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Tradeoffs between impact loading rate, vertical impulse and effective mass for walkers and heel strike runners wearing footwear of varying stiffness

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ABSTRACT

Humans experience repetitive impact forces beneath the heel during walking and heel strike running that cause impact peaks characterized by high rates and magnitudes of loading. Impact peaks are caused by the exchange of momentum between the ground and a portion of the body that comes to a full stop (the effective mass) during the period of the impact peak. A number of factors can influence this exchange of momentum, including footwear stiffness. This study presents and tests an impulse–momentum model of impact mechanics which predicts that effective mass and vertical impulse is greater in walkers and heel strike runners wearing less stiff footwear. The model also predicts a tradeoff between impact loading rate and effective mass, and between impact loading rate and vertical impulse among individuals wearing footwear of varying stiffness. We tested this model using 19 human subjects walking and running in minimal footwear and in two experimental footpads. Subjects walked and ran on an instrumented treadmill and 3D kinematic data were collected. As predicted, both vertical impulse (walking: $F(2,54)=52.0$, $p=2.6\mathrm{E}-13$; running: $F(2,54)=25.2$, $p=1.8\mathrm{E}-8$) and effective mass (walking: $F(2,54)=12.1$, $p=4.6\mathrm{E}-5$; running: $F(2,54)=15.5$, $p=4.7\mathrm{E}-6$) increase in less stiff footwear. In addition, there is a significant inverse relationship between impact loading rate and vertical impulse (walking: $r=-0.88$, $p<0.0001$; running: $r=-0.78$, $p<0.0001$) and between impact loading rate and effective mass (walking: $r=-0.88$, $p<0.0001$; running: $r=-0.82$, $p<0.0001$). The tradeoff relationships documented here raise questions about how and in what ways the stiffness of footwear heels influence injury risk during human walking and running.

1. Introduction

The human foot is subjected to repeated impact forces during walking and heel strike running, evident as visible impact peaks in vertical ground reaction forces. Impact peaks are caused by the inertial change in some portion of the body over a brief period of time, usually during the first 10–50 ms of stance. The generation and attenuation of impact forces have been the focus of much research because their potential role in the etiology of various repetitive stress injuries is unclear and intensely debated (Folman et al., 1986; Collins and Whittle, 1989; Nigg, 2001, 2010; Gill and O'Connor, 2003; Milner et al., 2006; Wen, 2007; Pohl et al., 2009; Daoud et al., 2012). In addition, how footwear affects the generation of impact forces has been heavily investigated because of the perceived role of footwear in mitigating discomfort and preventing injuries that may result from impact peaks (Hume et al., 2008; Nigg, 2010).

During the impact phase of stance, defined as the time period from the onset to the zenith of the impact peak, the impulse of the net external force changes the momentum of some portion of the body.

$$\int_{t_i}^{t_f} (F_z - m_{eff} g)\,dt = m_{eff}(v_f - v_i) \tag{1}$$

where t_i and t_f are the beginning and end times of the impact phase, F_z is the vertical ground reaction force, m_{eff} is the effective mass, g is acceleration due to gravity, and v_i and v_f are the vertical velocities of m_{eff} at t_i and t_f, respectively. We define m_{eff} as the portion of the body's mass that decelerates to zero during the period of the impact peak; m_{eff} therefore may contain mass from the foot, shank, thigh or other body segments (Dempster and Gaughran, 1967; Bobbert et al., 1991; Chi and Schmitt, 2005; Lieberman et al., 2010; Shorten and Mientjes, 2011). We define the impact peak as the first peak in vertical ground reaction force, and it thus contains the summation of both high and low frequency ground reaction forces. (Shorten and Mientjes, 2011).

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While the frequency components of the vertical ground reaction force are important for understanding how the body generates impact peaks, the purpose of this study is to understand how impact peak magnitude (F_{max}), impact loading rate (F'), and vertical impulse, variables that have been implicated in the etiology of various musculo-skeletal injuries, are influenced by footwear heel stiffness. Extensive experimental and modeling studies of the effects of footwear heel stiffness on F_{max} and F' have shown that softer footwear heels decrease F' largely due to increases in the time duration of impact (Δt) rather than changes in F_{max} (Light et al., 1980; Lafortune et al., 1996; Wakeling et al., 2003). Experimental results concerning F_{max} are largely inconclusive, with studies finding that less stiff footwear heels can increase, decrease or have no influence on F_{max} (Clarke et al., 1983; Nigg et al., 1987; Lafortune and Hennig, 1992; Hennig et al., 1993; Wakeling et al., 2003). Modeling studies predict that footwear heel stiffness should decrease F_{max} and that muscle activity in the lower limb can modulate F_{max} (Nigg and Liu, 1999; Zadpoor et al., 2007; Zadpoor and Nikooyan, 2010). Vertical impulse and m_{eff} have been studied in the context of kinematic variation (Chi and Schmitt, 2005; Lieberman et al., 2010), but have yet to be studied in the context of variations in footwear heel stiffness.

