IMPACT SHOCK MEASUREMENTS DURING RUNNING: CORRECTION FOR ANGULAR MOTION OF THE SHANK IS NECESSARY

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INTRODUCTION

Lower limb impact shock severity during running and other sporting activities is commonly quantified using accelerometers attached distally on the lower leg. This approach has been used to examine load attenuation mechanisms of the body and determine the efficacy of different cushioning systems in footwear and flooring surfaces (e.g. Hamill, 1999) and better differentiate between various impact conditions in comparison to data obtained from a force platform (Lafortune and Hennig, 1992).

But, accelerometry is not a straightforward approach and often methodological issues can influence the reliability of data from acceleration transducers. The magnitude of the signal obtained from the accelerometer is dependent on bone acceleration, mounting interaction, angular motion and gravity (Lafortune and Hennig, 1991; Nigg & Herzog, 1999). In order to distinguish the bone acceleration or impact shock the other components need to be identified. An important influence on the transducer signal that is often ignored or dismissed is the centripetal acceleration due to angular sagittal plane motion of the lower leg (shank) about the ankle during running ground contact (which acts in the opposite direction to the axial impact shock). This acceleration due to angular motion is governed by the square of the angular velocity of the shank and the distance of the transducer from the axis of rotation (ankle joint). Many researchers assume that a distal mounting of the accelerometer closer to the axis of rotation renders the influence of angular motion on the transducer axial shock signal to be minimal and that no correction is necessary. However this may not be the case if the magnitude of the angular velocity is underestimated due to high frequency components of the displacement data of the shank either being filtered out or not being adequately captured.

This study examined the influence of high frequency components of the angular velocity of the shank on the impact shock during running using a distally mounted transducer. The transducer signal corrections for angular motion were compared between shod and barefoot running.

METHODS

Two male recreational runners ran along a 20 metre runway at a comfortable running speed that they could reproduce consistently (approx. 4.5 $m.s^{-1}$). They contacted a force platform (Kistler) midway and ground reaction force and axial shank shock were recorded at 2000Hz using a miniature, uniaxial accelerometer (EGA100, Entran, Ltd) attached to a flat balsa wood base. The accelerometer and its mounting was glued to the skin overlying the antero-medial aspect of the tibia 12 cm above the medial maleolus (to approximate a distal location used in the literature). The transducer was fixed tightly with strapping and also the skin was pinched (stretched) and fixed in position below the transducer location. These procedures improved the mechanical coupling of the accelerometer mounting to the tibia and moved its resonance frequency to 75Hz or above.

Simultaneously, kinematics of the foot and shank were automatically captured at 1000Hz by a six-camera ProReflex system (Qualisys Inc., Sweden). Each subject performed 5-10 trials barefoot and while wearing well-cushioned running shoes. Sagittal plane displacement data of the shank was filtered using a butterworth low-pass filter with a 60Hz cut-off frequency before calculating shank angular velocity so that high frequency components were maintained. Over sampling at 1000Hz allowed noise in the data to be predominantly above 60Hz. Shank impact shock from the transducer was also low pass filtered at 60Hz (to avoid any resonance effects) and peak shock was quantified after the contributions of shank angular motion and gravity were accounted for. In addition, spectral analysis of the signal from the accelerometer was performed before and after correcting for angular motion and gravity.

RESULTS AND DISCUSSION

Peak angular velocity of the shank was 12.0 ±1.1 and 17.1 ±2.0 rad.s^{-1} for shod and barefoot running, respectively. These peak values were substantially higher than the 8 – 10 rad.s^{-1} values typically reported in the literature (e.g. Milliron and Cavanagh, 1990; Lafortune and Hennig, 1991) for cushioning knee flexion. The higher rates were directly associated with high frequency components in the 15-55Hz frequency range and most authors tend to filter out these components of the signal (e.g. low-pass cut-off frequency of 15Hz or less). The time of this peak was close to the time of peak axial shock for both shod and barefoot running and this overlap allowed the possibility of angular motion effects on the shock signal as described earlier. Peak impact shock was increased by 1.5-3g depending on the subject and shod condition after correction for angular motion and gravity (Table 1). The typical correction for gravity was about 1g during the initial impact phase as the shank was positioned predominantly within 20 degrees of vertical.

Subj.	BAREFOOT			SHOD		
	Correction (g)	Peak UN	Peak COR	Correction (g)	Peak UN	Peak COR
A	2.97±0.93	17.76 ±1.74	20.73 ±2.56	1.56 ±0.18	8.51 ±1.43	10.07 ±1.29
B	1.53± 0.59	16.34 ±1.60	17.87 ±2.07	1.86 ±0.48	8.67 ±0.97	10.52 ±1.27

Table.1 Mean peak uncorrected shock (UN) and corrected shock (COR)

For subject A in this study the difference between barefoot and shod peaks when corrected (10.67g) and uncorrected (9.25g) gave differing results. The magnitude of change in peak values was higher for the barefoot condition in subject A and higher for the shod condition in subject B. Power spectra of the shock curves revealed that total power below 60Hz was increased by 33% after correction. There were different increases in signal power for specific frequency bands. On average, power in the 8-13Hz band increased by 30.2 ± 7.1% and in the 18-23Hz band by 20.9 ± 8.6%. This was likely due to a modal frequency of around 8Hz in the angular motion correction curve.

CONCLUSIONS

Despite using a distally-mounted accelerometer on the shank, the high frequency components of shank angular velocity during early ground contact dictated that correction for angular motion was necessary in order to determine impact shock magnitude during running. This is particularly important for accurate comparisons between experimental impact conditions where angular velocity of the shank is modified (such as changes in running speed and stride length, uphill versus downhill running, and fatigued running conditions etc.). In this study, the comparison of peak impact shock between barefoot and shod running was influenced by the angular motion correction. In addition, the correction for angular motion influenced the shank shock power spectrum with gain in signal power particularly prevalent in the 8-13Hz frequency band. Consequently, investigations of shock transmission that determine the gain/attenuation spectra from shank shock to head shock should therefore also be interpreted with caution if corrections for angular motion have not been applied.

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