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Original article

Impact shock frequency components and attenuation in rearfoot and forefoot running

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Abstract

Background: The forefoot running footfall pattern has been suggested to reduce the risk of developing running related overuse injuries due to a reduction of impact related variables compared with the rearfoot running footfall pattern. However, only time-domain impact variables have been compared between footfall patterns. The frequency content of the impact shock and the degree to which it is attenuated may be of greater importance for injury risk and prevention than time-domain variables. Therefore, the purpose of this study was to determine the differences in head and tibial acceleration signal power and shock attenuation between rearfoot and forefoot running.
Methods: Nineteen habitual rearfoot runners and 19 habitual forefoot runners ran on a treadmill at 3.5 m/s using their preferred footfall patterns while tibial and head acceleration data were collected. The magnitude of the first and second head acceleration peaks, and peak positive tibial acceleration were calculated. The power spectral density of each signal was calculated to transform the head and tibial accelerations in the frequency domain. Shock attenuation was calculated by a transfer function of the head signal relative to the tibia.
Results: Peak positive tibial acceleration and signal power in the lower and higher ranges were significantly greater during rearfoot than forefoot running ($p < 0.05$). The first and second head acceleration peaks and head signal power were not statistically different between patterns ($p > 0.05$). Rearfoot running resulted in significantly greater shock attenuation for the lower and higher frequency ranges as a result of greater tibial acceleration ($p < 0.05$).
Conclusion: The difference in impact shock frequency content between footfall patterns suggests that the primary mechanisms for attenuation may differ. The relationship between shock attenuation mechanisms and injury is not clear but given the differences in impact frequency content, neither footfall pattern may be more beneficial for injury, rather the type of injury sustained may vary with footfall pattern preference.
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Keywords: Frequency domain; Impact shock; Running footfall patterns; Shock attenuation; Tibial acceleration

1. Introduction

Vertical impact variables, such as the magnitude and rate of the vertical impact peak and impact shock, have long been at the center of the running injury debate. The forefoot (FF) and midfoot (MF) running footfall patterns have recently been associated with lower rates of running injuries compared with rearfoot (RF) running.[1,2] The absence or reduction of the vertical ground reaction force (GRF) impact peak in FF and MF running has been the suggested explanation for these findings. However, impact variables, such as characteristics of the vertical GRF and impact shock, have been related to injury in some studies (e.g., Refs. 3–5) but not others (e.g., Refs. 6–8). For example, one study found a lower relative injury frequency in those considered to have high vertical impact force magnitudes or loading rates compared with individuals

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considered to have low vertical impact force magnitudes or loading rates.[9] Other vertical GRF variables, such as the active peak magnitude, may also be related to the development of running injuries[10−12] but this aspect has been virtually ignored in the running injury debate. One thing remains clear: running injuries develop because of complex interactions between many variables, regardless of footfall pattern. Further examination of impact related variables may reveal that the joints or tissues susceptible to injury may differ between footfall patterns.

The events surrounding the foot-ground collision during running are the main source of the impact shock that is transmitted through the leg and the rest of the body. This impact shock is closely related to vertical GRF characteristics and running kinematics.[13−17] Anything that affects segment velocity the instant before initial contact, such as running speed, stride frequency, and joint orientation, will determine the change in momentum of the foot and leg at initial contact and thus the magnitude and rate of the vertical impact peak and impact shock.[14,18−20] The frequency content of the impact shock will depend on the magnitude and timing of the vertical GRF.[13] Given the differences in vertical GRF characteristics and kinematics between footfall patterns, the impact shock resulting from each footfall pattern may exhibit different frequency content. The frequency content of impact parameters may be a significant contributor to running related injuries because the capacity of different tissues and mechanisms to transmit and attenuate the impact shock may be frequency dependent.[21]

The frequency content and signal power of the impact shock and tibial acceleration during stance are determined primarily by the acceleration of the leg segments and whole body center of mass (COM).[13] Specifically, the tibial acceleration profile in RF running contains a lower frequency range (4−8 Hz) representing voluntary lower extremity motion and the vertical acceleration of the COM during the stance phase and a higher frequency range (10−20 Hz) representing the rapid deceleration of the foot and leg at initial ground contact.[13−15,17,22] These lower and higher frequency ranges are also representative of the active peak and impact peak of the vertical GRF, respectively.[13,17] In the time domain, the existence of a prominent impact peak in RF running but a greater active peak magnitude in FF running[10,23,24] suggest that the signal power contained in these lower and higher frequency ranges may differ between footfall patterns and may also affect how these frequencies are attenuated.

