically during impacts when ankle angle is controlled), and quantify muscle stiffness directly. The strap effects highlighted the need to fully understand the relative contribution of ankle angle and muscle stiffness to the tibial response, as the voluntary contractions may produce changes in stiffness and kinematics.

The fact that not all subjects were able to successfully complete all contractions at the higher activation states (45% and 60% of their maximum activation states) during Session 2 (LG) was not viewed as a major limitation of the study. To verify that the reduced number of impacts during Session 2 did not affect the interpretation of the results, a separate statistical analysis was performed with the data for the 45% and 60% activation states removed. No significant changes to the interpreted results were noted compared to the analysis with the incomplete data included.

In conclusion, this study examined the effect of different activation states, including localized muscle fatigue, on the ability of plantar flexor and dorsiflexor muscles to attenuate forces during heel impact. Subjects were able to consistently contract their muscles to a given percentage of their maximum activation state. When the TA was fatigued, a decrease in peak acceleration and acceleration slope, coupled with an increase in time to peak acceleration suggested that the muscle became less stiff. The reduced stiffness of the wobbling mass will contribute to the dampening of the shock wave caused by impact. It is suggested that the localized muscle fatigue essentially causes the opposite effect of the increasing activation, which creates a more rigid structure that transmits shock waves of greater magnitude, in less time, to the knee.

Substantial increases in impact forces were demonstrated with increasing activation states (from baseline [2.20 BW] to 60% of the maximum activation state [2.41 BW]), and were reflected in slight increases in peak acceleration and acceleration slope, and decreases in time to peak acceleration during Session 1. These results may indicate an increased stiffness in the leg with rising activation states, which may alter the tibial response parameters.

Gender differences were found in tibial response and impact parameters. Future studies may want to match subjects of each gender based not only on strength, but also on leg length, circumference, mass, and heel pad thickness to account for different attenuation capabilities. The conclusions of this study should be reflected in other segments of the body, namely the upper limbs. Different activation levels could be induced to determine the effect on attenuation of impact forces applied in a way that is consistent with an extended arm bracing during falls.

Lastly, the definition of fatigue, based on MPF, deserves further consideration. The consensus indicates that a decrease in MPF of approximately 10–15%, is a significant indicator of fatigue. This study, coupled with findings of Flynn and colleagues (2004) challenge this assumption, as the MPF of nearly every subject declined by 20–30% or greater.

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