VALIDITY OF WEARABLE TECHNOLOGY TO MEASURE PEAK IMPACT DURING HIGH-INTENSITY TREADMILL RUNNING

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The purpose of this study was to identify the validity of an upper-body mounted accelerometer to measure peak acceleration during high-intensity treadmill running. A twelve camera motion analysis (MA) system was used as the criterion measure with markers placed on and close to the accelerometer. Ten peak impacts per participant were compared (*n* = 390). All accelerometer values were significantly different between the MA unit and T6 reflective marker's acceleration data. Smoothing accelerometer data at 8 and 6 Hz provides an acceptable indirect measure of peak impact acceleration performed during high-intensity running. Therefore, smoothing algorithms should be incorporated into the commercially available software that the devices are supplied with.

KEY WORDS: Motion analysis, accelerometer, acceleration, collision.

INTRODUCTION: In field team sports, impacts typically occur as a result of running foot-strikes, landings and player contact (Young, Hepner & Robbins 2012). Historically, impacts from running foot-strikes are measured in a laboratory setting, using large, expensive and immobile devices, such as force plates (Besier et al. 2001b; Keller et al. 1996). This situation limits the understanding of the 'in field' demands imposed on athletes. More recently, field team sports have adopted accelerometers integrated within commercially available wearable tracking devices to quantify whole body load. These devices are worn on the upper-body, and are used to quantify discrete and accumulated (over time) accelerations recorded by the tri-axial accelerometer. Research has shown these devices can automatically detect (Kelly et al. 2012) and quantify (number and intensity) collisions in rugby (Gabbett, Jenkins & Abernethy 2010). They can also quantify the physical demands imposed during basketball (Montgomery, Pyne & Minahan 2010) and Australian football (Young, Hepner & Robbins 2012). Excellent reliability of these devices has been shown in mechanical testing (CV = 0.91 to 1.9%; (Boyd, Ball & Aughey 2011), while poorer results (CV = 10.0 – 30.8%) were reported during jumping and landing when using ground reaction force as the criterion measure (Tran et al. 2010). The investigation of Tran and colleges (2010) also showed filtering the raw accelerometer data at 20 Hz improved the relationship with the criterion measure. Tran et al.'s (2010) choice of cut-off frequency was derived from previous counter-movement jump research (Bisseling & Hof 2006) with no attention to the actual frequency content of the acceleration time histories. Of further concern, the harness used to mount the tracking device on the participants was suggested to influence the accelerometers accuracy. No studies have sought to identify; the most optimal smoothing frequency, the influence of the harness mount and the validity of an upper-body mounted accelerometer integrated within wearable tracking devices to measure impact accelerations during high-intensity running. To gain confidence in the data produced by this technology and to enhance the understanding of the whole body physical demands imposed on athletes during high-intensity movements, the accuracy of the accelerometer output must be ascertained. The primary aim of this study was to examine the criterion validity of peak resultant acceleration values recorded by an accelerometer worn on the upper-body, against peak resultant acceleration values concurrently measured by a motion analysis (MA) system during high-intensity treadmill running. The secondary aim was to identify the most optimal smoothing frequency and any influence of the harness mount for an accelerometer worn on the upper-body.

METHODS: Thirty-nine recreationally active participants (twenty-eight males and eleven females: age 24.2 ± 2.5 years; height 1.79 ± 0.09 m; mass 71.6 ± 12.0 kg; mean ± SD) were recruited. The study protocol was approved by the Deakin University Ethics Committee and

