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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.

1.0 Introduction

The average person walks with approximately 6,000 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 over-estimated 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; 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): 4-12Hz); and 2) impact-related accelerations (impact PSD: 12-25Hz) (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 (10-20Hz: Mercer et al, 2002; 12-25Hz: 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.

2.0 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 6m 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 grams, 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 intercristal 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 pre-loaded 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 m.s^{-2} ± 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 30º.s^{-1} movement.

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

Both accelerometry and kinematic data were recorded continuously and pre-amplified via a conditioning unit (DLK900; Biometrics Ltd, UK) mounted on a belt around the waist of each subject. The data were sampled at a frequency of 500Hz via an analog-to-digital converter

(CED 1401 power, Cambridge, UK) using Spike2 data acquisition software (v6.10, CED, Cambridge, UK) with a resolution of 16 bits.

2.4 Data Analysis

The characteristic parameters of the recorded signals (Figure 1) were calculated from the third ipsilateral stride and averaged across the five trials performed in each condition using custom-written scripts developed in Spike2 v.6.10 analysis software (CED, Cambridge, UK).

2.4.1 Accelerometry.

Only axial accelerations were considered for analysis. Initially, the raw shank and spine time-domain signals were filtered with a 60Hz 4th order Butterworth low-pass filter for the quantification of peak positive acceleration (PPA) during stance phase (Sinclair et al, 2013). Following this, the signals were then filtered using a 4-50Hz finite impulse response band-pass function with a transition gap of 2.6Hz. Using this approach, the fundamental frequency of gait is omitted (~1Hz, Antonsson and Mann, 1985), whilst the frequency content containing the spectral power from segmental displacements (Angeloni et al, 1994) and in excess of 99% of the impact-related power (Lafortune et al, 1995) is preserved. Both acceleration signals were analysed from the time of ground contact to peak knee flexion; representing the absorption period of the stance phase. All data points outside this range were padded with zeros (Hamill et al, 1995; Shorten and Winslow, 1992) so that the time-domain range under analysis equalled 1.024s.

The PSD of shank and spine accelerations were derived using the Fast Fourier Transform (FFT) function (Figure 1). To overcome a limitation of the FFT in assuming a cyclical waveform, a Hanning window function was used to taper the start and end of each data block within a waveform to zero and prevent sharp discontinuities that may have caused additional frequency components within the result. Accordingly, a FFT block size of 512 (1.024s) meant a bin resolution of 0.98Hz given a sampling frequency of 500Hz. The resulting PSD was then normalised where the sum of powers from 0-50Hz was proportional to the RMS amplitude of the data in the time domain. Units of PSD were thus ***g***.Hz^{-1}. A transfer function describing the gain and attenuation (dB) between the shank and spine accelerations (Figure 1) was calculated as the logarithmic function of the PSD at each frequency bin (Hamill et al, 1995):

$$\text{Tranfer Function} = 10\log_{10}\left(\frac{\text{PSD}_{\text{spine}}}{\text{PSD}_{\text{shank}}}\right)$$

Active PSD was defined as the sum of powers up until the frequency bin containing the lowest power that delineated between low- and impact-related frequencies of the power spectrum (Figure 1). Generally, this cut-off point was within the range reported in the literature (8-12Hz, Hamill et al, 1995; Mercer et al, 2002; O'Leary et al, 2008); however, this varied on an intra- and inter-subject basis, and on occasions active PSD exceeded 12Hz in certain participants. The impact-related PSD was calculated by subtracting the active PSD from the total PSD of each acceleration signal. The variables used to quantify the shank and spine PSDs were: active and impact-related PSD (***g***.Hz^{-1}) and the frequency of their respective peaks (Hz). Additionally, peak attenuation (dB), its corresponding peak frequency (Hz), and the percentage of impact-related PSD attenuation between the shank and spine segments were measured (Figure 1).

2.4.2 Kinematics.

Sagittal plane knee and ankle angular displacement profiles were digitally filtered with a 10Hz finite impulse response low-pass filter using a transition gap of 1.3Hz, digitally differentiated to calculate movement velocity, then time-normalized to 110% of gait cycle commencing at 10% before heel-strike (Figure 1). The following discrete kinematic variables were extracted for analysis: stride time (s), joint angle at heel-strike (°), knee flexion/ankle plantarflexion range of motion (°), and time to peak displacement (% Gait Cycle (GC)).