We can use the impulse–momentum model (Eq. (1)) to investigate how footwear heel stiffness influences F_{max}, F', m_{eff} and vertical impulse by solving Eq. (1) for m_{eff}

$$m_{eff} = \frac{\int_{t_i}^{t_f} F_z dt}{\Delta v + g\Delta t} \tag{2}$$

Previous experiments using this impulse–momentum model on barefoot individuals have found that m_{eff} varies with gait pattern, kinematics and joint stiffness, and that m_{eff} averages 6.3% of body mass during walking heel strikes and ranges between 2 and 10% of body mass during heel strike running (Chi and Schmitt, 2005; Lieberman et al., 2010). m_{eff} is also expected to change with footwear heel stiffness because a less stiff interface between the foot and ground slows the exchange of momentum between the body and the ground. Decreasing the stiffness of footwear heels while holding all other variables constant increases Δt (Fig. 1) (Light et al., 1980; Nigg et al., 1987; Lafortune et al., 1996; Whittle, 1999; Shorten and Mientjes, 2011), which will result in a greater portion of the body coming to a stop during the period of the impact peak. Thus, less stiff footwear heels will result in larger m_{eff} within a given gait pattern. In turn, a larger vertical impulse will result from the increase in m_{eff} in less stiff footwear (Fig. 1).

The impulse–momentum model makes additional predictions. If less stiff footwear heels decrease F' as reported elsewhere and increase m_{eff} as predicted by the model, then there should be a tradeoff between F' and m_{eff} in walkers and heel strike runners wearing footwear of varying stiffness. Similarly, if less stiff footwear decreases F' and increases vertical impulse as predicted by the model, then there should be also be a tradeoff between these variables in walkers and runners wearing footwear of varying stiffness. The objective of this study is to test these predictions in human walkers and runners wearing footwear of varying stiffness.

2. Materials and methods

2.1. Subjects

Twenty-two healthy adult subjects (13 female – average (SD) body mass (kg): 59.2 (6.63), height (cm): 165 (7.99); 9 male – body mass (kg) 78.9 (7.64), height

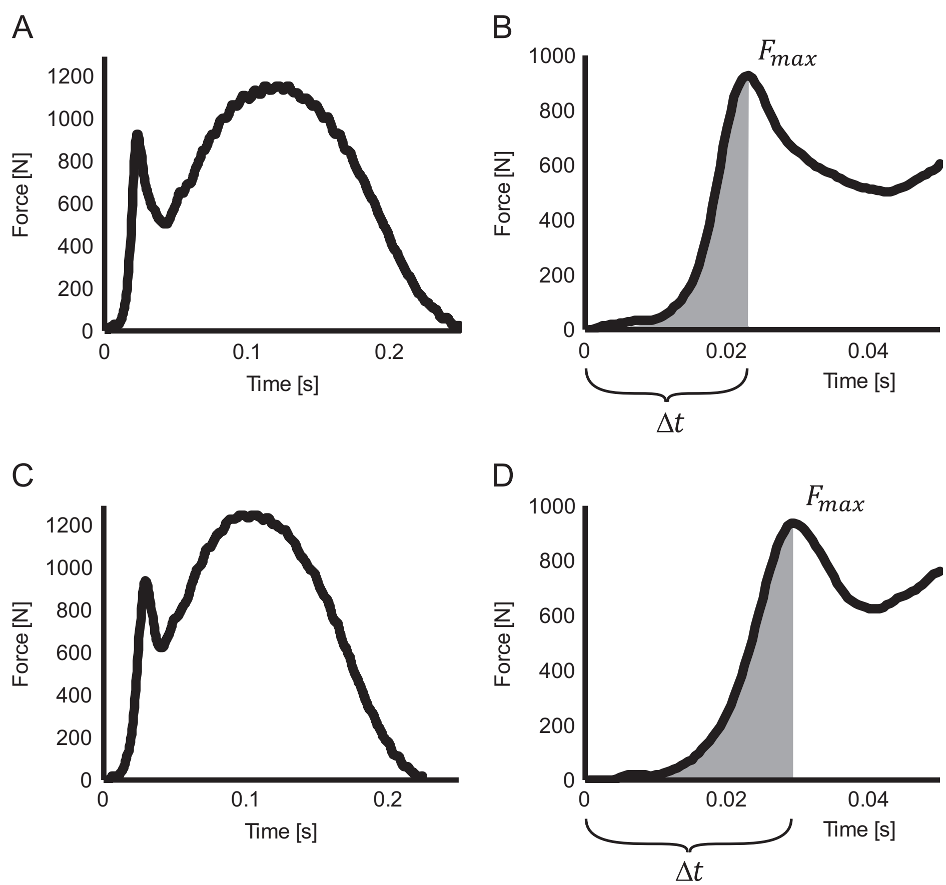

Fig. 1. Vertical ground reaction force of a barefoot heel strike runner on a hard surface (A) and corresponding vertical impact peak emphasized (B). Frames C and D are a vertical ground reaction force for a heel strike runner in soft footwear (C) with the corresponding vertical impact peak emphasized (D). The impact in the less stiff footwear increases Δt, leading to larger m_{eff}. A larger m_{eff} will in turn create a larger vertical impulse (the integral of force over Δt) in the less stiff footwear (shaded area in B vs. D).

(cm) 181 (6.93)) between the ages of 19 and 37 participated in this study, which was approved by the Institutional Review Board of Harvard University. Subjects gave their informed consent to participate and the experiments were conducted at the Skeletal Biology and Biomechanics Lab of the Department of Human Evolutionary Biology at Harvard University.

Criteria for subject inclusion in the data analysis were that the subject was able to wear minimal footwear comfortably and was able to heel strike in both walking and running for the full 30 s trial duration on all footpads (forefoot strikes were not included in the study because they generate no measureable impact peak (Lieberman et al., 2010)). Subjects were asked to heel strike in all walking and running trials, regardless of their natural strike pattern. Of the 22 subjects enrolled in the study, 19 were used for data analysis. In the running analysis, two subjects were removed because they were uncomfortable heel striking for the full trial duration. In walking, one subject was removed because of data collection problems and another was removed because heel strikes were not apparent in one of the trials. An additional subject was removed from walking and running analyses because of discomfort in the minimal footwear.