The impact shock must be attenuated to prevent the disruption of the vestibular and visual systems as a result of excessive head acceleration.[14,15,22,25,26] Attenuation occurs primarily through energy absorption from active muscles, changes in joint geometry, and deformation of passive tissues.[27−31] The body responds to greater impact magnitudes by increasing attenuation through a combination of active and passive mechanisms.[12,30] The reliance on certain shock attenuation mechanisms may depend on the frequency content of the impact shock. Passive mechanisms, such as deformation of the heel fat pad, the running shoe, ligaments, bone, muscle oscillation, and articular cartilage are responsible for attenuating the higher frequency waveforms generated at initial ground contact.[27−31] Pre-activation of muscle will change to increase damping of impact shock frequencies greater than 40 Hz.[32] However, muscle contractions specifically responding to the impact stimulus and some other attenuation mechanisms may only be effective at attenuating frequencies below 10 Hz because of muscle latency periods.[29,33] Active shock attenuation mechanisms include eccentric muscle contractions, increased muscle activation, changes in segment geometry, and adjustments in joint stiffness.[14,32,34−38] However, the body may have a reduced capacity for attenuating lower frequency components.[14,26] The capacity and degree of attenuation will be dictated by the frequency content of the impact shock and the mechanisms available for attenuation. A reduced capacity for attenuation by some tissues or mechanisms may result in a greater reliance on other tissues or mechanisms and could potentially result in a tissue becoming overloaded.[28,39,40]

Differences in impact parameters between RF and FF running have only been examined in the time domain to our knowledge. However, it may be important to examine impact parameters in the frequency domain because differences in the frequency content of the impact shock may alter the reliance on specific shock attenuation mechanisms in RF versus FF running and the degree of attenuation that occurs. A recent study found that RF running resulted in a greater percent difference in peak acceleration between the head and tibia signals in the time domain than FF running.[41] That study was an excellent first step investigating shock attenuation between footfall patterns using a transfer function in the time domain determine shock attenuation. However, given that frequency content dictates shock transmissibility,[21] important information may be lost regarding attenuation of specific frequency components and the mechanisms used for attenuation when using a time domain analysis.

Time domain differences in kinematics and vertical GRF characteristics between footfall patterns suggest that the impact shock may contain different frequency domain characteristics that are dictated by these kinematic and kinetic events. Specifically, the presence of the vertical GRF impact peak in RF running and greater vertical GRF active peak in FF running may result in differences in signal power of the higher and lower frequency ranges of the impact shock and the degree that shock is attenuated. Therefore, the purpose of this study was to determine the difference in the frequency content of the impact shock and its subsequent attenuation between footfall patterns. It was hypothesized that RF running would result in greater peak tibial acceleration and signal power in the higher frequency range, representative of the vertical GRF impact peak, compared with FF running whereas tibial acceleration power in the lower frequency range, representative of the vertical GRF active peak, would be greater in FF than in RF running. Although RF running results in greater tibial acceleration than FF running,[23] head acceleration may be similar because shock attenuation increases in response to greater impact loads to maintain head stability for proper vestibular and visual function.[14,17,22,26] Therefore, it was hypothesized that peak head acceleration and signal power in the lower and higher frequency ranges would not differ between footfall patterns. As a result of the previous observation that impact shock was greater with RF than FF running,[23] it was hypothesized that RF running would result in greater shock attenuation of the higher range frequency components than FF running. However, previous studies have indicated a reduced capacity for attenuation of lower frequency components,[14,26] therefore it was hypothesized that no difference would be observed in the degree of attenuation of the lower frequency components between footfall patterns.

2. Materials and methods

2.1. Participants

Nineteen habitual RF runners and 19 habitual FF runners participated in this study (Table 1). Sample size estimation determined that 12 runners per group were required to achieve a power of 0.8 and an alpha level of 0.05. All participants were healthy, experienced runners and did not have a history of cardiovascular or neurological problems. Inclusion criteria required that participants completed a minimum of 16 km/week at a minimum preferred running speed of 3.5 m/s and had not developed an injury to the lower extremity or back within the past year.

Participants were divided into an RF group or an FF group based on the footfall pattern habitually performed when distance running. The participants' habitual footfall pattern was determined by assessing the strike index, vertical GRF profile, and sagittal plane angle ankle at touchdown while the participants ran at his or her preferred speed over a force platform (OR6-5; AMTI, Watertown, MA, USA).[42] Given that approximately 20%−25% of runners are either MF or FF runners, participants classified as either MF or FF were place in the FF group to ensure appropriate statistical power. All participants read and completed an informed consent document and questionnaires approved by the University Institutional Review Board.

