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Predicting Impact Shock Magnitude: Which Ground Reaction Force Variable Should We Use?

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Abstract. Peak tibial acceleration (PTA) measured using accelerometers attached to the musculoskeletal system is considered the most effective method of quantifying impact shock magnitude as a result of footstrike during running. Ground Reaction Forces (GRFs) measured using force plates are also widely used to predict PTA. However it is not clear which is the most effective GRF variable to use. This has led to different variables being reported within biomechanics literature. This study aimed to identify which GRF variable is the most suitable for consistent and accurate prediction of impact shock magnitudes. Thirteen participants (10 male and 3 female) took part in this study. Simultaneous tibial accelerations and GRF information were recorded as participants ran at $4.0ms^{-1} \pm 5\%$ over a force platform. The relationship between various GRF parameters including, average vertical loading rates, peak instantaneous vertical loading rates (PIVLR), event times were compared to tibial shock magnitudes using Pearson correlations.

The GRF variables analysed identified that the strongest correlation ($r=0.469$) exists between the PIVLR and the PTA. This study therefore provides evidence that the most effective method of predicting PTA is via the PIVLR. This method does not require identification of a vertical GRF impact peak which is dependent on the individual researcher identifying the peak, and can be identified repeatedly across studies.

1. Introduction

There have been a number of investigations which have examined the detrimental characteristics of the human body's impact with the ground during human locomotion [1-6]. When not excessive in terms of frequency and magnitude, loading of the musculoskeletal system provides essential health benefits, including maintaining a suitable level of bone density [7-8]. However when a combined influence of the magnitude and frequency of impacts are excessive, epidemiological evidence suggests that overuse injuries such as stress fractures can occur [1].

It is commonly accepted that measuring impact shockwave transmission through the skeletal system at the tibia is most effectively accomplished using accelerometers attached directly to the underlying bone itself [9-11]. However, this methodology cannot be employed frequently as it causes much discomfort and requires invasive surgical procedures, thus its efficacy is compromised. The data from these studies has been contrasted to skin mounted accelerometers which provide a non-invasive method of estimating the actual tibial shock magnitude. Large differences have been found between the signals for skin and bone mounted accelerometers. However, it has been shown that through the use of a low-pass filter at an appropriate cut-off frequency that the large component of the signal present due to the skin interaction between the bone and accelerometer, can be attenuated and a good estimation of the bone acceleration can be recorded [12]. The skin mounted technique with the accelerometer attached tightly to the skin at the distal anterio-medial aspect of the tibia has since been used in many research publications, [1,13-17]. This method of positioning the accelerometer reduces skin interaction and minimises the effects of acceleration due to the angular motion of the tibia about the ankle joint [10].

Throughout the studies investigating GRF variables during the impact phase of human locomotive movement, there are a number of variables that are believed to be associated with the incidence of injury. Examples of these include; initial impact peak, average loading rate , instantaneous loading rate and time to peak loading rates [9,15-16,18-22].

Various ways of calculating the rate of loading have been employed in previous research. One methodology calculated the average loading rate from 20-80% of time to impact peak [16]. Similar methods calculating from 20-90% have also been used [23]. A method used by Munro and colleagues calculated the loading rate from 50N, to BW plus 50N [24]. Calculating a loading rate over the time it takes the GRF to increase by a BW does not require identification of a force peak and may therefore be a more consistence characteristic to use less prone to human error. The same is also true for studies that have reported the instantaneous loading rate [16,25-26]. By calculating the maximum difference found between each sample of GRF data, a peak loading rate during the impact phase can be found and recorded. The timing of the peak loading rate has not been examined by many studies and may be a factor that provides further information regarding the occurrence of injury in relation to GRFs. From the methodologies utilised in previous studies it is not clear if there is a conclusive way of analysing force data to most accurately predict impact shock magnitudes. A comparison of all the methodologies may provide evidence to allow identification of the most effective technique. Therefore the aim of this investigation was to determine the extent of the relationship between peak tibial accelerations and previously reported GRF parameters during human locomotion.