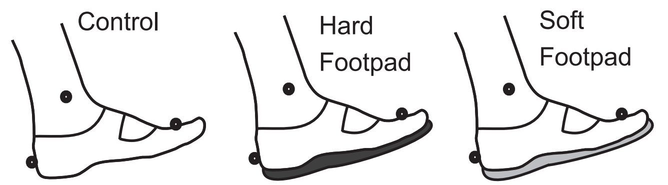

Fig. 2. Schematic of the control, hard footpad and soft footpad conditions used in this experiment. Footpads were attached to the bottom of minimal footwear using duct tape.

2.2. Experimental design and measurements

Subjects walked and ran in minimal footwear (model: M116 Sprint, Vibram USA, Concord MA, USA) and on two different experimental footpads (Fig. 2; 'hard pad': rubber, Product no. RB4000, On Deck Sports, Brockton, MA, USA; 'soft pad': foam, Product no. 150553488-32, Future Foam, Council Bluffs, IA, USA) cut specifically for each subject and attached to the bottom of the minimal footwear using duct tape. Footpads were chosen to decrease the interface stiffness between the foot and the ground – the 'hard' pad was less stiff than the control condition, and the 'soft' pad was less stiff than the 'hard' pad. The order in which these conditions were performed was randomized across subjects. Subjects walked and ran at Froude numbers of 0.28 and 1.2, respectively (actual forward velocities: 1.48 m/s to 1.68 m/s for walking; 3.06 m/s to 3.48 m/s for running). Froude number was controlled to ensure dynamic similarity between subjects that varied in leg length (Alexander, 2003). Before data collection, each subject practiced walking and running on the treadmill at the prescribed Froude number and their preferred step frequency in walking and running was recorded. Each subjects' preferred step frequency was played back via electronic metronome during each trial, and subjects were instructed to keep to this step frequency to the best of their ability to avoid complications with changes in support mechanics with changes in footpad stiffness (Kerdok et al., 2002).

We measured ground reaction forces and lower limb kinematics during the period of the impact peak. The impact peak was defined as the first peak in vertical force. A treadmill instrumented with a force-plate (BERTEC, Columbus, Ohio, USA)

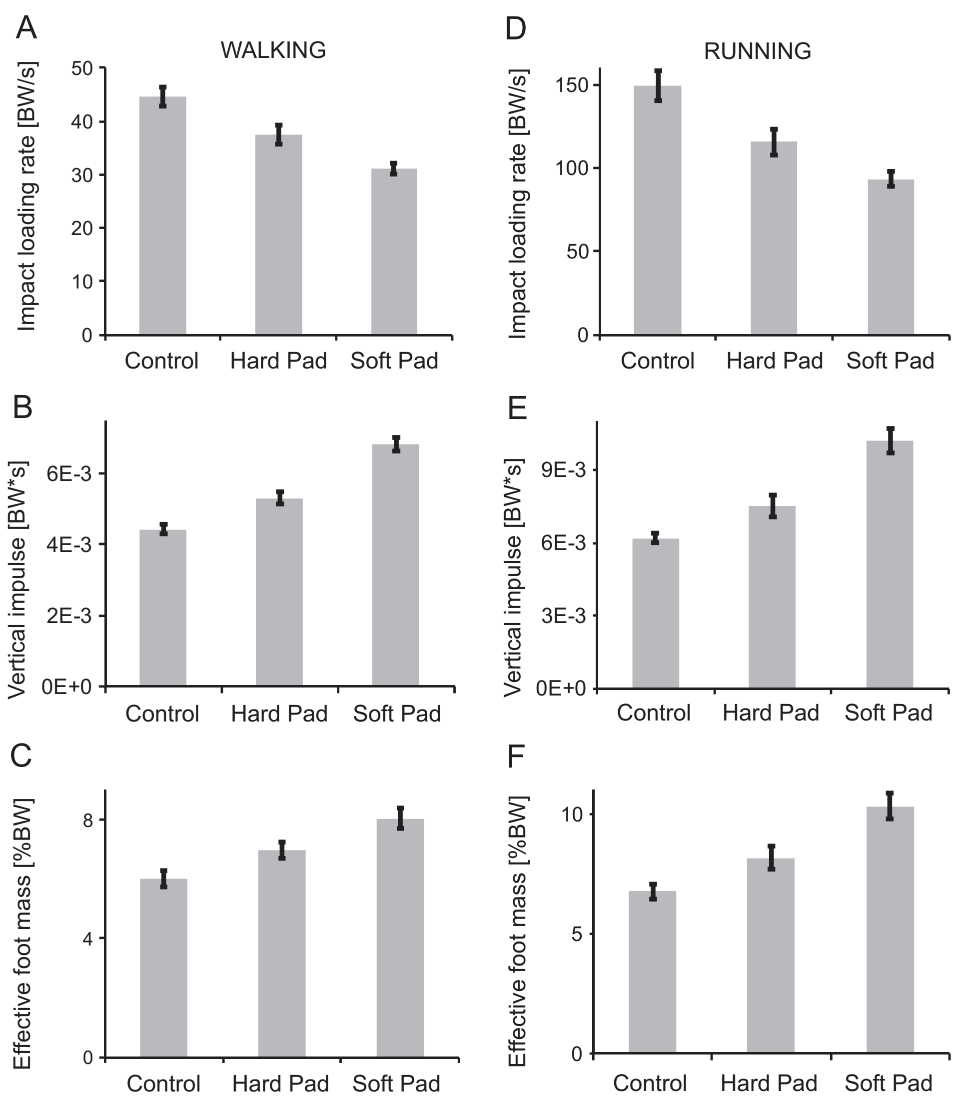

Fig. 3. (A–C) Impact loading rate, vertical impulse and effective foot mass for all three conditions in walking. (D–F) Impact loading rate, vertical impulse and effective foot mass for all three conditions in running. Error bars represent standard error. In each case, the control condition was significantly different from the hard condition and the hard condition significantly different from the soft condition (see text and Table 1 for details).