Furthermore, the average knee and ankle joint angular accelerations (rad.s^{-2}) were calculated from heel-strike to the initial peak flexion and plantarflexion velocities, respectively (Figure 1).

2.5 Statistical analyses

Outcome measures were tested statistically for normality of distribution with a Kolmogorov-Smirnov 1-sample test (PASW v.18.0, IBM Corp., USA) and compared using paired-samples t-tests to identify a condition effect (BF vs. SHOD) for all shank and spine PSD variables and the kinematic data. An alpha level for statistical significance was set at 0.05. A least-squares linear regression analysis of the shank active PSD and the absolute difference between the average knee and ankle joint angular accelerations was performed to assess the relationship between joint coupling motion and low-frequency accelerations. Pearson correlation (r) was used to identify the strength of this relationship.

3.0 Results

Preferred average walking speed established prior to testing was 1.21 ± 0.15m.s^{-1} and 1.23 ± 0.17m.s^{-1} in BF and SHOD conditions, respectively (t=-1.59, *df*26, *P*=0.062).

3.1 Accelerometry.

PPA in the time-domain signal was significantly higher at the shank (2.87 ± 1.2 vs. 1.58 ± 0.81***g***; t=8.49, *P*<0.0001) and spine (0.59 ± 2.5 vs. 0.48 ± 0.24***g***; t=3.58, *P*=0.0007) in the BF condition. In contrast, when the data was resolved into the frequency-domain, there was no significant differences between BF and SHOD for impact-related PSD at the shank (*P*=0.979) or the spine (*P*=0.178) (Table 1). The frequency where peak impact-related PSD occurred was however significantly higher in BF (*shank*: t=3.79, *P*=0.001; *spine*: t=2.56, *P*=0.017).

Active PSD was significantly higher at the shank (t=6.04, *P*<0.0001) and spine (t=2.85, *P*=0.008) in the BF condition. The frequency where peak active PSD occurred was also significantly higher in BF (*shank*: t=7.88, *P*<0.0001; *spine*: t=4.51, *P*<0.0001). There was no difference in peak attenuation (*P*=0.368), but its corresponding frequency was significantly higher in BF (t=3.062, *P*=0.005). A similar percentage of overall impact-related PSD attenuation was found between conditions (*P*=0.310).

The shank active PSD was strongly correlated with the absolute difference between the average knee and ankle joint accelerations in both conditions (BF: *r*=0.93, *P*<0.0001; SHOD: *r*=0.78, *P*<0.0001; Figure 2).

3.2 Kinematics.

Stride time was significantly shorter in BF (t=-6.97, *P*<0.0001). BF also demonstrated significantly greater knee flexion (t=6.10, *P*<0.0001) and ankle plantarflexion (t=-3.75, *P*=0.001) angles at initial contact when compared to SHOD (Table 2). The times to peak knee flexion and ankle plantarflexion were significantly earlier in the BF stride cycle (t=-7.29, *P*<0.0001; t=-2.44, *P*=0.022, respectively) with range of motion (ROM) being significantly less (t=-2.32, *P*=0.029; t=9.63, *P*<0.0001, respectively). There was no difference between conditions in average knee joint angular acceleration (*P*=0.844), however, the largest between-conditions difference was found for average ankle joint angular acceleration (t=10.15, *P*<0.0001); being significantly reduced in BF. Consequently, the mean absolute difference between these average joint accelerations was significantly higher in BF (t=7.49, *P*<0.0001).

4.0 Discussion

The purpose of this investigation was to investigate whether low-frequency kinematically-mediated accelerations, due to active movement, may cause an overestimation of the magnitude of tibial shock experienced during walking. Time-domain shank and spine accelerations were decomposed into the frequency-domain and compared between barefoot and shod conditions; two modalities which differ significantly with respect to the neuromuscular control associated with ground contact (von Tscharner et al, 2003). Akin to observations made during running (Hamill et al, 1995; Shorten and Winslow, 1992), the present study noted two distinct peaks in the spectral distribution of these accelerations that represent active and impact-related components of the signal. Separate analysis of these components confirmed the experimental hypotheses of the present study. Firstly, impact-related PSD experienced at the shank during barefoot walking is not significantly different to that measured during shod walking. Additionally, no significant difference was observed in the overall impact-related PSD attenuation between the shank and spine segments. Secondly, the magnitude of low-frequency (active) PSD recorded at the shank was significantly greater in barefoot walking and furthermore, this parameter was found to be strongly correlated with the absolute difference in average knee and ankle joint angular accelerations in both conditions.

