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# Low-frequency accelerations over-estimate impact-related shock during walking

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ABSTRACT

During gait, a failure to acknowledge the low-frequency component of a segmental acceleration signal will result in an overestimation of impact-related shock and may lead to inappropriately drawn conclusions. The present study was undertaken to investigate the significance of this low-frequency component in two distinctly different modalities of gait: barefoot (BF) and shod (SHOD) walking. Twenty-seven participants performed five walking trials at self-selected speed in each condition. Peak positive accelerations (PPA) at the shank and spine were first derived from the time-domain signal. The raw acceleration signals were then resolved in the frequency-domain and the active (low-frequency) and impact-related components of the power spectrum density (PSD) were quantified. PPA was significantly higher at the shank ($P < 0.0001$) and spine ($P = 0.0007$) in the BF condition. In contrast, no significant differences were apparent between conditions for shank ($P = 0.979$) or spine ($P = 0.178$) impact-related PSD when the low-frequency component was considered. This disparity between approaches was due to a significantly higher active PSD in both signals in the BF condition ($P < 0.0001$; $P = 0.008$, respectively), due to kinematic differences between conditions ($P < 0.05$). These results indicate that the amplitude of the low-frequency component of an acceleration signal during gait is dependent on knee and ankle joint coordination behaviour, and highlight that impact-related shock is more accurately quantified in the frequency-domain following subtraction of this component.



## 1. Introduction

The average person walks with approximately 6000 steps taken per day (Tudor-Locke et al., 2009) and with each step the body is exposed to an impact force in excess of bodyweight (Ounpuu, 1994). Within this impact force, transient forces exist which are determined by the rate of change in momentum of the contacting foot with respect to the ground causing impact-related accelerations (shock) to be transmitted up the musculoskeletal system. Inadequate attenuation of these accelerations, through alterations in the body's internal damping mechanisms has been suggested as a primary etiological agent underlying headaches and a number of pathological and injurious conditions (Whittle, 1999).

Footwear is a primary determinant of transient forces at initial contact (Whittle, 1999); understanding how these can be modulated by way of various mid-sole interfaces/technologies have led to considerable advancements in shoe development over recent decades for potentially enhancing shock attenuation. However, significantly lower peak impact force (derived from ground reaction force) has been reported in barefoot compared to footwear-mediated locomotion (Divert et al., 2005; Hamill et al., 2011; Keenan et al., 2011; Squadrone and Gallozzi, 2009). Yet paradoxically, there is considerable evidence to suggest that tibial accelerations (or shock) are significantly higher in barefoot locomotion (Clarke et al., 1983; Forner et al., 1995; Lafortune, 1991; McNair and Marshall, 1994; Sinclair et al., 2013). These studies may well have overestimated the magnitude of tibial shock through inclusion of low frequency accelerations due to movement.

The frequency range of impact-related shock from ground contact occurs between 10 and 35 Hz (Nigg and Wakeling, 2001; Voloshin et al., 1985; Wakeling and Nigg, 2001). Frequencies below this are synonymous with accelerations due to movement (Angeloni et al., 1994; Hamill et al., 1995; Shorten and Winslow, 1992), which should not be included in the description of impact-related shock. To do so may lead to inappropriately drawn conclusions and rehabilitation prescriptions with respect to various pathological and injurious conditions. As such, the importance of correctly measuring impact-related shock cannot be over-stated.

During gait, the use of accelerometers for measuring impact-related shock in response to ground contact is common practice, and this has been widely used for understanding the effects of footwear (Clarke et al., 1983; Forner et al., 1995; Lafortune, 1991;

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Lafortune et al., 1996; O'Leary et al., 2008; Ogon et al., 2001; Sinclair et al., 2013), orthotic intervention (Laughton et al., 2003) and prosthesis design (Adderson et al., 2007); as well as the induced segmental accelerations caused by musculoskeletal trauma (Milner et al., 2007), fatigue (Voloshin et al., 1998) and changes in spatio-temporal gait parameters (Derrick et al., 1998; Hamill et al., 1995; Mercer et al., 2002; Voloshin, 2000). A number of these studies however, were based on time-domain analysis and did not account for the presence of low-frequency accelerations induced by movement that become superimposed onto actual impact-related accelerations (Shorten and Winslow, 1992).

An alternative method for interpreting impact-related shock is spectral analysis of the time-domain signal (Derrick et al., 1998; Hamill et al., 1995; Mercer et al., 2002; O'Leary et al., 2008; Shorten and Winslow, 1992; Sinclair et al., 2013; Voloshin et al., 1985). When viewed in the frequency-domain, a typical segmental acceleration profile during running demonstrates two distinct peaks, representing: (1) low-frequency kinematically-mediated accelerations (active power spectrum density (PSD): 4–12 Hz); and (2) impact-related accelerations (impact PSD: 12–25 Hz) (Hamill et al., 1995; Mercer et al., 2002; O'Leary et al., 2008; Shorten and Winslow, 1992). The benefit of using this method is that the impact-related content can be easily discerned from the low-frequency accelerations due to movement. However, even with this approach there are examples in the literature of subjective delineation of impact-related frequencies (10–20 Hz: Mercer et al., 2002; 12–25 Hz: O'Leary et al., 2008). As such, these studies have failed to consider the intra- and inter-subject variability in gait that will inevitably alter the active PSD between strides and subjects. Correct identification of the active PSD component within a segmental acceleration signal should therefore be a primary consideration when interpreting impact-related shock.