recorded ground reaction force data at 2 kHz. Kinetic data were low-pass filtered at 100 Hz prior to data analysis. The impact peak was considered to begin when the vertical ground reaction force value exceeded 3 standard deviations above treadmill noise and ended at F_{max}. Lower limb kinematic data were collected at 1000 Hz with an eight camera Oqus system (QUALYSIS, Gothenburg, Sweden). Markers (12.7 cm in diameter) were placed on the skin over the anterior superior iliac spine (ASIS), greater trochanter, medical and lateral femoral condyles, medial and lateral malleoli, the calcaneal tuberosity, and the distal joints of the 5th and 2nd metatarsals.

We measured F_{max}, Δt, impact velocity (Δv) in the vertical direction, as well as knee and ankle angles in the sagittal plane. We considered only the vertical components of the kinetic and kinematic variables because over 90% of the total ground reaction force during the impact phase of gait is due to the vertical force (Cavanagh, 1990). Δv was calculated as the change in lateral malleolus position divided by the change in time for the four frames immediately prior to the beginning of the impact peak. Knee and ankle angles were measured at the beginning of the impact peak and at F_{max}. Knee angle was measured between the greater trochanter, lateral femoral condyle and lateral malleolus markers, and ankle angle was measured between the lateral femoral condyle, lateral malleolus and 5th metatarsal markers. Heel strikes were verified by comparing the plantar foot angle (the angle made between the treadmill horizontal and a line formed between the posterior calcaneus and 5th metatarsal markers) during locomotion to the plantar foot angle made during a standing trial.

We calculated F', m_{eff} and vertical impulse during the period of the impact peak. F' was calculated as F_{max} divided by Δt. Vertical impulse was calculated as the integral of the impact peak over Δt. m_{eff} was calculated using Eq. (2) above. Only the vertical components of force and velocity were used to calculate m_{eff}. In running, we calculated lower extremity stiffness, which was defined as F_{max} divided by the vertical displacement of the greater trochanter.

2.3. Data analyses

Individual steps were removed from the analysis when F_{max} was 3 standard deviations from the average F_{max} for the given subject and condition. We then analyzed 25 steps from the right leg per subject per condition. Averages for all variables were taken from these 25 steps and used in all subsequent analyses.

Regression and ANOVA analyses were performed in MATLAB (v. 2011a, Mathworks, Inc) and JMP Pro 10. We used one-way ANOVA to test how the experimental footpads affected kinematic and kinetic variables, and walking and running trials were considered separately. Bonferroni corrections were used to correct p-values for multiple comparison tests. In all cases that required multiple comparisons, a comparison between each of the three conditions (control, hard pad, soft pad) were made. We used mixed models with subject as a random effect to test relationships between F' and vertical impulse and F' and m_{eff}. For all statistical analyses, significance was assigned to p-values < 0.05.

3. Results

3.1. Effect of footpad stiffness on measured and calculated variables

In both walking and running, F' was significantly different between conditions (Fig. 3A and D; Table 1; walking: $F(2,54)=18.12$, $p=9.5E-7$; running $F(2,54)=15.33$, $p=5.3E-6$). F' was 19% and 29% greater in the control condition than on the hard pad for walking and running, respectively (walking: $p=2.7E-7$; running: $3.4E-6$). F' was 20% and 24% greater on the hard pad than on the soft pad for walking and running, respectively (walking: $p=0.0001$; running: $p=0.0002$).

Vertical impulse was 28% and 35% greater in the soft pad than in the hard pad for walking and running, respectively (Fig. 3B and E; Table 1; walking: $p=3.4E-8$, running: $p=6.7E-8$). Vertical impulse was 20% and 21% greater on the hard pad then in the control condition during walking and running, respectively (Fig. 3B and E; walking: $p=2.6E-6$, running: $p=0.01$). F_{max} was not significantly different between conditions (Table 1; walking: $F(2,54)=0.67$, $p=0.52$, running: $F(2,54)=0.21$, $p=0.81$). Δt was significantly different between conditions (Table 1; walking: $F(2,54)=35.6$, $p=1.4E-10$; running $F(2,54)=33.9$, $p=2.9E-10$). Δt was 13% and 26% longer in the soft pad than in the hard pad for walking and running, respectively (Table 1; walking: $p=2E-8$; running: $p=8.6E-8$). Δt was 21% and 24% longer in the hard pad than in the control condition for walking and running, respectively (Table 1; walking: $p=1E-6$; running: $p=0.0001$).

During both walking and running, m_{eff} (measured in %BW) differed significantly between conditions (Fig. 3C and F; Table 1; walking: $F(2,54)=12.08$, $p=4.6E-5$; running: $F(2,54)=15.52$, $p=4.7E-6$). During walking, m_{eff} averaged 6.0%, 7.0% and 8.1% of body weight in the control, hard footpad and soft footpad

Table 1
The mean and standard deviation of impact kinetic variables (F_{max}, F', vertical impulse, m_{eff}, Δt, Δv, and lower extremity stiffness) for all conditions in both walking and running. P-values are recorded from ANOVA tests between conditions. See text for Bonferroni corrections for multiple comparisons.