The finding that impact-related PSD at the shank was not significantly different between BF and SHOD conditions is in disagreement with earlier investigations, which used time-domain analysis and reported tibial shock to be significantly higher during experimental or simulated barefoot walking (Forner et al, 1995; Lafortune, 1991; Lafortune et al, 1996). Three rational explanations can be provided to explain this discrepancy, which should be considered in parallel. Firstly, an important consideration pertains to how the acceleration signal is processed before being subsequently expressed in magnitude of $\boldsymbol{g}$. There are numerous examples within the literature where the low-frequency (active) component of the time-domain acceleration signal has not been acknowledged. Indeed, the present findings demonstrate that when raw data is filtered with a generic 60Hz low pass filter and peak positive acceleration is derived from the time-domain signal, ***g*** is shown to be significantly greater during barefoot walking. However, this acceleration signal contains an active component, which is kinematically-mediated (Hamill et al, 1995; Shorten and Winslow, 1992) and should therefore not be included in the assessment of impact-related shock. Furthermore, the present findings also demonstrate that the cut-off for active PSD can in fact be as low at 6Hz in some participants and exceeds 12Hz in others. Hence, the magnitudes of tibial shock (***g)*** reported in the literature may well be over-estimated, and which may also have facilitated incorrectly drawn conclusions. Recently, the peak tibial shock was reported to be significantly greater during barefoot running when compared to conventional and barefoot-inspired footwear (Sinclair et al, 2013). Combined with a significant increase in the median frequency of the shank acceleration signal, the authors concluded that barefoot runners are more susceptible to musculoskeletal injury. In light of the present findings, a reassessment of the data provided by Sinclair et al, (2013) is warranted before injurious claims can be made.

Secondly, the present study observed a significant decrease in stride time in BF. This indirectly confers with a reduced stride length; a notable feature of barefoot walking (Keenan et al, 2011). Hamill et al, (2011) drew on the work of others and suggested that impact-moderating behaviour is found through such an adaptation. A reduced stride length imposes the lowest tibial accelerations when compared to increases in length (Derrick et al, 1998). Heiderscheit et al, (2011) noted that knee angle at initial contact increases with an increase in step rate (decreased stride length) and is accompanied with a flatter foot placement and lower probability of impact-transient occurrence. These are consistent with the kinematic data reported in the present study. It would seem that a main characteristic of barefoot walking is a pre-programmed adaptation prior to ground contact in order to reduce the effective mass at impact, which indicates the proportion of body mass that responds to impact force. Estimates of effective mass during barefoot walking approximate 6% of body mass and decreases further during activities which require increased knee flexion (Chi and Schmitt, 2005). By altering segmental geometry prior to ground contact concomitant changes occur in joint stiffness, segmental deformations and segment moments of inertia (Derrick, 2004); all of which contribute to reducing the magnitude of impact force.

Finally, if it is accepted that impact shock is influenced by stride length, which in turn is a surrogate of speed, then fixing speed for a homogenous data set might not be an accurate representation of shock attenuation. This approach does not consider inter-subject variation and consequently imposes alterations in spatio-temporal gait parameters in participants. Forner et al, (1995) fixed their walking velocity at 2.0m.s^{-1} and noted in excess of a 150% increase in tibial acceleration in barefoot walking when compared to differing mid-sole constructions. Similarly, Lafortune, (1991) asked one subject to walk at 1.5m.s^{-1} and noted a two-fold increase in tibial acceleration in a barefoot condition compared to when a hard leather–soled shoe was worn. In the present study, preferred walking speed in both conditions was adopted and the group means (BF: 1.21 ± 0.15m.s^{-1}; SHOD: 1.23 ± 0.17m.s^{-1}) highlight a reduction in walking speed compared to those implemented in the afore-mentioned studies. Moreover, the reported speeds demonstrate that one tends to walk slower in a barefoot condition; therefore the present findings are derived from a more ecologically valid representation of impact loading during barefoot walking.