2.2. Experimental setup

Three-dimensional kinematics of the right leg and foot were recorded using a 9-camera high-resolution motion capture system (Qualysis Inc., Gothenburg, Sweden) at 240 Hz to verify the footfall pattern performed by each participant. Kinematic data collection procedures and reflective marker placement are described elsewhere.[43] Low-mass (<4 grams), uniaxial, piezoelectric accelerometers (ICP®; PCB Piezotronics, Depew, NY, USA) were attached to the center of the forehead and the distal anteromedial aspect of the tibia.[22] Each attachment site was chosen to reduce the effects of soft tissue vibration.[44] The axis of each accelerometer was aligned with the vertical axis of the lower leg while the participant was standing. The vertical axis of the lower leg was aligned with the vertical axis of the laboratory coordinate system. The accelerometers were sampled at 1200 Hz and voltage was amplified by a factor of 10. Lower extremity motion and accelerometer data were collected synchronously.

2.3. Protocol

Participants wore neutral racing flats (RC 550; New Balance, Brighton, MA, USA) provided by the laboratory. Accelerometers were secured to the head and distal anteromedial tibia by rubber straps tightened to participant tolerance. Participants warmed up for several minutes before data were collected by running on the treadmill (Star Trac; Unisen, Inc., Irivine, CA, USA) with their habitual footfall pattern. The RF group was instructed to land with a heel-strike and the FF group was instructed to land with a toe-strike to reduce any affect of treadmill running on their footfall kinematics. The sagittal plane kinematics of all participants on the treadmill were not statistically different than their footfall pattern performed during the over-ground screening. After the warm-up, participants ran for 2 min on the treadmill at 3.5 m/s with their habitual footfall pattern before accelerometer and motion capture data were recorded. Data were collected for the last 15 s of the 2-min running period.

2.4. Data analysis

The sagittal plane ankle joint angle during the stance phase was determined from the processed kinematic data according to previously reported methods.[43] Time domain and frequency parameters from the tibia and head accelerometers were calculated using a custom MATLAB program (Mathworks, Inc., Natick, MA, USA). Time domain parameters from the tibia and head accelerometers were determined from 15 stance phases performed by each participant. A least-squares best fit line was subtracted from the raw data of each signal to remove any linear trend.[17] Data were then filtered with a second order Butterworth low-pass filter with a cut-off frequency of 60 Hz.[16] The first (HP1) and second (HP2) peak of the head acceleration signal occurred between 1% and 30% of stance and 31%−101% of stance, respectively. Peak positive tibial

Table 1
Participant characteristics of the rearfoot group (RF) and the forefoot group (FF) (mean ± SD).

	Male/female (*n*)	Age (year)	Height (m)	Mass (kg)	Pref. speed (m/s)	Distance/week (km)
RF	12/7	26.7 ± 6.1	1.75 ± 0.09	70.10 ± 10.00	3.47 ± 0.90	42.85 ± 29.04
FF	14/5	25.4 ± 6.2	1.76 ± 0.10	68.78 ± 9.51	3.73 ± 0.24	53.18 ± 25.53
p	−	0.515	0.788	0.680	0.220	0.252

acceleration (PPA) was identified as the peak occurring between 1% and 20% of stance.

Unfiltered head and tibial acceleration data from each stance phase were detrended then padded with zeros to equal 2048 data points, ensuring periodicity.[17] The power of the stance phase head and tibia acceleration in the frequency domain was determined by calculating the power spectral density (PSD) using a square window. Examination of the acceleration signals collected over the entire stance phase follows the periodic assumptions of Fourier analysis and allows for examination of frequencies below 15 Hz.[45,46] The PSD was performed on frequencies 0 to the Nyquist frequency (F_N) and normalized to 1 Hz bins.[14,22] After binning, the PSD was normalized in order for the sum of the powers from 0 to F_N to be equal to the mean squared amplitude of the data in the time domain. Examining the PSD results revealed two primary peaks or local maxima for the tibial and head acceleration signal power in both RF and FF running that were outside of the lower (4–8 Hz) and higher (10–20 Hz) ranges previously identified in the literature for RF running.[13–15,17] As a result, we expanded the lower and higher frequency ranges investigated to 3–8 Hz and 9–20 Hz, respectively, to more appropriately include the dominating frequency components of each footfall pattern. The frequency at which peak power occurred within the lower and higher frequency range of the tibial (TPF_{low}, TPF_{high}) and head (HPF_{low}, HPF_{high}) acceleration signal was determined. Signal power magnitude in the frequency domain was quantified by the integral of the signal power contained in the lower and higher frequency ranges in the tibial (TSM_{low}, TSM_{high}) and head (HSM_{low}, HSM_{high}) acceleration signals.[15]