written informed consent was obtained from all participants prior to testing. A single commercially available wearable tracking device (minimaxS4, Catapult Innovations, Australia) containing a tri-axial accelerometer (Konix, USA) was utilised in this study. The sampling frequency of the accelerometer was 100 Hz and the full-scale output range was ±10 G in each axis. The device was worn in a manufacturer supplied harness (Catapult, Australia) which located the device firmly in the centre of the upper-back at the level of T2. All testing was performed indoors with the GPS functionality of the device disabled. Three-dimensional (3D) positional data were collected at 200 Hz using a 12-camera MA system (Raptor-E, Motion Analysis Corporation, USA). Eleven reflective markers were attached to the body. A marker located on the lateral malleolus of each foot was used to identify foot-strike. A marker located on the unit itself and at T6 level was used to assess criterion validity. The T6 marker location was chosen to allow comparison without any harness influence on the marker, while being close enough to ascertain a reasonable comparison. All trials were completed on a calibrated motorised treadmill (Quinton Q65, Quinton Instrument Company, Washington, USA). Following familiarisation with all equipment, procedures, and the exercise protocol, participants were asked to run at a high-intensity (females 5 m/s; males 5.83 m/s) on a treadmill for 30 seconds. Acceleration (from the accelerometer) and positional data off each reflective marker were recorded simultaneously. In total, 390 comparisons were made per marker. Raw accelerometer recordings were corrected for gravity using the manufacturer supplied algorithms so that the accelerometer displayed 0 G while stationary. This data along with the raw position data from the MA system were spectrally analysed using a customised Fast Fourier transform (FFT) in Microsoft Excel (version 14.0.6112.5000). This program displayed the spectra of the signal and provided the researcher with a visual means to choose an optimal cut-off frequency for smoothing. A 4^{th} order, dual pass, digital Butterworth filter was applied to smooth the accelerometer acceleration and MA position data. The MA smoothed position data were then differentiated twice to calculate acceleration. To aid in synchronising the device signals, participants performed three countermovement jumps prior to testing, while standing within the 3D space. A customised MATLAB program (R2012a 7.14.0.739, USA) was used to synchronize the captured signals and to detect peak values during previous manually labelled steps. The accelerations captured during the countermovement jumps were used to find the offset between the devices, using cross correlation. Subsequently, the offset between the devices was subtracted from the time domain of the MA data and peaks were identified, based on previously labelled time stamps, in the original frequency of the captured accelerations. All data were log transformed as the data were not normally distributed and displayed heteroscedasticity. Non-parametric Spearman's Rho correlation coefficients (r) and typical error of the estimate expressed as a percentage (CV%) were subsequently used. Our operational definition for the CV% was small (CV% < 5), moderate (CV% >5 and < 20) and large (CV% ≥ 20). A one way (Device) ANOVA was used to determine if peak impact values recorded by each measurement device for each movement were significantly different from each other. If significant interactions existed, multiple pairwise comparisons (with Bonferroni correction) were used to identify where the significant differences lay.

RESULTS: Visual inspection of the FFT outputs suggested the optimal cut off frequency to be 6 Hz for the MA data and between 6 and 10 Hz for the accelerometer data. Additionally, a 20 Hz cut off was analysed to make the current results comparable to previous work. There was a significant main effect of device on impact acceleration, $F(6, 228) = 587.02$, $p < 0.01$. All accelerometer variables were significantly different ($p < 0.01$) from both MA measures (Table 1a and 1b). The 6, 8 and 10 Hz accelerometer measures showed the strongest correlations ($r > 0.8$) and moderate measurement errors (CV% < 10) compared to the MA marker on the unit. All accelerometer measures showed weaker conformity when compared to the MA marker located at T6 (Table 1b). Both MA measures were significantly different from each other but displayed a strong correlation ($r = 0.75$) and moderate measurement error (CV% = 11.3). Smoothing positively influenced the accuracy of the accelerometer data.

Table 1a and 1b

Summary of results for high-intensity treadmill running: Accelerometer acceleration for raw, smoothed at 20, 10, 8 and 6 Hz, compared to reflective marker acceleration located on the unit (Table 1a) and on the skin at the level of T6 (Table 1b; *n* = 390).