A strong linear relationship was demonstrated between the shank active PSD and the absolute difference in knee and ankle joint average angular acceleration in both conditions. The rationale for correlating these two variables was based on the known knee joint-mediated de-coupling of the shank segment relative to the foot as a mechanism for reducing the effective mass of the body at ground contact (Derrick, 2004). The motions of these segments are known to be less coordinated (out-of-phase coupling behaviour) during barefoot locomotion (Kurz and Stergiou, 2004). Furthermore, active PSD is known to be kinematically-mediated (Shorten & Winslow, 1992); therefore, its magnitude should intuitively be dependent on the manner in which the shank rotates about the ankle joint during load response of stance phase. Hence, we investigated whether the shank active PSD is correlated with the coupling of knee and ankle joint kinematics by subtracting the change in ankle joint angular velocity between initial contact and peak knee flexion from the respective change in knee joint angular velocity. The significant correlation reported here suggests a good place for future work to search for either alternative correlates or to define an underpinning mechanism responsible for this component of a time-domain acceleration signal.

Derrick, (2004) theorised that reductions in effective mass impose higher tibial accelerations. The present findings indicate that whilst this is the case, they are not necessarily impact-related; rather, the higher shank accelerations observed during barefoot walking in the time-domain are kinematically-mediated and significantly correlated with sagittal plane knee and ankle joint angular kinematics. The greater the absolute difference between these joint's average angular acceleration, the greater the low-frequency PSD of the shank acceleration. This relationship was more consistent in the BF condition with the linear regression model accounting for 86% (r=0.93) of the variance between participants, whereas this was reduced to 61% (r=0.78) during shod walking. There was no statistical significance observed for average knee flexion angular acceleration between conditions, therefore the source of this variance in shod walking must have derived from ankle joint angular acceleration. Indeed, this measure contained the largest condition-dependent effect size of all dependent outcome measures, highlighting a dependency on an accurate and controlled foot placement during barefoot walking. Increased ankle stiffness has been postulated as a mechanism responsible for the differences in impact characteristics observed between barefoot and shod runners (Hamill et al, 2011). The present findings suggest this is also the case in barefoot walking, where significant reductions in ankle joint range of motion and average plantarflexion acceleration were demonstrated by participants. A stiffer ankle combined with a compliant knee should result in greater kinematically-mediated low-frequency accelerations at the shank due to the uncoupling of knee angular displacement relative to that at the ankle (Derrick, 2004). Alternatively, increased ankle plantarflexion ROM coupled with knee flexion, as observed in the SHOD condition, equates to an anti-phase movement (ankle extension-knee flexion). Therefore, low-frequency accelerations should be reduced as there is limited opposing distal motion that prevents the knee from un-coupling from the ankle joint to induce this active PSD.

The frequency where peak impact-related PSD occurred at was significantly greater during barefoot walking for both shank and spine accelerations. This was also the case for the frequency corresponding to peak attenuation, but there was no statistical difference between conditions in the overall attenuation of the impact-related PSD. The reported values for the peak frequency of impact-related PSD fall within the range reported in the literature (10-35Hz; (Nigg and Wakeling, 2001; Voloshin et al, 1985; Wakeling and Nigg, 2001). It is well accepted that the musculoskeletal system tunes itself to effectively dampen all frequencies in and around the impact-related bandwidth (Nigg and Wakeling, 2001; Wakeling and Nigg, 2001). The frequencies of damping coefficients recorded from muscles exposed to vibration have been shown to exceed the vibration frequency of the input signal (Wakeling and Nigg, 2001). In the present study, the peak attenuation frequency exceeded the respective peak impact-related PSD frequencies. As such, it is a matter of debate as to whether an increase in peak impact-related PSD frequency induced through barefoot locomotion is potentially harmful as has recently been suggested (Sinclair et al, 2013). It is logical that peak impact frequency be higher in a condition where a foot-ground-interface is absent. However, in light of the present evidence, this may alternatively represent a natural tuning to impact force resulting in an adequate damping of the energy from the shock wave. Indeed, no statistical difference was found in the impact-related PSD at the spine. Interestingly, the present results show that the average peak frequency of the spine impact-related PSD decreased with respect to the same measure at the shank during barefoot walking. In contrast, the opposite was true in the shod condition (Table 1) indicating that higher peak impact-related frequencies are experienced more so at the spine than at the shank.