A transfer function has been previously used to determine the degree of shock attenuation in human running by calculating the ratio of each frequency bin between the tibial and head signal[14,15,19,22] (i.e., the transmissibility of each frequency component[21]). The transfer function was calculated across all frequencies from 0 to F_N to determine the degree of shock attenuation occurring between the tibia to the head by:

$$\text{Shock attenuation} = 10 \times \log_{10}(PSD_{head}/PSD_{tibia})$$

For each frequency, the transfer function calculated the gain or attenuation, in decibels, between the tibia and head signals. Positive values indicated a gain, or increase in signal strength, and negative values indicated attenuation, or decrease in signal strength. A gain in lower frequency components is typically a result of changes in head vertical velocity and voluntary segment motion during the stance phase whereas negative values indicate attenuation in signal power as the impact shock travels through the body.[17,22] Shock attenuation magnitude was quantified by the integral of the transfer function result within the lower (ATT_{low}) and higher frequency ranges (ATT_{high}).

2.5. Statistical analysis

For the lower and higher frequency ranges, tibial and head peak signal power and signal magnitude were averaged across all stance phases of each participant and then across group. Group means for shock attenuation magnitude were also determined in this manner. A one-way analysis of variance was used to determine the difference between time and frequency domain acceleration variables between the RF and FF groups running with their habitual footfall pattern ($\alpha = 0.05$) using SPSS Statistics version 21.0 (IBM, Amonk, NY, USA). Effect sizes (d) were also calculated to determine if the differences between groups were biologically meaningful (small $d \leq 0.3$, moderate $d \leq 0.5$, large $d \leq 0.8$).[47]

3. Results

3.1. Head and tibial acceleration in the time domain

Ankle joint angles measured during the treadmill running confirmed that the RF group ran with a dorsiflexion angle at touchdown whereas the FF group ran with a plantar flexion angle at touchdown (Fig. 1).

Tibial and head acceleration in the time domain were plotted in Fig. 2. There was no significant difference in HP1 or HP2 between footfall patterns ($p > 0.05$) (Table 2). However, RF running resulted in a greater PPA compared with FF running ($p = 0.009$).

3.2. Head and tibial acceleration in the frequency domain

Tibial and head acceleration signals in the frequency domain were plotted in Fig. 3A and B, respectively. HPF_{low} was statistically greater during FF compared with RF running ($p = 0.001$). TPF_{high} was statistically greater during RF compared with FF running ($p < 0.001$). No statistical difference was observed for HPF_{high} or TPF_{low} ($p > 0.05$) (Table 2).

No statistical difference was detected between footfall patterns for HSM_{low} or HSM_{high} ($p > 0.05$) (Table 2). Both

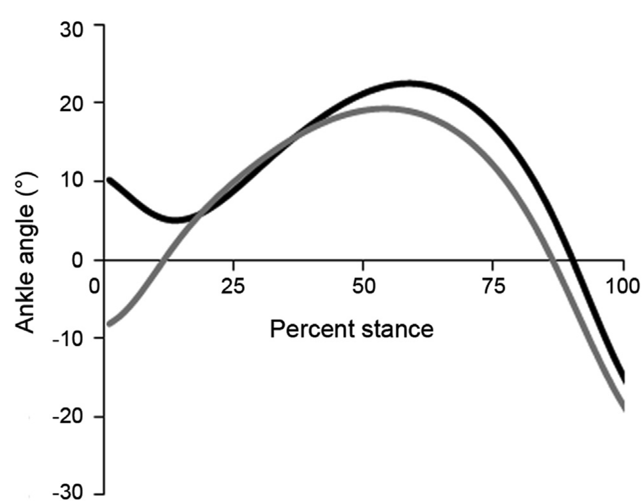

Fig. 1. Mean time series of sagittal plane ankle joint motion compared between the rearfoot (black) and forefoot (grey) groups performing their habitual footfall pattern during the treadmill running.