2. Methods

2.1. Participants

Thirteen adults (Age 30.0 $\pm$ 9.4 years; Height 1.74 $\pm$ 0.06 m; Mass 70.6 $\pm$ 8.1 kg) comprising of 10 male and 3 females, volunteered to take part in this study. All were injury free at the time of data collection and completed an informed consent form. Ethical approval for this investigation was obtained from a university ethical board.

2.2. Procedure

Participants were required to run between two sets of timing gates positioned 4m apart and either side of the force platform. The participants had a 10m run up to the plate with 10m after the plate to slow down. They were instructed to run through the second set of gates before slowing down (Figure 1). A thick crash mat was used against the end wall to allow the participants to stop safely and reduce the risk of injury through collision with the wall. Each participant was required to perform 8 good trials. A trial was considered good when the participant landed with their right foot fully in contact with the force plate with no observable adjustments made to target the force plate. Participants were required to run at $4 m.s^{-1} \pm 5\%$ measured by the timing gates.

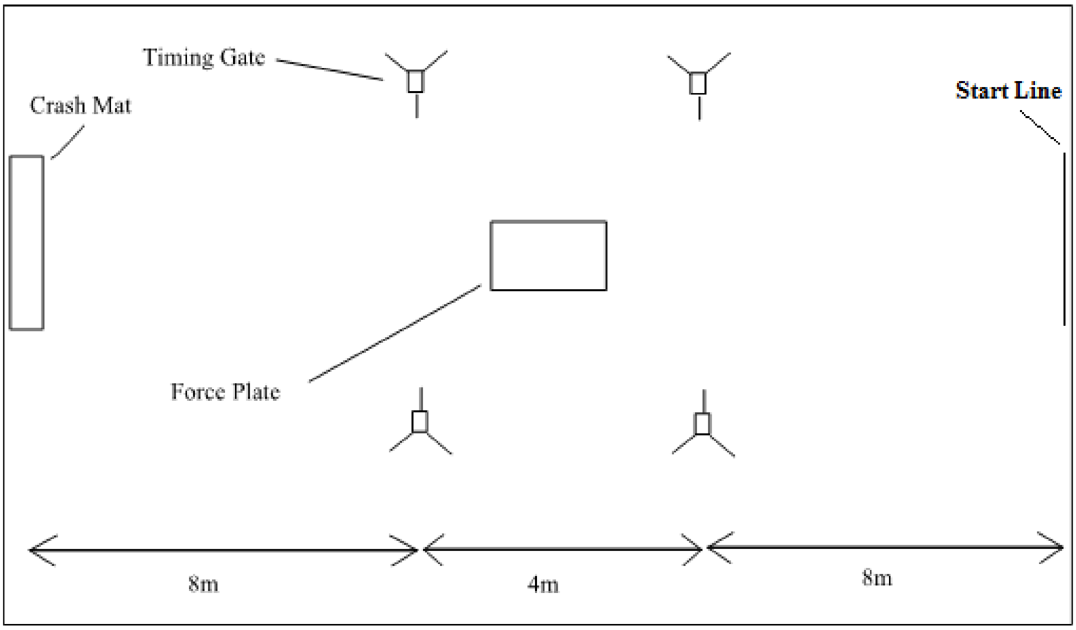

Figure 1 Setup of the biomechanics lab for data collection

A tri-axial accelerometer (Biometrics ACL300) was mounted to a lightweight carbon-fibre plate via a securely glued lightweight bolt and thread attachment. The total weight of the accelerometer and mounting system was 9g. The carbon-fibre plate was firmly attached to the shank via surgical adhesive tape. By using skin stretching techniques the plate was attached tightly so the accelerometer was positioned on the distal anterior-medial aspect of the tibia and 8cm above the medial-malleolus. The accelerometer was orientated to measure the acceleration along the longitudinal axis of the tibia. This accelerometer and attachment system was the same used in a recent publication investigating differences in fencing footwear [17]. By positioning the accelerometer near the malleolus the radius of the motion of the sensor about the ankle joint was minimised.