Finally, whilst accelerometry is an accepted method for evaluating movement patterns during walking (Kavanagh and Menz, 2008); an inherent limitation associated with its use is the potential contamination of the time-domain signal with artefacts due to skin movement. These effects are minimised with the use of low-mass accelerometers and pre-loading of the instrument, which compresses the soft-tissue, in turn increasing its stiffness and the resonant frequency of the tissue-accelerometer system (Forner-Cordero et al, 2008). However, even with this accepted approach, the resonant frequency of the tissue-accelerometer system is still less than 100Hz (Shorten and Winslow, 1992). Hence, during post-processing the time-domain acceleration signals were band-passed between 6-50Hz since it has been shown that 99% of the frequency content of tibial acceleration is below 50Hz. By adopting this approach, we potentially negated the influence of soft-tissue artefact within the time-domain signals; nonetheless, caution should be made regarding the possible contamination of the signals, particularly at the spine (Kitazaki and Griffin, 1995).

The frequency resolution used to calculate the power spectrum of the accelerations may present another limitation of the present study. Segmental accelerations were sampled at 500Hz, resulting in a bin resolution of 0.98Hz. It is possible that greater bin resolution achieved through a higher sampling frequency might have given more accurate delineation between the active and impact-related segments of the power spectrum. Whether this would have translated into a stronger relationship between the joint kinematics and the active PSD component is uncertain and perhaps worthy of future investigation. However, given the size of the differences found between barefoot and shod conditions for many spectral parameters, it is unlikely that enhancing the resolution further would have affected the biomechanical relevance of the main findings of the study.

In conclusion, the findings of the present study demonstrate a few of the adaptations made during barefoot walking that aim to reduce the impact-related shock in the absence of a foot-ground interface to levels present in a shod condition. These include spatio-temporal alterations and changes in sagittal plane knee and ankle joint angular kinematic profiles, which de-couple presumably to reduce the effective mass of the system at ground contact. As a consequence, the kinematically-mediated low-frequency active component of a shank-mounted acceleration signal is significantly greater in barefoot compared to shod walking, without any difference noted in the amplitude of the impact-related PSD signal. This discrepancy with conclusions drawn from previous studies most likely reflects an acknowledgement of the active PSD component within a segmental time-domain signal and incorporating ecological validity within the present experimental design. In light of the inherent nature of intra- and inter-subject gait variability, it is concluded that impact-related shock is more accurately quantified in the frequency-domain on a subject-trial basis following subtraction of the low-frequency component of the acceleration signal.

Conflict of interest statement

There are no conflicts of interest.

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Table 1. Mean±SD (*n*=27) power spectrum density (PSD) parameters during barefoot (BF) and shod (SHOD) walking. † indicates P<0.05, ‡ P<0.01.

