



KINEMATIC REGULATION OF TIME AND FREQUENCY DOMAIN COMPONENTS OF ACCELERATIONS MEASURED AT THE TIBIA DURING HEEL-TOE RUNNING

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ABSTRACT

Purpose. The transmission of tibial accelerations through the musculoskeletal system may contribute to the aetiology of injuries. Therefore, determining the mechanisms that regulate impact accelerations may have potential clinical significance. This study aimed to determine the influence of lower extremity kinematics on the regulation of both time and frequency domain characteristics of tibial accelerations during running. **Methods.** Forty participants ran at $4.0 \text{ m} \cdot \text{s}^{-1} \pm 5\%$. Three-dimensional joint kinematics from the hip, knee and ankle were measured using an eight-camera motion analysis system operating at 250 Hz. Regression analyses treating time and frequency domain tibial acceleration parameters as criterion variables were used to identify lower extremity parameters associated with the passive regulation of impact accelerations. **Results.** The overall regression model yielded an adj. $R^2 = 0.13$, $p \leq 0.01$. Knee flexion velocity at footstrike was identified as a significant regulator of tibial accelerations in the time domain. No kinematic variables were identified as significantly related to the frequency domain properties of the signal. **Conclusions.** The findings of the current investigation suggest that sagittal plane knee flexion velocity at footstrike can regulate the magnitude of impact loading linked to the development of chronic injuries.

Key words: regulation, kinematics, accelerometer

Introduction

Distance running provides numerous physiological health benefits [1] but has also been shown to cause high rates of chronic injuries estimated to affect between 30% and 75% of runners every year [2, 3]. One of the manifestations of the foot impacting the surface during running is the generation and transmission of a vertical transient shock wave that propagates through the musculoskeletal system and is considered to be a key aetiological component in the development of injury [4].

The shock waves generated during running can be quantified by examining the impact forces experienced at the tibia by accelerometry. The most accurate way of measuring tibial acceleration magnitude during dynamic movements is by mounting an accelerometer directly to the tibia via a bone pin [5]. However, this method is not used frequently as it is invasive and requires surgical intervention. Skin-mounted accelerometers, in contrast, feature easy application although acceleration signals from these devices have been reported to be approximately twice the magnitude of bone-mounted systems [5]. Nevertheless, unwanted high-frequency signal components caused by skin interaction can be reduced with the use of a Butterworth low-pass filter. Hence, tibial accelerations are quantified more commonly by the attachment of an accelerometer to the

skin frequently on the anterior-medial aspect of the tibia [4–11].

As the impact transient that is generated by footstrike propagates through the musculoskeletal system [12], a number of mechanisms exist for the regulation of internal or external impact accelerations [12]. Many analyses in this area have examined the influence of footwear with varying levels of midsole density on the impact load experienced by the lower extremities [4, 10–11]. The current consensus is that footwear with more advanced midsole cushioning properties serves to attenuate both the time and frequency domain properties of tibial acceleration signal [10].

Typically, authors quantify lower extremity impact severity by examining the amplitude of tibial shock signals as a function of time. Previous investigations have shown that it is also possible to separate the components of the signal due to tissue resonance and acceleration as a result of foot contact using a frequency analysis algorithm known as fast Fourier transform (FFT) [9]. This was an important breakthrough as higher frequency accelerations have been linked to the aetiology of injury [10, 11]. Using this method, the median power frequency of the acceleration signal can be obtained, allowing frequency shifts in signals mediated through extrinsic parameters such as surface or footwear to be quantified and subjected to statistical analyses.

Passive tissues and joint movements have also been shown to moderate the magnitude of impact load [13]. Denoth [14] proposed the theory of effective mass whereby impact acceleration following footstrike is governed

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by the knee flexion angle at touchdown. Lafortune et al. [16] and Bobbert et al. [15] examined the influence of knee flexion angle and surface stiffness on tibial acceleration magnitude using the human pendulum approach. A strong effect was observed for surface stiffness, yet only minor impact acceleration differences were observed for alterations in knee angle at foot contact from 0° to 40° flexion. Additional analyses have examined the potential influence of the amplitude and velocity of foot motions in the frontal and sagittal planes during initial foot contact [17, 18]. However, the results of these studies are conflicting and reliable conclusions have yet to be drawn.

