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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.

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