The accelerometer signal was set to 100mV/g providing a measurement range of ±100g. The sampling frequency was set to 1000Hz. The Analogue Data signal was recorded through Qualisys Track Manager software (OMG, Oxford), via a biometrics data collection device attached via a 20m wire. Force data was recorded through a force plate sampling at 1000Hz embedded in the ground of the biomechanics laboratory. The analogue signal was recorded simultaneously with the accelerometer data through Qualisys Track Manager. The accelerometer signal was processed through a Butterworth zero-lag low-pass filter at a cut-off frequency of 60Hz. This filter was used to exclude the component of the signal due to skin artifact and the resonance of the device, in line with the findings from previous research [12]. Software was developed (Matlab, Mathworks) to process and calculate the GRF variables incorporating manual identification of the GRF peaks from graphical data. The results were checked for consistency by analysing the GRF data individually using Microsoft Excel. The results from both methods were found to be consistent.

2.3. Statistical Analysis

Multiple bivariate Pearson correlation analyses were performed to compare the relationship between the various GRF characteristics and the tibial acceleration measured. Statistical procedures were calculated using SPSS 17.0 with significance accepted at the $p \leq 0.05$ level.

Table 1: Ground reaction force and tibial acceleration parameters

Parameter	Code	Unit
Peak Tibial Acceleration	**PTA**	**g**
1st Vertical Force Peak	**VFP1**	**BW**
Peak Instantaneous Vertical Loading Rate	**PVLR**	**BW.s^{-1}**
Average Vertical Loading Rate	**AVLR**	**BW.s^{-1}**
Average Vertical Loading Rate From 50N to 50N Plus BW	**AVL50NT50NBW**	**BW.s^{-1}**
Average Vertical Loading Rate from 20 To 80% of 1st Vertical Force Peak	**AVL20T80**	**BW.s^{-1}**
Average Vertical Loading Rate from 20 To 90% of 1st Vertical Force Peak	**AVL20T90**	**BW.s^{-1}**
Breaking Force Peak	**BFP**	**BW**
Time to peak vertical loading rate from foot down.	**TPVLR**	**ms**
Time to 1st vertical force peak from foot down.	**TVFP1**	**ms**
Time to peak tibial acceleration from foot down.	**TPTA**	**ms**

3. Results and Discussion

The VGRF (Figure 2), vertical loading rate (Figure 3) and tibial acceleration (Figure 4) data generated conventional peaks with characteristics that were expected, including peak tibial accelerations and loading rates occurring prior to the identified first vertical force peak. A first impact peak was easily identifiable for most of the data recorded, however in some cases where there were double peaks or only minor deformation of the vertical-force time curve. Thus it was necessary to take a best estimation of the impact peak from the graphed data. The mean impact force peak values reported in Table 2 were similar to those from previous human locomotion research (Cavanagh and Lafortune, 1980, McClay et al., 1994, Kersting and Bruggemann, 2006).