Therefore, there still remains a paucity of research regarding the protection afforded by joint alignment from transient impact accelerations during running. The aim of the current investigation was to determine the influence of 3-D lower extremity kinematics on the passive regulation of both time and frequency domain characteristics of tibial acceleration during running. This study tests the hypothesis that 3-D kinematic parameters can serve as significant regulators of both the time and frequency components of tibial accelerations during running.

Material and methods

Forty male participants who were free of musculoskeletal injury volunteered to take part in the study. The mean characteristics of the participants were: age 30.41 ± 5.25 years, height 178.31 ± 6.47 cm and body mass 78.43 ± 5.57 kg. All participants were classified as natural rearfoot strikers by exhibiting a clear first peak in their vertical ground reaction force. A statistical power analysis was conducted using G*Power software (Faul, Erdfelder, Lang, Buchner; Germany) using a moderate effect size [19] to reduce the likelihood of a type II error and determine the minimum number participants needed for this investigation. It was found that the acquired sample size was sufficient to provide for more than 80% statistical power. The study was approved by the university's ethical committee, and all participants provided written informed consent in accordance with the guidelines outlined in the Declaration of Helsinki.

Participants ran at a fixed velocity of $4.0 \text{ m} \cdot \text{s}^{-1}$ over a Kistler force plate (Kistler Instruments Ltd., UK) embedded in the floor (Altrosports 6mm, Altro Ltd., UK) of a 22 m long biomechanics laboratory. Running velocity was quantified using Powertimers 300 series infrared timing gates (Newtest Oy, Finland) where a maximum deviation of $\pm 5\%$ from the set velocity was allowed. Stance time was defined between the times when 20 N or greater vertical force was applied to the force platform [20].

Kinematics and tibial acceleration data were synchronously collected. Kinematic data was captured at 250 Hz via an eight camera optoelectronic motion cap-

ture system (Qualisys Medical AB, Sweden). Calibration of the system was performed before each data collection session. Only calibrations with average residuals of less than 0.85 mm for a 750.5 mm wand length and points above 3000 in all cameras were accepted prior to data collection.

The marker set used for the study was based on the calibrated anatomical systems technique (CAST) [21]. In order to define the right foot, shank and thigh, retro-reflective markers were attached unilaterally to the calcaneus, 1st and 5th metatarsal heads; medial and lateral malleoli; medial and lateral epicondyle of the femur; and greater trochanter. To define the pelvis, additional retro-reflective markers were placed on the anterior (ASIS) and posterior (PSIS) superior iliac spines. Rigid tracking clusters were positioned on the shank, thigh and pelvis. The pelvic cluster was positioned at the sacrum. Each rigid cluster comprised four 19 mm diameter spherical reflective markers mounted to a thin sheath of lightweight carbon fibre with length to width ratios in accordance with Cappozzo et al. [22]. The foot was tracked using additional markers which were glued to the foot close to the positions of the calcaneus and 1st and 5th metatarsal heads. Hip joint centre was determined using regression equations based on the positions of the ASIS and PSIS markers [23]. A static trial was conducted with the participant in the anatomical position in order for the positions of the anatomical markers to be referenced in relation to the tracking clusters, after which they were removed.

A tri-axial (ACL 300, Biometrics, UK) accelerometer sampling at 1000 Hz was utilized to measure accelerations at the tibia. The device was mounted on a piece of lightweight carbon fibre material using a protocol previously outlined by Sinclair et al. [4, 9–11]. The combined mass of the accelerometer and mounting instrument was 9 g. The voltage sensitivity of the signal was set to 100 mV/g, allowing adequate sensitivity with a measurement range of ± 100 g. The device was attached securely to the distal anterior-medial aspect of the tibia in alignment with its longitudinal axis 8 cm above the medial malleolus. This location was selected to attenuate the influence of ankle rotation on acceleration magnitude [24]. Strong non-stretch adhesive tape was placed over the device and leg to avoid overestimating acceleration due to tissue artefact.