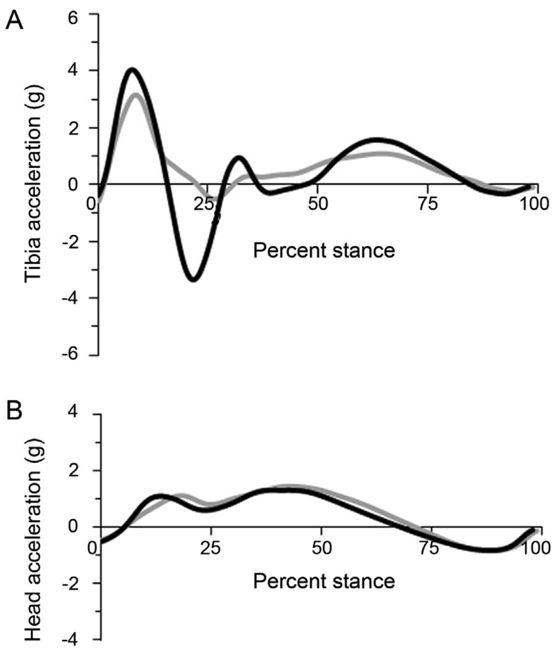

Fig. 2. Mean tibial (A) and head (B) acceleration in the time domain compared between the rearfoot (black) and forefoot (grey) groups performing their habitual footfall pattern.

Table 2
Time domain and frequency domain characteristics of rearfoot (RF) and forefoot (FF) groups performing their habitual footfall pattern. Variables included: peak positive tibial acceleration (PPA); first (HP1) and second (HP2) head acceleration peaks; frequency of peak power of the tibial ($TPF_{low,high}$) and head ($HPF_{low,high}$) acceleration signal within the lower and higher frequency ranges; signal power magnitude of the tibial ($TSM_{low,high}$) and head ($HSM_{low,high}$) acceleration signal within the lower and higher frequency ranges; and shock attenuation magnitude in the lower (ATT_{low}) and higher (ATT_{high}) frequency ranges. Positive values for $ATT_{low,high}$ indicate a gain in signal power whereas negative values indicate attenuation of signal power (mean ± SD).

	RF running	FF running	*p*	*d*
PPA (g)	5.07 ± 1.49	3.87 ± 1.36	0.009	0.8
HP1 (g)	0.51 ± 0.28	0.47 ± 0.19	0.923	0.2
HP2 (g)	1.01 ± 0.24	1.06 ± 0.26	0.221	0.2
TPF_{low} (Hz)	6.4 ± 0.5	7.2 ± 1.5	0.055	0.7
TPF_{high} (Hz)	14.3 ± 2.2	10.7 ± 2.8	<0.001	1.4
HPF_{low} (Hz)	3.7 ± 0.5	4.3 ± 0.5	0.001	1.2
HPF_{high} (Hz)	12.7 ± 1.7	11.8 ± 2.8	0.211	0.4
TSM_{low} (g^2/Hz)	0.355 ± 0.092	0.158 ± 0.101	<0.001	2.0
TSM_{high} (g^2/Hz)	0.860 ± 0.341	0.248 ± 0.253	<0.001	2.1
HSM_{low} (g^2/Hz)	0.194 ± 0.073	0.235 ± 0.085	0.121	0.5
HSM_{high} (g^2/Hz)	0.048 ± 0.024	0.041 ± 0.018	0.305	0.3
ATT_{low} (dB)	−17.9 ± 16.2	18.0 ± 21.3	<0.001	1.9
ATT_{high} (dB)	−165.1 ± 43.3	−88.7 ± 40.1	<0.001	1.8

TSM_{low} and TSM_{high} were statistically greater during RF running than FF running ($p < 0.001$) (Table 2).

3.3. Impact shock attenuation

The lowest frequency that was attenuated was 5.1 ± 0.5 Hz (mean ± SD) in RF running and 6.9 ± 0.9 Hz in FF running