	Walking				Running			
	Mean (SD)			p-Value	Mean (SD)			p-Value
	Control	Hard	Soft		Control	Hard	Soft	
F_{max}(BW)	0.80 (0.11)	0.76 (0.11)	0.77 (0.08)	0.52	1.66 (0.29)	1.60 (0.31)	1.65 (0.23)	0.81
F' (BW/s)	44.6 (8.21)	37.4 (7.63)	31.1 (4.27)	9.5E−7	149 (37.4)	116 (34.0)	93.4 (19.6)	5.3E−6
Vertical impulse (BW*ms)	4.4 (0.6)	5.3 (0.7)	6.8 (0.9)	2.6E−13	6.2 (0.9)	7.5 (1.9)	10.2 (2.2)	1.8E−8
m_{eff} (%BW)	6.0 (1.2)	7.0 (1.2)	8.1 (1.4)	4.6E−5	6.8 (1.4)	8.2 (2.0)	10.3 (2.4)	4.7E−6
Δt (ms)	17.9 (2.2)	20.4 (2.7)	24.6 (2.4)	1.4E−10	11.4 (1.5)	14.4 (2.9)	17.9 (2.5)	2.9E−10
Δv (m/s)	0.56 (0.09)	0.56 (0.07)	0.60 (0.08)	0.19	0.80 (0.11)	0.77 (0.09)	0.81 (0.10)	0.47
Lower extremity stiffness (BW/m)	N/A				255 (112)	183 (66.6)	141 (41.3)	3.0E−4

Table 2
The mean and standard deviation of kinematic variables during impact (ankle angle approach, ankle plantarflexion, knee angle approach, knee flexion) for all conditions in both walking and running. P-values are recorded from ANOVA statistical tests. See text for Bonferroni corrections for multiple comparisons.

	WALKING				RUNNING			
	Mean (SD)			p-Value	Mean (SD)			p-Value
	Control	Hard	Soft		Control	Hard	Soft	
Ankle angle approach (deg.)	103 (3.6)	103 (3.9)	103 (3.7)	0.93	97.2 (3.8)	96.8 (4.4)	97.9 (4.8)	0.77
Ankle plantarflexion (deg.)	2.65 (1.0)	3.03 (1.1)	3.41 (1.3)	0.14	3.36 (1.1)	3.85 (1.8)	3.56 (2.6)	0.14
Knee angle approach (deg.)	163 (5.1)	163 (5.1)	163 (5.2)	0.99	156 (3.7)	156 (3.9)	155 (3.5)	0.98
Knee flexion (deg.)	2.88 (0.7)	2.81 (0.9)	3.4 (1.0)	0.13	2.79 (0.8)	3.37 (1.0)	4.42 (1.1)	2.3E−5

conditions, respectively. During running, m_{eff} averaged 6.8%, 8.2% and 10.3% of body weight in the control, hard footpad and soft footpad conditions, respectively. Δv was not significantly different between conditions (Table 1; walking: $F(2,54)=1.68$, $p=0.2$, running $F(2,54)=0.77$, $p=0.47$). Lower extremity stiffness was significantly different between conditions during running (Table 1; $F(2,51)=9.75$, $p=0.0003$). Lower extremity stiffness was 30% greater in the hard pad than in the soft pad, and 39% greater in the control condition than in the hard pad.

During walking, no difference was found between conditions for the ankle angle at the beginning of the impact peak (Table 2; $F(2,51)=0.06$, $p=0.93$) or for the change in ankle angle during the impact peak (Table 2; $F(2,51)=2.07$, $p=0.14$). Also in walking no difference was found between conditions for the knee angle at the beginning of the impact peak (Table 2; $F(2,48)=0$, $p=0.99$) or for the change in knee angle during the duration of impact peak (Table 2; $F(2,48)=2.16$, $p=0.13$).

In running, there was no difference between conditions for the ankle angle at the beginning of the impact peak (Table 2; $F(2,51)=0.27$, $p=0.77$) or the change in ankle angle during the impact peak (Table 2; $F(2,51)=2.07$, $p=0.14$). There was no difference between conditions in knee angle at the beginning of the impact peak (Table 2; $F(2,48)=0.02$, $p=0.98$), but knee flexion angle during the duration of the impact peak was significantly different between conditions (Table 2; $F(2,48)=13.42$, $p=2.3\text{ E}-5$). Subjects had 31% more knee angle flexion when wearing the soft footpad than the hard pad ($p=1.5\text{E}-6$), and 21% more knee angle flexion when wearing the hard pad than in the control condition ($p=0.008$).

3.2. F' vs. vertical impulse and F' vs. m_{eff}

F' varied inversely with vertical impulse in both walking and running (Fig. 4A and B; Table 3; walking: $r=-0.88$, $p<0.0001$; running: $r=-0.78$, $p<0.0001$). F' also varied inversely with m_{eff} in both walking and running (Fig. 4C and D; Table 3; walking: $r=-0.88$, $p<0.0001$; running: $r=-0.82$, $p<0.0001$).