A successful trial was defined as one where the participant ran within the specified velocity range, where all tracking clusters were in view of the cameras, the foot made full contact with the force plate and no evidence of gait modifications due to the experimental conditions was observed. Runners completed ten successful trials.

The trials were processed in Qualisys Track Manager (Qualisys Medical AB, Sweden) in order to identify anatomical and tracking markers and then exported as C3D files. Kinematic parameters were quantified using

Visual 3-D (C-Motion, USA) after marker data were smoothed using a low-pass Butterworth 4th order zero-lag filter at a cut off frequency of 12 Hz. This frequency was selected as being the frequency at which 95% of the signal power was below stance phase marker data from the current study. Three-dimensional kinematics of the hip knee and ankle joints were calculated using an XYZ Cardan sequence of rotations (where X is flexion-extension, Y is abduction-adduction and Z is internal-external rotation) [25]. All data were normalized to 100% of the stance phase and then the processed gait trials were averaged. The 3-D kinematic measures from the hip, knee and ankle extracted for statistical analysis were 1) angle at footstrike, 2) angle at toe-off, 3) range of motion during stance, 4) peak angle during stance, 5) relative range of motion from footstrike to peak angle, 6) angular velocity at footstrike, 7) angular velocity at toe-off and 8) peak angular velocity.

The acceleration signals were filtered using a 60 Hz Butterworth zero-lag 4th order low pass filter in accordance with the recommendations of Lafortune and Hennig [7] to prevent any resonance effects. Peak positive axial tibial acceleration was defined as the highest positive acceleration peak measured during the stance phase. To analyse data in the frequency domain, a fast Fourier transformation function was performed and median power frequency content of the acceleration signals was calculated. The frequency content of the tibial acceleration signal was examined using Labview software (National Instruments, USA).

Exploratory factor analysis was used to select a smaller number of variables to be included in regression analysis. This preliminary analysis yielded eight separate factors and the variables with the highest loading for each factor were extracted. These variables were: peak knee abduction, peak ankle dorsiflexion, coronal plane hip angle at toe-off, peak knee internal rotation, sagittal plane knee velocity at footstrike, peak transverse plane ankle velocity and sagittal plane hip velocity at footstrike. Multiple regression analyses, with peak tibial acceleration and median power frequency of the tibial acceleration signal treated as the criterion variables and the aforementioned 3-D kinematic parameters as independent variables, were performed using a forward stepwise procedure with significance accepted at the $p \leq 0.05$ level. The independent variables were examined for collinearity prior to entry into the regression model using a Pearson's correlation coefficient matrix, where those exhibiting a high collinearity of $R \geq 0.7$ were removed. All statistical procedures were conducted using SPSS ver. 20.0 software (IBM, USA).

Results

The overall regression model for the time domain of the tibial accelerations yielded an $R = 0.39$, $R^2 = 0.15$ and adj. $R^2 = 0.13$, $p \leq 0.01$. One biomechanical param-

eter, knee angular velocity at footstrike adj. $R^2 = 0.13$ ($B = 0.39$, $t = 2.66$) was found to be a significant predictor of peak tibial acceleration. Regression analyses for the frequency components of the tibial acceleration signal revealed that no kinematic variables served as significant regulators of the medial tibial acceleration frequency $R = 0.34$, $R^2 = 0.11$ and adj. $R^2 = 0.09$, $p > 0.05$.

Discussion

The aim of this study was to examine the passive kinematic mechanisms linked to the regulation of tibial accelerations during running. This study represents the first to analyse the discrete kinematic parameters associated with the attenuation of both time and frequency domain characteristics of tibial acceleration signals using regression analyses.