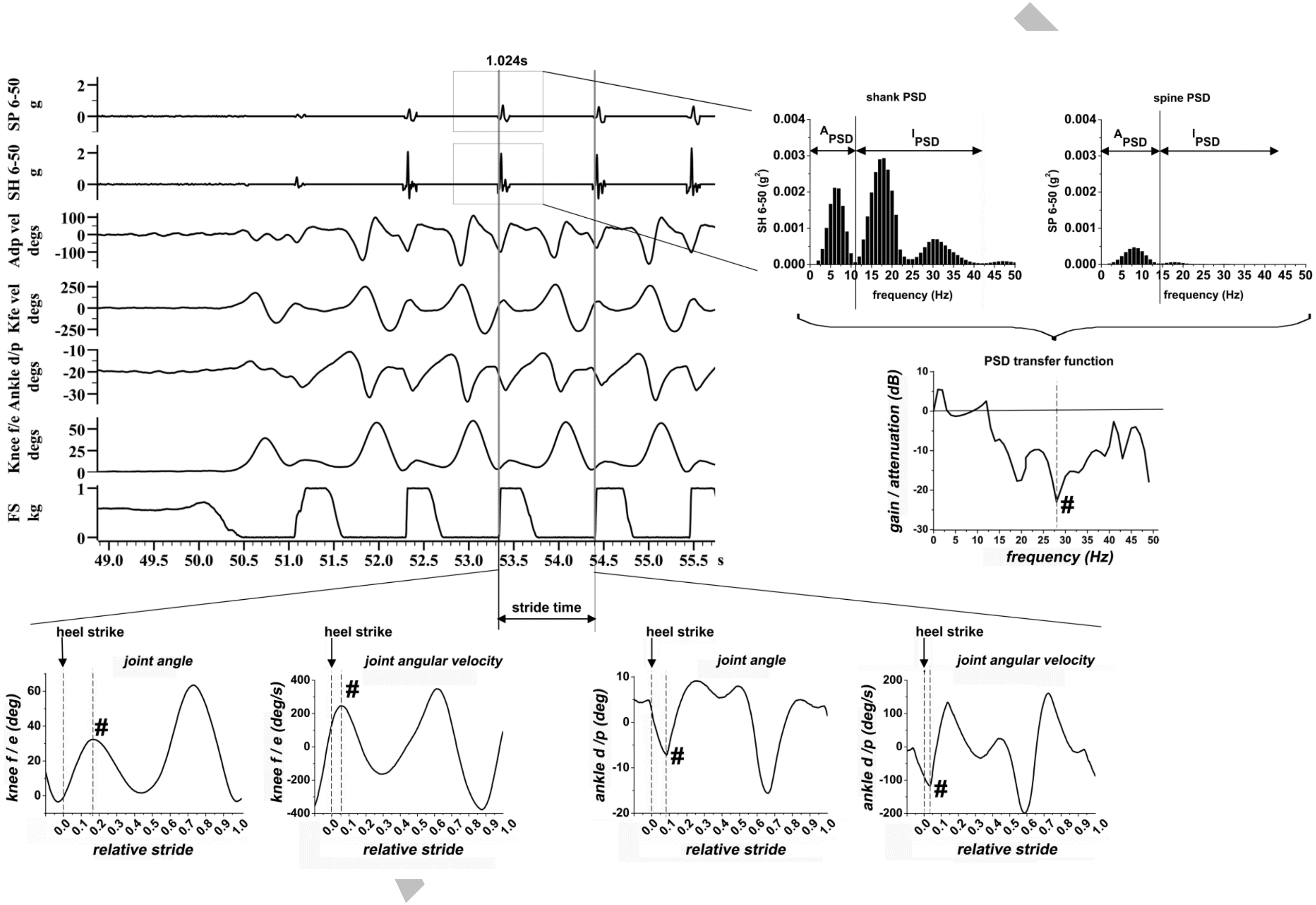
	BF		SHOD		
	Mean	SD	Mean	SD	
ACTIVE PSD (g.Hz^{-1})					
Shank	0.266	0.138	0.142	0.089	‡
Spine	0.153	0.073	0.114	0.056	‡
PEAK ACTIVE PSD FREQUENCY (Hz)					
Shank	8.6	1.5	6.6	1.2	‡
Spine	7.9	1.8	6.8	1.3	‡
IMPACT-RELATED PSD (g.Hz^{-1})					
Shank	0.309	0.181	0.308	0.165	
Spine	0.029	0.031	0.020	0.014	
PEAK IMPACT-RELATED PSD FREQUENCY (Hz)					
Shank	23.1	6.7	17.7	4.4	‡
Spine	21.9	5.7	19.0	5.3	†
PEAK ATTENUATION (dB)	-21.37	4.98	-20.58	3.69	
PEAK ATTENUATION FREQUENCY (Hz)	35.6	11.3	29.0	9.3	‡
IMPACT-RELATED PSD ATTENUATION (shank - spine %)	84.9	28.5	91.0	7.5	