To the authors' knowledge, this approach has yet to be explored in the analysis of walking and therefore warrants investigation. In light of the kinematic adaptations induced by barefoot locomotion (Squadrone and Gallozzi, 2009), it is likely that this will translate into a higher active PSD component underlying a time-domain shank acceleration signal (Shorten and Winslow, 1992). Therefore, the present study was undertaken to investigate the significance of this component during barefoot and shod walking. We hypothesised that the active PSD component within a shank acceleration signal will be significantly greater in barefoot than shod walking and this will be correlated with kinematic parameters that differentiate gait pattern between conditions. This, rather than differences in impact-related PSD, may explain the higher acceleration signal in the barefoot condition when interpreted in the time-domain. Furthermore, previous work has shown that footwear reduces shock transmission to the spine (Ogon et al., 2001). However, in this study, low-frequency accelerations were not acknowledged in the interpretation of the time-domain signals. Hence, we evaluated shock attenuation between the shank and spine in barefoot and shod walking in the frequency domain.

## 2. Methods

### 2.1. Participants

Twenty-seven participants ($n = 27$; mean ± SD, 12 Male: 27.8 ± 7.5 yrs, 1.74 ± 0.06 m, 71.2 ± 9.8 kg; 15 female: 26.1 ± 6.2 yrs, 1.66 ± 0.05 m, 59.2 ± 6.7 kg) gave their written informed consent to participate in the study, which had received prior University Research Ethics Committee approval. All participants reported from initial screening that they were free from any current musculoskeletal injury or pathology that might otherwise have biased the resulting outcome measures.

### 2.2. Experimental protocol

Prior to testing, each participant's preferred walking speed was ascertained from five preliminary barefoot (BF) and shod (SHOD) walking trials, which were calculated by speed gates (Newtest, Finland) separated 6 m apart along a walkway. This approach was adopted so that a true adaption to ground impact was established since a move away from preferred walking speed negatively influences shock attenuation (Derrick et al., 1998; Heiderscheit et al., 2011). Hence, the acceptable range for individual walking speed within each main trial was determined by one standard deviation either side of their averaged preferred speed.

The experimental protocol required participants to perform five main walking trials in BF and SHOD (*Kalenji Success*, 0.39 EVA, Shore 55C) conditions. Sufficient time was given for familiarisation and respective trials were counterbalanced to exclude order effect on the outcome measures. All trials commenced with right-sided gait initiation and all data were taken from the right lower extremity of participants.

### 2.3. Data collection

#### 2.3.1. Accelerometry

Two tri-axial accelerometers (ACL300; range: ±10*g*, weight: 10 g, resolution: 0.0025*g*; Biometrics Ltd., UK) were located on the shank and spine segment to compare the transmissibility of impact-related shock between conditions. One was positioned at the distal antero-medial aspect of the tibia, proximal to the medial malleolus (Hamill et al., 1995; Mercer et al., 2002), and the second – midway between the superior aspect of both iliac crests, representing the third lumbar vertebrae (L3). Similar to Ogon et al. (2001), the spinal accelerometer was positioned at L3 for enhanced reliability of identification with respect to the intercrystal line formed by palpation of iliac crests (Chakraverty et al., 2007). The third lumbar vertebrae is regarded as the optimal site for the measurement of spinal accelerations since the effects of contamination from rotational trunk motion are minimised with respect to linear acceleration output (Kavanagh and Menz, 2008).

Prior to attachment, the accelerometers were calibrated within a custom-made frame with the *y*-axis referenced to a global vertical orientation. The skin areas corresponding to the aforementioned attachment sites were shaved where necessary. The accelerometers were first securely fixed to the skin and then preloaded with zinc oxide medical tape in order to minimise the effect of soft-tissue vibrations on the acceleration signal (Shorten and Winslow, 1992). The validity of the ACL300 accelerometer was confirmed by way of an electromagnetic exciter driven by a crystal oscillator, which elicits a standard level of acceleration of $10\ \mathrm{m\ s^{-2}} \pm 3\%$ (Type 4294; Brüel&Kjær, Denmark).

#### 2.3.2. Kinematics

Two electro-goniometers (SG150, SG110; accuracy ± 2°; Biometrics Ltd., UK) were calibrated using a manual goniometer and positioned to measure sagittal plane motion about the knee and ankle joints. They were first securely fixed to the skin and reaffirmed with zinc oxide medical tape. The validity of the SG150 sensor was confirmed by comparing differentiated knee joint angular displacement data ($n = 1$) to those recorded by isokinetic dynamometry (Kin Kom, Chattanooga Group Inc., USA) during $30°\ \mathrm{s^{-1}}$ movement.

A foot-switch (Biometrics Ltd., UK) attached to the posterior aspect of the right heel determined the time of each initial contact. The channel sensitivity and excitation output of the switch were set at 300 mV and 3000 mV respectively, in accordance with the manufacturer's guidelines.