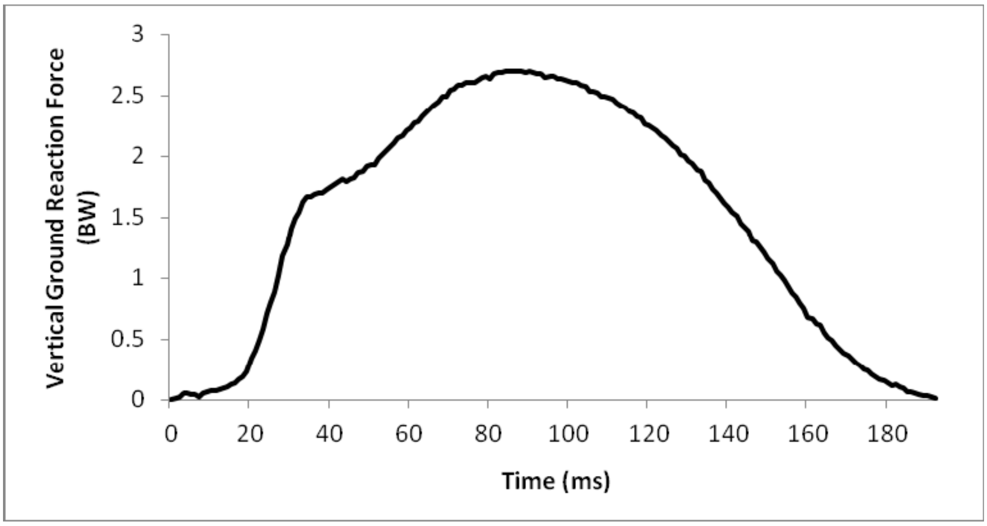

Figure 2: Representative vertical ground force data during stance

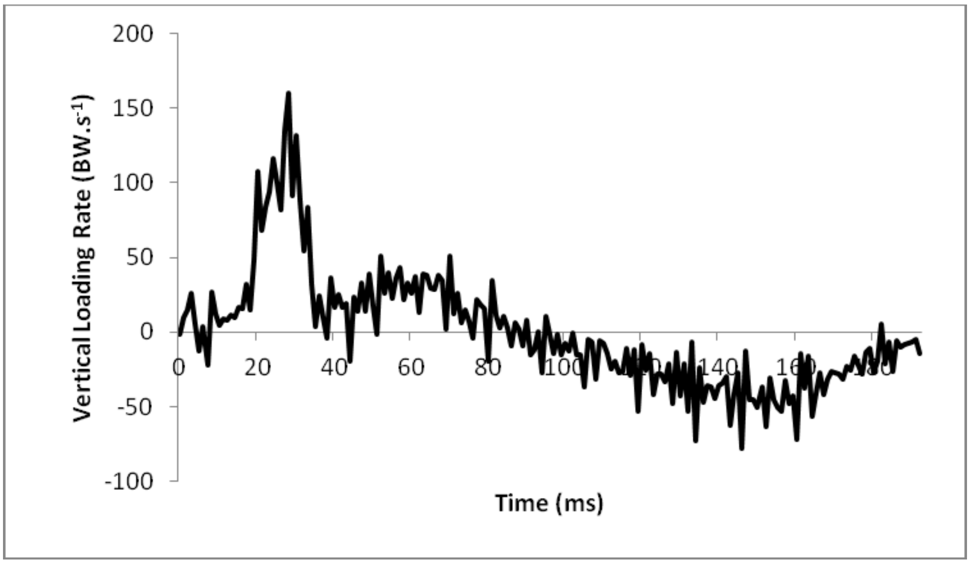

Figure 3: Representative vertical loading rate data during stance

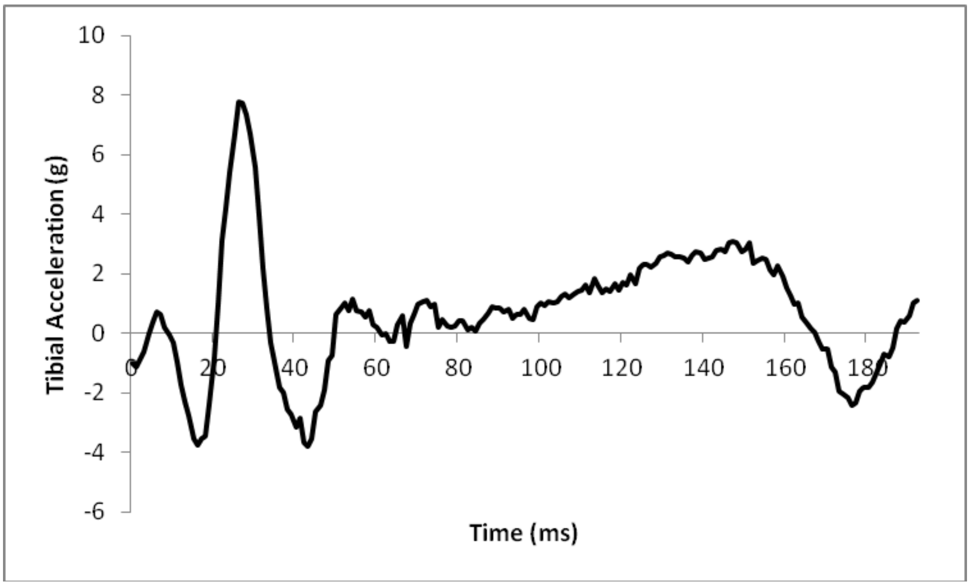

Figure 4: Representative tibial acceleration data during stance

Table 2 Mean (± standard deviation) GRF and tibial acceleration values from all participant's data

Variable	Mean Value	
VFP1 (BW)	**2.56**	**±0.3**
AVLR (BW.s^{-1})	**106.7**	**±26.4**
PVLR (BW.s^{-1})	**246.9**	**±61.6**
PTA (g)	**9.5**	**±3.3**
TVFP1 (ms)	**25.3**	**±5.4**
TVFP1 (ms)	**15.9**	**±4.3**
Time to Peak tibial acceleration (ms)	**21.7**	**±5.9**

Table 3 Correlation values between GRF variables and tibial acceleration peak values (**=P<0.05 **=P<0.001*)

Correlation Value	Pearson Correlation	Sig. (2-tailed)
VFP1	**-0.051**	**0.595**
PVLR	**0.469****	**0.000**
AVLR	**0.274****	**0.004**
AVL50NT50NBW	**0.291****	**0.002**
AVL20T80	**0.439****	**0.000**
AVL20T90	**0.439****	**0.000**
BFP	**0.326****	**0.000**
Stance Time	**-0.267****	**0.004**
TPVLR	**-0.359****	**0.000**
TVFP1	**-0.336****	**0.000**
TPTA	**0.035**	**0.718**

The data reported across the participants for the timings of events (Table 3) shows a slightly stronger negative correlation for the time to peak loading rate (r=-0.36, P≤0.05) than the 1st impact peak (r=-0.34, P<0.05). The time to peak tibial acceleration was not found to be significantly correlated (P>0.05) to the magnitude of the PTA, suggesting that the timing of the PTA during stance does not influence its magnitude. This result differs to correlations reported between the same variable and the PTA from bone mounted accelerometer research, which reported a strong correlation (r=-0.8). This provides further evidence of the effectiveness of the bone mounted technique.

From the results presented in Table 3 there are moderate to weak correlations between loading rates and peak tibial accelerations measured by a skin mounted accelerometer. Overall, the results suggest that the strongest significant correlation (r=-0.469, P<0.001) is that of the PVLR. Furthermore, the data indicates that the AVL20T80 and the AVL20T are also more effective at identifying the magnitude of PTA than measuring the AVLR. The strength of the reported correlations are less than values reported for both bone mounted accelerometer research [9] and in similar skin mounted accelerometer research [16,27]. However, the previous skin mounted investigations identified the PVLR as having the strongest correlation, which agrees with this current research. The bone mounted research did not report the PVLR variable.

4. Conclusion

This study provides evidence that the most effective way of predicting tibial accelerations is by measuring the PIVLR. This method does not require identification of a VIFP which is dependent on the individual researcher identifying the peak. In some cases this may be problematic and lead to errors. The PIVLR can be identified efficiently by software such as that developed for this research, without the time consuming identification of VIFPs. It is therefore recommended that in similar future research, the PIVLR should be reported.

5. Acknowledgements

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