The first key observation of the current investigation was that a single kinematic parameter (knee flexion velocity at footstrike) served as a significant regulator of time domain peak tibial accelerations. This observation makes intuitive sense and corresponds with previous analyses [14]. The knee has been shown to influence effective mass [14], and increased knee flexion rates have been observed in studies using hard surfaces and footwear with reduced midsole cushioning properties [16]. Furthermore, in analyses using simulation modelling, the knee flexion rate at footstrike has been shown to regulate impact loading as a result of footstrike. This investigation has confirmed these observations in vivo using human runners rather than a musculoskeletal modelling technique. The findings from this study may have specific implications for the modification of running technique aimed towards reducing impact shock. The outcomes of previous analyses using biomechanical feedback to alter running mechanics in order to reduce impact kinetics have generally been inconclusive [11]. Nonetheless future work should still focus on implementing interventions to improve shock attenuation, particularly in runners who are susceptible to impact related injuries.

An additional observation of this study is that the frequency domain characteristics were not attenuated by any of the discrete kinematic parameters entered into the regression model. Therefore, it appears that the time domain peak tibial accelerations and frequency domain median power frequency are two distinct and separate components and necessitate different attenuation mechanisms. Passive mechanisms have previously been found to be responsible for damping the high frequency components of impact forces and the impact shock wave [26–29], although the results of the present study cannot confirm this. Previous analyses have documented that the distal heel pad and running footwear with more advanced midsole cushioning systems are able to reduce the high frequency components of tibial acceleration signal. Thus, it appears based on the current

observation that extrinsic as opposed to intrinsic factors are responsible for the attenuation of the frequency components of the tibial acceleration signal.

Whilst the results of the current study appear to show that lower extremity kinematic motions can significantly influence impact acceleration magnitude, there remains a large proportion of unexplained variance. Therefore, the remaining variance with regards to understanding the mechanisms behind the regulation of impact remains unknown. Future research may wish to consider additional parameters in an attempt to determine where the remaining variance lies. That this study did not evaluate the electromyographic potentials of the lower extremity muscles may serve as a limitation as muscle pre-activation, particularly in the period prior to foot-strike, has been proposed as one of the mechanism that may attenuate loading of the lower extremities during running [11, 30]. It is recommended that future analyses examine this pre-activation mechanism in conjunction with 3-D kinematic analyses.

The use of cushioned or shock-absorbing shoe midsoles has been suggested as a mechanism to regulate the impact forces associated with running, independent of kinematic adjustments. It is understood that cushioned footwear midsoles provide a deceleration mechanism that is able to attenuate the magnitude of tibial accelerations during running [10]. Therefore, the depth of the midsole interface in both loaded and unloaded conditions, which were not considered in the current study, should be the focus of both comparative and correlational analyses of impact loading in the future.

Tibial accelerometry is a complex function and methodological procedures can affect the efficacy of collected data in both the time and frequency domain [4]. The tibial acceleration signal is influenced by centripetal acceleration due to the sagittal plane angular motion of the tibia [4]. It has been established that adjustments for angular motion are still required despite the utilization of a distally mounted device [7]. Further exploration is needed to determine the necessary signal corrections for angular acceleration and the influence that they may have on the time and frequency domain properties of acceleration signals.

In addition to this, alternatives to the traditional FFT technique have been advocated. Joint time-frequency distribution (JTFD) analysis has also used to examine how the power spectrum of a signal alters over time [31]. Compared with the more traditional FFT, which provides only the average of the power spectrum, the JTFD technique allows the instantaneous power spectrum to be studied. For this reason it has been suggested as a more suitable method for the quantification of transient waveforms [32]. It has been shown that higher frequencies of tibial accelerations were observed using JTFD analysis during running in comparison with FFT [33]. It is recommended that future analyses examine the efficacy

of different frequency analysis techniques and the influence of kinematic motions on their regulation.

A final limitation of the current investigation that needs to be addressed is the all-male sample. Previous analyses have established that impact acceleration mechanics differ between sexes [34–35]. Thus, it is unlikely that the results of the current investigation can be generalized to female runners. Furthermore, it has also been documented that females exhibit different lower extremity kinematics compared with males [5, 36], with it being unlikely that female runners regulate impact accelerations in the same manner as males. Future research should therefore seek to repeat the current investigation using a female sample.

Conclusions

This study confirms that kinematic parameters can influence tibial acceleration magnitude during running. Future works should consider the influence of running kinematics on the passive attenuation of the frequency components of tibial accelerations in order to reduce the incidence of injury in runners.

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