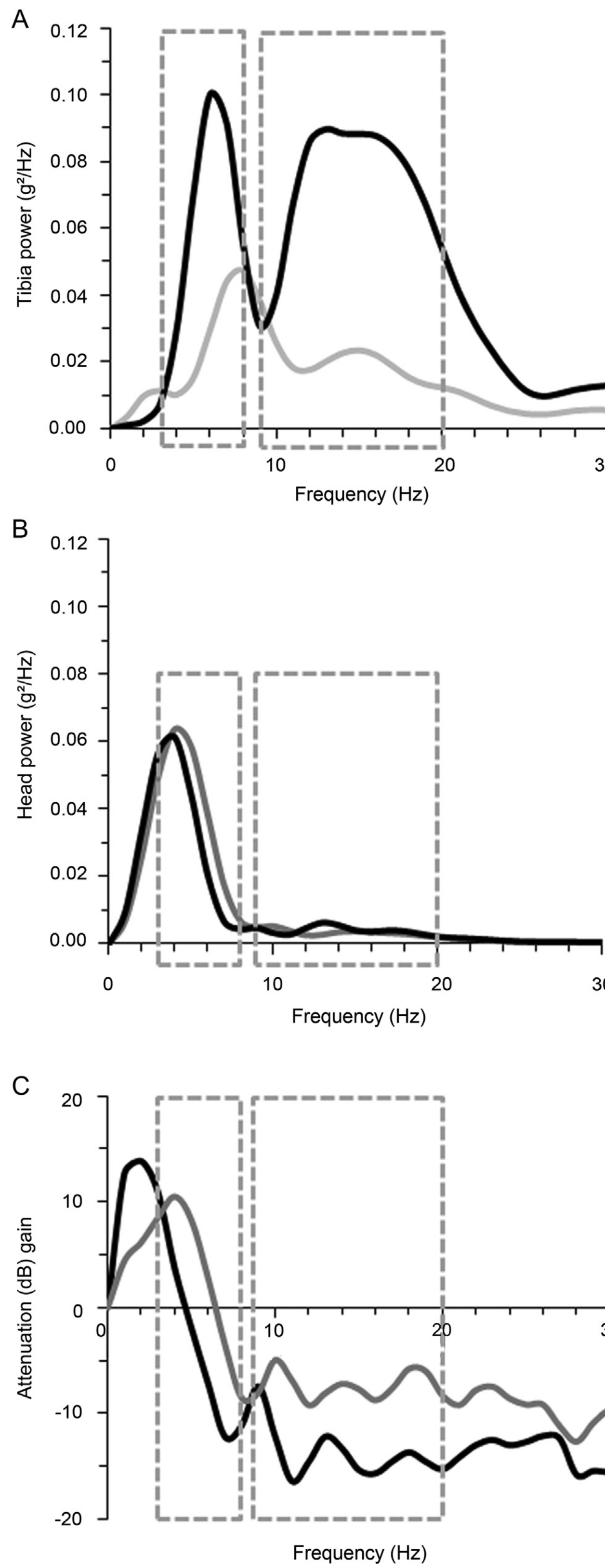

Fig. 3. Mean power spectra of the tibia (A) and head (B) acceleration signals and (C) transfer function compared between the rearfoot (black) and forefoot (grey) groups performing their habitual footfall pattern. Boxes indicate the lower (3−8 Hz) and higher (9−20 Hz) frequency ranges investigated.

($p < 0.001$, $d = 2.5$) (Fig. 3C). RF running resulted in attenuation of frequencies contained in ATT_{low} whereas FF running resulted in a gain of these frequencies ($p < 0.001$) (Table 2). ATT_{low} was positive in FF running because the gain of frequencies between 3 and 5 Hz was larger than the attenuation of frequencies between 6 and 8 Hz (Fig. 3C). RF running resulted in significantly greater ATT_{high} than FF running as indicated by a larger negative value for ATT_{high} ($p < 0.001$) (Table 2).

4. Discussion

The aim of this study was to determine if there were differences in the frequency content of impact shock and its subsequent attenuation between RF and FF running patterns. The first hypothesis, that RF running would result in greater peak tibial acceleration and signal power in the higher frequency range (9–20 Hz) than FF running, was supported whereas tibial acceleration power in the lower frequency range (3–8 Hz) would be greater in FF than in RF running, was not supported. The higher frequency range is representative of the vertical impact peak and the rapid deceleration of the foot and leg following initial ground contact.[13,17] RF running resulted in greater tibial acceleration power in the higher range because of the greater peak positive acceleration observed in the time domain with this pattern compared with FF running. Previous studies have indicated that joint kinematics at impact, such as greater heel vertical velocity, greater lower leg angle, decreased knee flexion angle, and a greater stride length result in greater tibial acceleration at impact.[14,15,18,19,26,48,49] Some of these kinematic features have been observed previously with RF compared with FF running.[23] Therefore, the joint geometry characteristics of RF running at foot-ground collision result in greater peak tibial acceleration magnitude in the time domain and greater tibial acceleration signal power in the frequency domain compared with FF running.