4. Discussion

In this study, we investigated how variations in footwear heel stiffness influenced several aspects of walking and heel strike running impact peaks (including F_{max}, F' and vertical impulse) that have been implicated in the etiology of various repetitive stress injuries. Our study used impulse–momentum mechanics, which models impact events as the exchange of momentum that occurs between the ground and some portion of the body (m_{eff}) over a brief period of time (Δt). It is important to note that the impact peak (as defined above) does not occur instantaneously, and that several portions of the body including the foot, the shank, the thigh and other body segments may contribute to m_{eff}. This means that the impact peaks we examined in this study were composed of both low and high frequency forces (Shorten and Mientjes, 2011). However, the focus of this study was not to examine how footwear influences high and low frequency components of the impact peak, nor was it to investigate contributions of different parts of the body to the total m_{eff}. Rather, our focus was on F', F_{max} and vertical impulse, variables that are often cited as risk factors for several musculo-skeletal injuries (Voloshin et al., 1981; Folman et al., 1986; Collins and Whittle, 1989; Gill and O'Connor, 2003; Davis et al., 2004; Milner et al., 2006; Wen, 2007).

The prediction that both vertical impulse and m_{eff} would increase in walkers and heel strike runners wearing less stiff footwear was

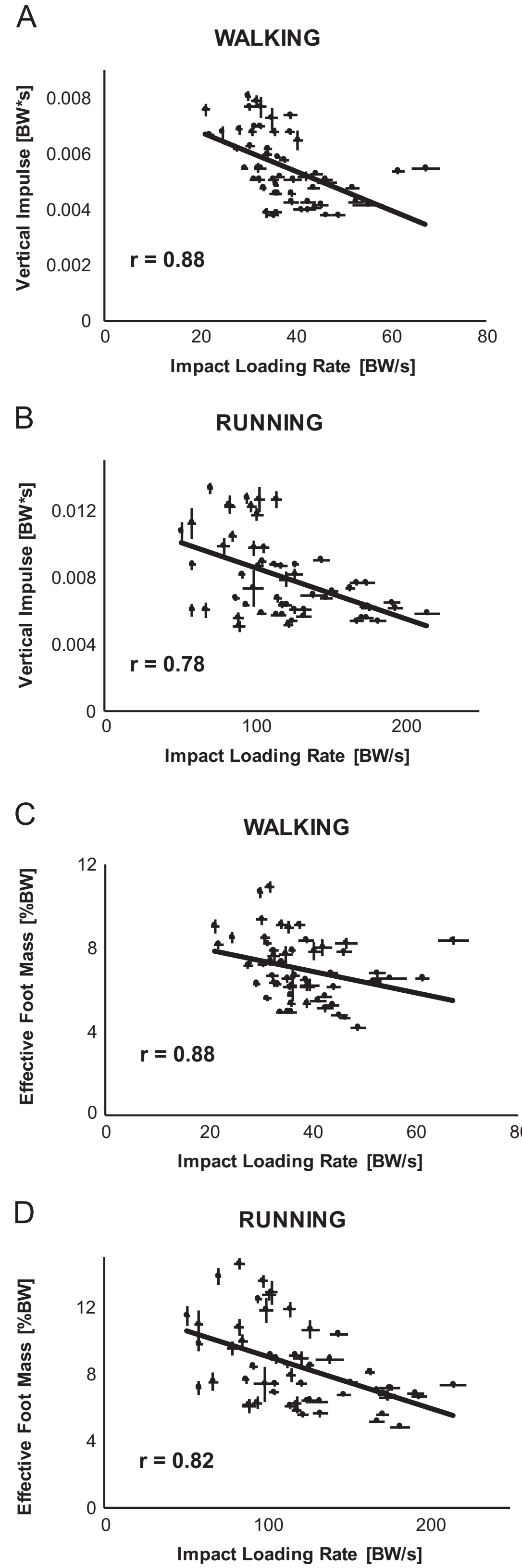

Fig. 4. Repeated measures regression of impact loading rate vs. vertical impulse in walking (A) and running (B). Repeated measures regression of impact loading rate vs. effective foot mass in walking (C) and running (D). Error bars represent standard error. Strong correlations and significant relationships were found for each relationship (see text and Table 3 for details).

Table 3
Correlation coefficients and p-values for the relationships between F' and vertical impulse as well as for F' and m_{eff} for both walking and running.

	Walking	Running
F' vs. vertical impulse	$r=-0.88$; $p<0.0001$	$r=-0.78$; $p<0.0001$
F' vs. m_{eff}	$r=-0.88$; $p<0.0001$	$r=-0.82$; $p<0.0001$

supported by experimental data. Our data also supported the predictions that tradeoff relationships exist between F' and vertical impulse as well as between F' and m_{eff}. In sum, less stiff footwear heels decrease F' while increasing m_{eff} and the vertical impulse of the ground reaction force during the impact phase of walking and heel strike running.

m_{eff} was influenced solely by changes in Δt in the different conditions, and not by changes in F_{max}, Δv or sagittal plane knee and ankle kinematics at the beginning of the impact peak. Knee flexion during Δt for running significantly increased in the softer footpads, likely indicating reduced knee joint stiffness. However, the changes were minimal and unlikely to have had a profound effect on running m_{eff} values because walking m_{eff} changed with condition despite no change in knee or ankle angles during the period of the impact peak (Table 1 and 2).