1a	*Device*	*Mean ± SD*	*r*	*CV%*
	MA Unit	*1.66 ± 0.30*		
Accelerometer	Raw	3.16 ± 0.82*	0.62†	13.4
	20 Hz	2.74 ± 0.66*	0.73†	11.6
	10 Hz	1.80 ± 0.36*	0.83†	9.3
	8 Hz	1.61 ± 0.30*	0.84†	9.1
	6 Hz	1.41 ± 0.23*	0.85†	9.1

1b	*Device*	*Mean ± SD*	*r*	*CV%*
	MA T6	*1.47 ± 0.22*		
Accelerometer	Raw	3.16 ± 0.82*	0.45†	14.4
	20 Hz	2.74 ± 0.66*	0.50†	13.9
	10 Hz	1.80 ± 0.36*	0.56†	13.2
	8 Hz	1.61 ± 0.30*	0.59†	13.0
	6 Hz	1.41 ± 0.23*	0.61†	12.9

Acceleration values are means ± SD; * The mean difference (Accelerometer vs. MA marker) is significant at the 0.01 level (based off the log transformed data); † The correlation is significant at the 0.01 level. MA, motion analysis; SD, standard deviation; T6, thoracic vertebrae six reflective marker; r, Spearman Rho correlation coefficient; CV%, typical error of the estimate presented as a percentage.

DISCUSSION: In field team sports, the use of wearable tracking devices is widespread and continues to grow. The accuracy of accelerometers integrated within this technology for quantifying high-intensity movements is largely unknown. In this study, peak impact accelerations in an upper-body mounted accelerometer were examined against a MA system during high-intensity treadmill running. Multiple cut-off frequencies were also assessed to identify optimal smoothing parameters. On the criterion validity of the unit, significant differences ($p < 0.01$), strong correlations ($r = 0.84$) and moderate measurement errors (CV% = 9.1) suggest the 8 Hz accelerometer measure is the best measure of impact acceleration. On the global whole body impact load (compared to the T6 marker's acceleration), smoothing at 6 Hz appears more appropriate. The FFT suggested that the majority of MA signal power was contained in the lower 7 harmonics (below 6 Hz), which falls within cut-off frequency ranges (5 – 20 Hz) typically reported in the literature (Kristianslund, Krosshaug & van den Bogert 2012). If a higher cut-off frequency were applied, more noise would be allowed to pass through the filtering process. Differentiation would then amplify this noise, leading to erroneous higher derivatives (Giakas & Baltzopoulos 1997). The FFT results for the accelerometer were less clear. There was still signal power above the 10th harmonic, but it had the characteristics of "noise" (Winter 2009). As such, multiple frequencies (6, 8 and 10) were analysed. The current investigation suggests that raw accelerometer data and a 20 Hz (previously validated (Tran et al. 2010)) cut-off frequency are both unacceptable to measure peak impact acceleration. This data were nearly double the actual peak recorded by the MA system (Table 1a). Smoothing improved the relationship with the criterion measure and reduced the measurement error found (Table 1a). For example, the 8 Hz cut-off frequency resulted in a 0.06 G difference with the unit MA measure. The harness used to attached the accelerometer to the wearer was suggested as a possible cause of the error in previous research (Tran et al. 2010). In the present study, significant overestimation of impact acceleration from the unit measure compared to the T6 measure support this observation. These results suggest that the harness causes excessive movement of the unit outside of that experienced by the body (i.e. whipping movement of the unit increases the impact acceleration recorded). Therefore, comparison of validity should be against that of the T6 measure to remove the influence of the harness. As such, a 6 Hz cut-off frequency appears to be most appropriate (Table 1b). The ability to use accelerometers to accurately quantify impact loads in the field (over the laboratory) gives practitioner's a unique insight into the impact load imposed on the musculoskeletal system during training and competition. This has implications for the prescription of training loads, injury management programs and recovery practices.

Conclusion: In field team sports, the accurate quantification of peak foot-strike impact accelerations will aid sports scientists and coaches to understand the physical demands imposed on athletes during training and competition. The findings of the present investigation support the use of accelerometer data to quantify high-intensity running impacts, if the data is smoothed appropriately. Cut-off frequencys within 6-8 Hz should be incorporated into the commercially available software that the units are supplied with. Further research is required to assess the validity of these devices during different movements and the devices ability to measure load in multiple movement that simulate field team sport.

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