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Original research

The effects of preferred and non-preferred running strike patterns on tissue vibration properties

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ABSTRACT

Objectives: To characterize soft tissue vibrations during running with a preferred and a non-preferred strike pattern in shoes and barefoot.
Design: Cross-sectional study.
Methods: Participants ran at 3.5 m s^{-1} on a treadmill in shoes and barefoot using a rearfoot and a forefoot strike for each footwear condition. The preferred strike patterns for the subjects were a rearfoot strike and a forefoot strike for shod and barefoot running, respectively. Vibrations were recorded with an accelerometer overlying the belly of the medial gastrocnemius. Thirteen non-linearly scaled wavelets were used for the analysis. Damping was calculated as the overall decay of power in the acceleration signal post ground contact. A higher damping coefficient indicates higher damping capacities of the soft tissue.
Results: The shod rearfoot strike showed a 93% lower damping coefficient than the shod forefoot strike ($p < 0.001$). A lower damping coefficient indicates less damping of the vibrations. The barefoot forefoot strike showed a trend toward a lower damping coefficient compared to a barefoot rearfoot strike. Running barefoot with a forefoot strike resulted in a significantly lower damping coefficient than a forefoot strike when wearing shoes ($p < 0.001$). The shod rearfoot strike showed lower damping compared to a barefoot rearfoot strike ($p < 0.001$). While rearfoot striking showed lower vibration frequencies in shod and barefoot running, it did not consistently result in lower damping coefficients.
Conclusions: This study showed that the use of a preferred movement resulted in lower damping coefficients of running related soft tissue vibrations.

1. Introduction

Impact forces in heel–toe running are caused by a collision between the foot and the ground and the sudden deceleration of the mass of certain body segments. As a result of this collision, a shock wave is initiated, which travels through the body from the foot to the head.[1] Particular focus has been given to a potential association between impact forces, loading rates and lower extremity stress fractures.[2] Based on the contradictory and sometimes inconclusive results, it was proposed that the effects of impact forces on the musculoskeletal system are not well understood.[3]

It is suggested that further information about the effects of impact forces on the human body may be found through the investigation of soft tissue vibrations initiated by the impact-related shock waves. The vibrations of soft tissue compartments during locomotion are typically heavily damped.[4–6] A potential reason for this damping may be that the locomotor system tries to minimize detrimental effects of repetitive vibrations. For example, excessive high vibration frequencies are thought to cause bone and ligament damage during repetitive loading cycles.[7,8] In humans, it has been shown that exposure to repetitive vibrations can result in a decrease of motor unit firing and force production.[9] In a repetitive movement such as running, soft tissue vibrations, shock attenuation and damping occur during every step. It was previously suggested that the damping of impact related soft tissue vibrations may serve as an indicator for the amount of work muscles have to do in order to minimize vibrations.[3,5,6] It has been shown that horses have muscles with appropriate architecture and properties to damp impact related vibrations.[10] When these muscles were activated, vibrations were highly damped in the first 100 ms following impact.

Previous research in the field of human locomotion has shown that shock wave attenuation and damping can be affected by stride characteristics,[11,12] speed[12] and leg muscle activation.[5,6,13] However, to date it is not clear what the differences are in vibration damping between barefoot and shod running. To the best of our knowledge, the difference in the magnitude of damping between shod and barefoot running has not yet been investigated. It is

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suggested that the time- and frequency-dependent damping properties might play a role to reduce impact related vibrations that may have detrimental effects if they occur repetitively in an excessive manner. Thus, it is of interest to understand how damping characteristics change between different running conditions or strike types.

While a variety of variables have been compared between barefoot and shod running as well as between different strike patterns, little mention has been made as to the comparison of preferred and non-preferred strike patterns for a specific footwear condition. However, some research has been done on the preference of a specific strike pattern with respect to surface hardness as well as speed[14,15] showing that surface hardness and running speed influence the preference of a strike pattern.

The preference of a strike pattern for a given footwear condition will further be referred to as a preferred movement pattern in this manuscript. The connection between damping behavior and a preferred movement pattern is not clear. It is known that optimal locomotion performance is often achieved through preferred movement patterns (e.g. speed, strike frequency, step length). We, therefore, are interested to test the hypothesis that the characteristic damping behavior for a preferred movement might be optimized in a similar way.

In summary, it remains unknown whether the damping behavior is primarily driven by different strike patterns, the difference of shod and barefoot running or by the use of a preferred movement pattern. Thus, the purpose of this study was to compare the damping of vibrations caused by impacts during running in a shod and a barefoot condition. Both conditions were tested for a preferred and a non-preferred strike pattern. It was hypothesized that the preferred strike pattern in each condition requires less damping compared to a non-preferred strike pattern (H1).

2. Methods

Twelve healthy, trained, male participants were recruited for this study (25.56 years (2.88), 178 cm (8.31), 71 kg (6.69), mean and SD).