Our calculation of m_{eff} uses a measure of instantaneous velocity of the lower extremity at impact which we obtained by averaging the displacement of the lateral malleolus over 4 ms prior to impact. Measuring impact velocity at the foot may over- or under-estimate the velocity of the portion of the body that stops during impact. However, any discrepancies that this might cause in absolute values of m_{eff} would be consistent across conditions and would likely have no effect on the differences in m_{eff} we found between conditions. Moreover, our results from control conditions are consistent with previously published data. We found m_{eff} to average 6.0% BW (SD $\pm$ 1.2) in the walking control condition, which is in agreement with the value of 6.3% BW found by Chi and Schmitt (2005) for barefoot walkers. We found m_{eff} during running to average 6.8% BW (SD $\pm$ 1.4) in the control condition, which is greater than the average of 5.3% BW found by Chi and Schmitt (2005)but identical to the value found by Lieberman et al., 2010 for barefoot heel strike runners. One potential reason for the discrepancy between our data and Chi and Schmitt (2005) in heel strike running m_{eff} is that our runners had a forward speed between 3.06 and 3.42 m/s, while runners in Chi and Schmitt (2005) averaged 2.65 (SD $\pm$ 0.44) m/s. F_{max} tends to increase as forward speed increases (Nigg et al., 1987), which is likely indicative of larger m_{eff} at faster running speeds. Although the hypothesis that m_{eff} increases with running speed has not been tested, this would explain why this study and Lieberman et al. (2010) (running speeds between 4 and 6 m/s) found greater m_{eff} than Chi and Schmitt (2005).

This study has several limitations. Instrumented treadmills may influence walking and running kinematics, kinetics and muscle activation compared to embedded force plates (Nigg et al., 1995; Wank et al., 1998). In addition, our definition that the impact peak begins when the vertical force reaches 3SD above treadmill noise (this averaged 25 N across subjects) likely influences the values of Δt, Δv, F', m_{eff} and the vertical impulse. For example, increasing the threshold for the beginning of the impact peak would decrease both Δt and Δv. Despite these methodological limitations, our values for F_{max}, Δv, Δt and m_{eff} are comparable to other studies that have used embedded force plates or that have slightly different definitions for the beginning of impact (Gill and O'Connor, 2003; Chi and Schmitt, 2005; Lieberman et al., 2010), suggesting our methodology does not confound our results.

Further, F' in previous studies has been measured using smaller force intervals (e.g. from 200 N to 90% of F_{max} (Williams et al., 2000; Lieberman et al., 2010)). The method of calculating F' in this study includes the toe and peak regions of the impact peak, where the rate of change of force is not constant. We measured F' using this method because F_{max} during walking often reached no greater than 400 N for some individuals. We would have had low temporal and spatial resolution for measuring F' and foot motion had we chosen a smaller force interval.

An additional limitation of this study is that the footpads necessarily added mass to the subjects' feet. While an ideal experiment would have used experimental footpads of equal mass, we think it is unlikely that the observed changes in m_{eff} were due to the actual mass added by the footpads or differences in mass between the footpads. The hard and soft footpads had an average mass of 0.12 kg and 0.024 kg across subjects, respectfully. The change in m_{eff} from the hard pad to the control condition was 0.97% and 1.4% of body weight in walking and running, respectively. These percentages are equivalent to 0.63 and 0.91kg in a 65 kg individual, respectively, which are much greater than the mass of the hard pad alone. Further, the soft pad condition resulted in greater m_{eff} than the hard pad despite having lower mass than the hard pad. Therefore, it is improbable that the mass of the footpads had more than a negligible effect on changes in m_{eff} observed in this study.

This study was designed to test predictions about how impact kinetics change as a result of variations in footwear heel stiffness. An important implication is that the tradeoff relationship between F' and vertical impulse in the experimental footwear used in this study likely also exists for walkers and runners using any kind of footwear. Because both F' and vertical impulse have been hypothesized to be risk factors for some repetitive stress injuries, walking or heel strike running in less stiff footwear heels may decrease injury risk from impact loading rates but increase injury risk from larger vertical impulses. This hypothetical trade-off has yet to be tested, but merits further investigation in order to better understand the effects of different types of footwear heels. Future work in this area should consider how common shoe materials influence F' and vertical impulse, as well as investigate the relationship between specific repetitive stress injuries and elevated values of F' and vertical impulse. A related question is whether larger vertical impulses due to extended impact time durations (as found in this study) have the same effect on skeletal tissues as large vertical impulses due to elevated F_{max}.

An important follow-up question deriving from these results is how muscles modulate the way impact forces are generated and dampened. Lower limb muscle activity changes around the time of impact when individuals walk and run in footwear of varying stiffness (Wakeling and Nigg, 2001; Wakeling et al., 2002, 2003). These changes may occur in order to modulate F_{max} or to reduce vibrations of soft tissues (Wakeling et al., 2003; Zadpoor and Nikooyan, 2010). It is unknown, however, how these changes in muscular activity influence the vertical impulse and m_{eff}. Also, models that use muscle activity to explain experimental findings concerning impact kinetics would benefit from incorporating changes in m_{eff} and lower extremity stiffness documented here (Zadpoor and Nikooyan, 2010). Changes in muscle activity also function to increase damping when heel striking in less stiff footwear (Wakeling et al., 2003), but it remains unclear how changes in muscle activation influence m_{eff} or the vertical impulse.

Conflicts of interest

The authors have no known conflicts of interest.