Joint and segment configurations at terminal swing will influence the velocity of the foot and leg just prior to impact and thus the force and rate of deceleration required to terminate foot and leg movement at initial contact. As a result, impact shock magnitude and the frequency that peak power occurs will also be affected. In the higher frequency range, peak power of the tibial acceleration signal occurred at a lower frequency during FF running compared with RF running. This finding may indicate that FF running results in a lower rate of tibial acceleration in the time domain at initial contact compared with RF running. Greater ankle compliance during FF running[23,50] may contribute to a lower rate of tibial deceleration after ground contact compared with RF running and may also delay the time that the vertical impact peak GRF occurs in FF running. A delay in the timing of the impact peak may explain why it is not visible in the time domain vertical GRF during FF running.[51]

Tibial acceleration signal power in the lower frequency range, representing the vertical GRF active peak and active movement of the leg and foot and whole body COM during stance,[13,17] was hypothesized to be greater during FF running than RF running because previous studies observed greater vertical GRF active peak magnitude in the time domain with FF running.[23,24] Contrary to this hypothesis, RF running resulted in greater tibial acceleration power magnitude in the lower range compared with FF running. Previous observations that RF running results in greater knee flexion excursion and velocity,[23] a greater stride length, and a greater contact time[52] may explain these results because these kinematics result in greater tibial signal power magnitude of frequencies below 10 Hz.[14,17,22] However, the frequency that peak tibial acceleration signal power occurred in the lower range was similar between footfall patterns indicating a similar rate of tibial acceleration during stance after initial contact. This result suggests that the dominant frequency component contributing to the overall tibial acceleration waveform in the lower frequency range is similar between footfall patterns. Similar rates of tibial acceleration may be a consequence of a greater tibial acceleration power magnitude occurring over a longer ground contact time in RF running compared with a lower rate of tibial acceleration magnitude occurring over a shorter ground contact time in FF running.

In support of the second hypothesis, the first and second peaks of the head acceleration signal in the time domain and the head acceleration power in the lower and higher frequency ranges were not different between footfall patterns. Additionally, there was no difference in the frequency that peak head acceleration occurred within the higher range. These results were expected because the body is able to respond to varying impact situations to maintain head stability[14,17,22,26] and suggest that the body is able to sufficiently attenuate the impact shock that occurred during both footfall patterns. However, peak power of the head acceleration signal within the lower frequency range was greater in FF running compared with RF running whereas there was no difference between patterns in the frequency of peak power in the higher range. This result indicates that the head and whole body COM oscillates at a greater dominate frequency in FF than in RF running, which reflects a greater rate of head acceleration in the time domain. The rate of acceleration of head may be greater with FF running because of the shorter contact time available to reverse the COM downward velocity after impact and may contribute to greater vertical GRF active peaks[10,23,24,53] than RF running. The rate of head and COM acceleration may be greater with FF running despite previous findings that this pattern minimizes vertical COM excursion.[23]

Our third hypothesis was partially supported as there was greater impact attenuation through the body for RF running compared with FF running in both the lower and higher frequency ranges rather than just the higher frequency range. The present study supports previous findings that RF running increased impact shock attenuation measured in the time domain[41] and peak tibial acceleration compared with FF running.[23,24] Greater attenuation of the higher frequency components resulting from the foot-ground collision was likely a result of the body responding to greater tibial acceleration in the time and frequency domains compared with FF running. However, in the lower frequency domain, FF running resulted in a gain of signal power whereas RF running resulted in attenuation of these frequencies. The difference in the lowest frequency that was attenuated in RF compared with FF running may explain these results and why RF running resulted in greater attenuation of frequencies in the lower range compared with FF running. During RF running, the lowest frequency that was attenuated was between 4 and 6 Hz across participants whereas the lowest frequency that was attenuated during FF running was between 5 and 9 Hz (Fig. 3C). A gain in signal power of the lower frequency components is typically a result of vertical oscillation of the COM and joint flexion occurring during stance that generate signal power of these lower frequency components.[13,17] This increase in signal power of low frequency components is reflected by the observation that FF running resulted in greater power of the head acceleration signal in this frequency range compared with RF running.