The participants were active heel strike runners, experienced in shod running with a weekly mileage ranging from 15 to 30 km. All subjects had no previous experience with barefoot running or minimal footwear. Participants were free from lower extremity injury for the last 6 months prior to data collection. The University of Calgary's Conjoint Health Research Ethics Board approved the study protocol, and all participants gave their written consent after being informed of the nature of the study.

Each participant ran on a treadmill (Quinton Q65 with a rigid steel deck, Quinton 645 controller) at a speed of $3.5\,\mathrm{m\,s^{-1}}$ in their own comfortable running shoes as well as barefoot. The treadmill was set to a slope of 1% incline to reflect the energetic cost of outdoor running. Prior to data collection, participants were asked to run barefoot and shod without any instructions by the researcher to identify their preferred strike pattern using two accelerometers. For all participants in this study, the preferred strike pattern for the barefoot condition was a forefoot strike (FFS) while the preferred strike pattern for the shod condition was a rearfoot strike (RFS). Each subject was wearing their own running shoes. The shoes were typical running shoes with a cushioned elevated heel and similar material properties to ensure that the damping coefficient was not significantly influenced by the shoe itself. The following four running conditions were tested for each subject. Each running conditions was tested for 5 min. While the first 4 min were used as a familiarization period, the last 60 s were used for data collection. Running in shoes was tested using the preferred RFS pattern as well as a FFS. Similarly, the preferred FFS and a RFS condition were tested while participants were running barefoot. The order of testing was randomized. Between the experimental conditions participants were given at least a 3-min rest period to avoid the influence of fatigue. Subjects were asked to indicate if they needed more resting time before continuing with the next condition.

Vibrations were measured from the soft tissue compartment overlying the muscle belly of the medial gastrocnemius using a skin-mounted tri-axial accelerometer (ADXL 78, Analog Devices USA, measuring range 35 g, frequency response of 0–400 Hz) with a sampling frequency of 2400 Hz. The accelerometers were attached to the skin surface using medical adhesive glue and then pre-loaded to the skin with stretch adhesive bandage to improve congruence of motion with the soft tissues. The vertical axis of the accelerometer was aligned to be parallel to the long axis of the shank. Two single axis accelerometers were used to identify time of ground contact. The first single axis accelerometer was attached to the heel and the second one to the fifth metatarsophalangeal joint (MTPJ). If a rapid deceleration of the heel accelerometer occurred prior to the one of the MTPJ the strike pattern was classified as a RFS. Similarly, a forefoot strike was identified if the deceleration of the MTPJ accelerometer occurred earlier than the one of the heel accelerometer.

Accelerometer data from the axis that was aligned parallel with the longitudinal axis of the tibia collected during the shod and barefoot running were analyzed using MATLAB (Version 7.10.0, The MathWorks Inc., Natick, MA, USA). A baseline correction was used for the accelerometer data to correct for the influence of gravity. Twenty steps were analyzed for each subject in each of the four conditions. Soft tissue vibrations were quantified for a time window of 200 ms post-ground contact capturing the impact shock related vibrations. A time window of 200 ms was chosen since the power of the acceleration signal diminished below 5% of the maximum power by this time for all participants.

The frequency content of all steps was calculated using a Fast Fourier Transformation (FFT) and was represented by a power spectrum. For each power spectrum the peak (first mode), mean and median frequencies were calculated and averaged across all steps for each condition.

A filter bank of 13 non-linearly scaled wavelets was originally developed for the analysis of electromyograms[16] and was adjusted for the analysis of vibration signals of soft tissue compartments.[4] The wavelet analysis represents a convolution of the real and imaginary part of each wavelet with the raw signal. The power is obtained by adding the squared values or the real and imaginary part of the transformed signal. Summing up the power extracted by all the wavelets resulted in the overall power of the acceleration signal, which typically decays exponentially.[4] The logarithm of this decay was used to transform an exponential to a linear relationship, with the slope of this linear portion representing the damping coefficient of the vibration signal (Fig. 1). The decay of the logarithm of the overall power of the vibration signal was analyzed using a least square minimization algorithm to determine the overall damping coefficient. For a detailed description of the computational considerations of the methodology, the reader is referred to the manuscript by Enders et al.[4]

Mean damping coefficients were calculated for each subject, for all conditions. A higher value of the calculated damping coefficient indicates a higher dampening of the soft tissue vibrations. The damping coefficients were normalized to the preferred strike pattern in the shod running condition within each subject by dividing the damping coefficient of each trial by the mean damping coefficient that was obtained for the preferred strike pattern in the shod running condition.

Statistical analysis was made using SPSS version 19 (SPSS Inc., Chicago, IL). Analysis of variance (ANOVA) was applied after normal distribution of the damping coefficients was confirmed. Differences