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References

Alexander, R., 2003. Principles of Animal Locomotion. Princeton University Press, Princeton, NJ.
Bobbert, M.F., Schamhardt, H.C., et al., 1991. Calculation of vertical ground reaction force estimates during running from positional data. J. Biomech. 24 (12), 1095–1105.
Cavanagh, P.R., 1990. Biomechanics of distance running. Human Kinetics Publishers, Champaign.
Chi, K.J., Schmitt, D., 2005. Mechanical energy and effective foot mass during impact loading of walking and running. J. Biomech. 38 (7), 1387–1395.
Clarke, T.E., Frederick, E.C., et al., 1983. Effects of shoe cushioning upon ground reaction forces in running. Int. J. Sports Med. 4 (4), 247–251.
Collins, J.J., Whittle, M.W., 1989. Impulsive forces during walking and their clinical implications. Clin. Biomech. 4 (3), 179–187.
Daoud, A.I., Geissler, G.J., et al., 2012. Foot strike and injury rates in endurance runners: a retrospective study. Med. Sci. Sports Exercise 44 (7), 1325–1334.
Davis, I., Milner, C.E., et al., 2004. Does increased loading during running lead to tibial stress fractures? A prospective study. Med. Sci. Sports Exercise 36 (5) (S58-S58).
Dempster, W.T., Gaughran, G.R., 1967. Properties of body segments based on size and weight. Am. J. Anat. 120 (1), 33–54.
Folman, Y., Wosk, J., et al., 1986. Cyclic impacts on heel strike – a possible biomechanical factor in the etiology of degenerative disease of the human locomotor system. Arch. Orthop. Trauma Surg. 104 (6), 363–365.
Gill, H.S., O'Connor, J.J., 2003. Heelstrike and the pathomechanics of osteoarthrosis: a pilot gait study. J. Biomech. 36 (11), 1625–1631.
Gill, H.S., O'Connor, J.J., 2003. Heelstrike and the pathomechanics of osteoarthrosis: a simulation study. J. Biomech. 36 (11), 1617–1624.
Hennig, E.M., Milani, T.L., et al., 1993. Use of ground reaction force parameters in predicting peak tibial accelerations in running. J. Appl. Biomech. 9 (4), 306–314.
Hume, P., Hopkins, W., et al., 2008. Effectiveness of foot orthoses for treatment and prevention of lower limb injuries – a review. Sports Med. 38 (9), 759–779.
Kerdok, A.E., Biewener, A.A., et al., 2002. Energetics and mechanics of human running on surfaces of different stiffnesses. J. Appl. Physiol. 92 (2), 469–478.
Lafortune, M.A., Hennig, E.M., 1992. Cushioning properties of footwear during walking – accelerometer and force platform measurements. Clin. Biomech. 7 (3), 181–184.
Lafortune, M.A., Hennig, E.M., et al., 1996. Dominant role of interface over knee angle for cushioning impact loading and regulating initial leg stiffness. J. Biomech. 29 (12), 1523–1529.
Lieberman, D.E., Venkadesan, M., et al., 2010. Foot strike patterns and collision forces in habitually barefoot versus shod runners. Nature 463 (7280), 531–535.
Light, L.H., McLellan, G.E., et al., 1980. Skeletal transients on heel strike in normal walking with different footwear. J. Biomech. 13 (6), 477–480.
Milner, C.E., Ferber, R., et al., 2006. Biomechanical factors associated with tibial stress fracture in female runners. Med. Sci. Sports Exercise 38 (2), 323–328.
Nigg, B.M., 2001. The role of impact forces and foot pronation: a new paradigm. Clin. J. Sport Med. 11 (1), 2–9.
Nigg, B.M., 2010. Biomechanics of Sport Shoes. University of Calgary, Calgary, Alberta, Canada.
Nigg, B.M., Bahlsen, H.A., et al., 1987. THE influence of running velocity and midsole hardness on external impact forces in heel toe running. J. Biomech. 20 (10), 951–959.
Nigg, B.M., Deboer, R.W., et al., 1995. A kinematic comparison of overground and treadmill running. Med. Sci. Sports Exercise 27 (1), 98–105.
Nigg, B.M., Liu, W., 1999. The effect of muscle stiffness and damping on simulated impact force peaks during running. J. Biomech. 32 (8), 849–856.
Pohl, M.B., Hamill, J., et al., 2009. Biomechanical and anatomic factors associated with a history of plantar fasciitis in female runners. Clin. J. Sport Med. 19 (5), 372–376.
Shorten, M., Mientjes, M.I.V., 2011. The 'heel impact' force peak during running is neither 'heel' nor 'impact' and does not quantify shoe cushioning effects. Footwear Sci. 3 (1), 41–58.
Voloshin, A., Wosk, J., et al., 1981. Force wave transmission through the human locomotor system. J. Biomech. Eng.-Trans. ASME 103 (1), 48–50.
Wakeling, J.M., Liphardt, A.M., et al., 2003. Muscle activity reduces soft-tissue resonance at heel-strike during walking. J. Biomech. 36 (12), 1761–1769.
Wakeling, J.M., Nigg, B.M., 2001. Soft-tissue vibrations in the quadriceps measured with skin mounted transducers. J. Biomech. 34 (4), 539–543.
Wakeling, J.M., Pascual, S.A., et al., 2002. Altering muscle activity in the lower extremities by running with different shoes. Med. Sci, Sports Exercise 34 (9), 1529–1532.
Wank, V., Frick, U., et al., 1998. Kinematics and electromyography of lower limb muscles in overground and treadmill running. Int. J. Sports Med. 19 (7), 455–461.
Wen, D., 2007. Risk factors for overuse injuries in runners. Curr. Sports Med. Rep. 6 (5), 307–313.
Whittle, M.W., 1999. Generation and attenuation of transient impulsive forces beneath the foot: a review. Gait Posture 10 (3), 264–275.
Williams, D.S., McClay, I.S., et al., 2000. Lower extremity mechanics in runners with a converted forefoot strike pattern. J. Appl. Biomech. 16 (2), 210–218.
Zadpoor, A.A., Nikooyan, A.A., 2010. Modeling muscle activity to study the effects of footwear on the impact forces and vibrations of the human body during running. J. Biomech. 43 (2), 186–193.
Zadpoor, A.A., Nikooyan, A.A., et al., 2007. A model-based parametric study of impact force during running. J. Biomech. 40 (9), 2012–2021.