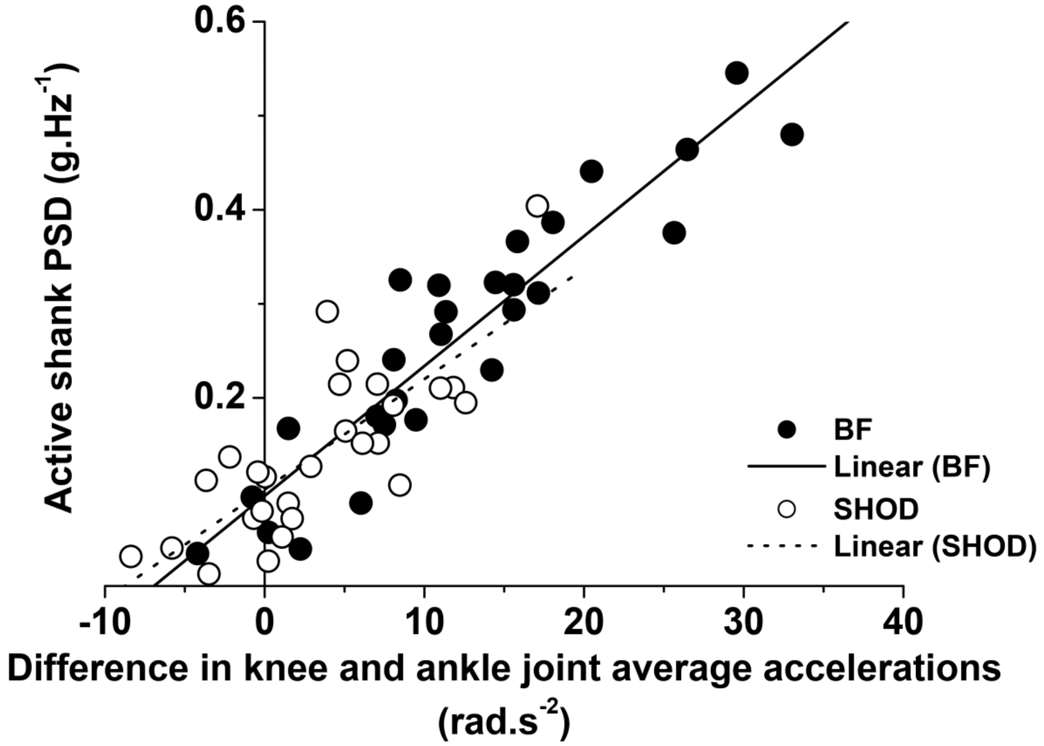
Table 2. Mean ± SD (*n*=27) sagittal plane kinematic variables during barefoot (BF) and shod (SHOD) walking. +'ve: knee flexion and ankle dorsiflexion, -'ve: ankle plantarflexion. † indicates $P<0.05$, ‡ $P<0.01$.

	BF		SHOD		
	Mean	**SD**	**Mean**	**SD**	
STRIDE TIME (s)	1.03	0.08	1.07	0.08	‡
GROUND CONTACT (°)					
KNEE	5.1	4.0	2.9	3.5	‡
ANKLE	-0.7	1.8	0.6	2.0	‡
ROM (°)					
KNEE	13.4	4.6	14.0	5.1	†
ANKLE	-6.8	2.5	-9.9	2.6	‡
TIME TO PEAK DISPLACEMENT (% stance)					
PEAK KNEE FLEXION	13.9	1.4	15.4	1.8	‡
PEAK ANKLE PLANTARFLEXION	8.0	1.3	8.4	1.4	†
AVERAGE ACCELERATION (rad.s^{-2})					
KNEE	21.0	9.0	20.7	7.2	
ANKLE	-8.6	4.6	-17.4	5.8	‡
difference (knee-ankle)	12.3	9.3	3.4	5.9	‡

Figure 1. Schematic diagram of study parameters. *Top Left*: Filtered kinematic and accelerometry data. Vertical lines represent the third ipsilateral gait cycle (stride time) to which the kinematic waveforms are normalised against. *Bottom*: Normalised knee and ankle joint angular displacements and velocity. Vertical line at '0.0' relative stride represents initial contact and '#' at the second vertical line represents peak knee flexion / velocity and ankle plantarflexion / velocity. Top Right: Shank and spine power spectrum densities (PSD) resolved from an FFT window ~ 1.024s incorporating data points that were zero-padded either side of the time-domain signal between initial contact and peak knee flexion. Each PSD comprises an active (A_{PSD}) and an impact-related (I_{PSD}) component. The gain or attenuation of power at each frequency between the shank and spine segments is described with a transfer function (*below*), with '#' indicating peak attenuation and frequency where this occurred.

Figure 2. Relationship between shank active PSD and the difference between the absolute average values of knee and ankle joint accelerations in barefoot (BF: r=0.93) and shod (SHOD: r=0.78) conditions.


1.024s
SP 6-50 g
SH 6-50 g
Adp vel degs
Kfe vel degs
Ankle d/p degs
Knee f/e degs
FS kg
s
shank PSD
spine PSD
A_{PSD}
I_{PSD}
SH 6-50 (g^2)
SP 6-50 (g^2)
frequency (Hz)
PSD transfer function
gain / attenuation (dB)
frequency (Hz)
#
stride time
heel strike
joint angle
joint angular velocity
knee f / e (deg)
knee f / e (deg/s)
ankle d /p (deg)
ankle d /p (deg/s)
relative stride



0.6
0.4
0.2
Active shank PSD (g.Hz-1)
-10
0
10
20
30
40
Difference in knee and ankle joint average accelerations (rad.s-2)
BF
Linear (BF)
SHOD
Linear (SHOD)

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