Impact shock attenuation occurs by a combination of active and passive mechanisms. Passive mechanisms are responsible for attenuating higher frequency components and include deformation of the shoe, heel fat pad, ligaments, bone, articular cartilage, and oscillation of soft tissue compartments.[28,29] Frequencies greater than 40 Hz are also attenuated by preactivation of muscle in preparation for ground contact.[32] Active shock attenuation mechanisms specifically responding to the impact stimulus and those that occur later in stance may be responsible for attenuating lower frequency components[29,33] and include eccentric muscle contractions, increased muscle activation, changes in segment geometry, and adjustments in joint stiffness.[14,34−37] When greater shock attenuation is required as a result of greater input energy, it is typically accomplished by active mechanisms such as increasing energy absorption by the muscles crossing the joints of the lower extremity.[14] Eccentric muscle contractions may be the primary mechanisms that attenuate forces transmitted through the body.[30] However, different segment and joint positions can affect the transmissibility of the impact shock and the primary mechanisms responsible for attenuation.[26,34] For example, increasing knee flexion may shift the degree of shock attenuation from passive tissue to muscular contractions by increasing the amount of knee extensor eccentric activity.[15] Muscle activity will affect joint stiffness which has also been shown to adjust in response to greater impact loading.[32] Results from previous studies investigating lower extremity joint compliance suggest that a compliant ankle is responsible for active shock attenuation during FF running more so than the knee whereas a compliant knee is responsible for active shock attenuation during RF running than the ankle.[23,50] Relying more on the knee than the ankle for shock attenuation may partially explain the greater shock attenuation observed with RF rather than FF running in the present study.

The differences in impact loading have been at the center of the footfall pattern debate. A recent retrospective study[1] and a recent survey study[2] found that those who use an MF or FF pattern have fewer injuries than those who use an RF pattern. These authors and others have suggested that MF and FF running may reduce the risk of developing running related injuries as a result of reduced impact loading compared with RF running.[1,24,48] These studies were excellent first steps toward furthering our knowledge of injury rates between footfall types. However, more research is needed given the limitations of survey studies and that statistical significance was only found in the retrospective study when male and female data were combined. Additionally, speculating that the findings were a result of differences in impact parameters between footfall patterns was a logical conclusion given that many studies have found a relationship between impact parameters and the development of running related injuries (e.g., Refs. 3−5). However, many other studies have not found this relationship (e.g., Refs. 6−8). Running injuries, regardless of footfall pattern, are the result of a complex interaction of many variables in addition to impact loading, such as excessive joint excursion and moments, greater vertical GRF active peak, and muscle weakness.[11,54] Results from the present study may assist with understanding why different types of injuries may be more common with one footfall pattern than another by providing insight on potential tissues and mechanisms responsible for attenuating shock with each footfall pattern. The capacity and reliance of different tissues and mechanisms to attenuate impact shock may be frequency dependent.[21]

The primary frequency content of acceleration due to impact shock and segment motion during stance of each footfall pattern may alter the reliance on the mechanisms used for shock attenuation and how specific tissues adapt or are injured with each footfall pattern. The present study indicates that RF running may result in a greater reliance on passive mechanisms because the power of higher frequency components of the tibial acceleration signal was greater compared with FF running. Bone deformation may be the primary passive shock attenuation mechanism during any activity.[30] Several studies have shown that impact forces similar to those experienced during RF running result in beneficial adaptations to bone, tendon, and muscle.[55−57] Damage to bone, articular cartilage, vertebral discs, and other passive tissues may result if they are overloaded,[30,40,58] and thus may be more at risk for injury from RF running. However, overload and injury also occur from MF and FF running[1,55,56,59,60] despite generating less impact energy than RF running. Given that FF running does not make heel contact, it cannot take advantage of passive mechanisms like the heel fat pad or shoe cushioning in the heel to attenuate impact forces resulting from the collision with the ground. Therefore, the proportion of shock that would otherwise be attenuated by these mechanisms must be applied to other tissues that may not have the same capacity for shock attenuation. As a result, FF running may have a greater reliance on kinematics and eccentric contractions of the plantar flexors to sufficiently attenuate impacts thus a greater risk of injury to the tissues involved. For example, the muscles of the triceps surae may not be as effective as the quadriceps at changing muscle activity to increase frequency damping due to the smaller mass of the triceps surae.[32] The potential differences in the reliance of impact attenuation mechanisms between footfall patterns may have implications on not only the risk of injury, but the tissues or joints that are susceptible to injury.

5. Conclusion

RF running resulted in greater impact loads and impact shock attenuation compared with FF running. Varying amounts of shock attenuation between footfall patterns suggest that the body has the capacity to manage a range of impulsive loads in order to protect the head from excessive acceleration. The difference in impact shock frequency content between footfall patterns suggests that the primary mechanisms for attenuation may also differ. Although RF running may elicit an increased requirement of the body tissues to attenuate greater impulsive loads which may be detrimental, it is also possible that the tissues adapt to greater impulsive loads in a beneficial manner. However, the threshold between injury and adaptation is currently unknown.

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