COMPARISON OF TIBIAL IMPACT ACCELERATIONS: VIDEO VS ACCELEROMETER

Erich J. Petushek[1], Chris Richter[2], David Donovan[3], William P. Ebben[4] and Randall L. Jensen[1]

Dept. Health, Physical Education & Recreation, Northern Michigan University, Marquette, MI, USA[1]
Health & Human Performance Dept., Dublin City University, Dublin, Ireland[2]
Dept. Physics, Northern Michigan University, Marquette, MI, USA[3]
Dept. Health, Exercise Science & Sport Management, UW-Parkside, Kenosha, WI, USA[4]

This study compared tibial axial accelerations measured by video analysis and accelerometry. Twenty-two recreationally active adults performed three countermovement jumps. The landing tibial axial accelerations were assessed with video and an accelerometer. High reliability was demonstrated for the root mean square error between the assessment methods (ICC_{ave} = 0.872). Repeated measures ANOVA results revealed no instrumentation differences in the magnitude of the two acceleration peaks (toe and heel contact) and no difference between trials. However, first and second peaks occurred 9.6 and 4.0 ms earlier, respectively, when assessed by video. Accelerometry is a valid and reliable alternative to video analysis for the assessment of tibial impact accelerations if temporal characteristics are not of interest.

KEYWORDS: countermovement jump, landing, instrumentation, reliability, motion analysis.

INTRODUCTION: The impact from jump landings has been of interest to scientists and practitioners in order to understand plyometric intensity (Jensen & Ebben, 2007) and knee injuries (Murphy et al., 2003). Specifically, high tibial impact accelerations have been associated with increased risk of lower extremity injury (Moran & Marshall, 2006; Madigan & Pidcoe, 2003). Tibial accelerations are also an important parameter when using inverse dynamics to calculate lower extremity joint moments upon impact. Therefore, the quantification of tibial impact acceleration is critical for injury prevention and proper exercise progression.

Imaging systems using video or infrared cameras are commonly used to quantify tibial accelerations (Bisseling & Hof, 2006; Winter, 1990). This approach relies on differentiation, which amplifies the error for each successive kinematic derivative. Other problems associated with this assessment method may include: slow video feeds, timely digitization or labeling, difficulty choosing optimal filtering technique and cost. Other kinematic tools such as accelerometers may be a more affordable and time saving alternative to video.

Accelerometers are relatively inexpensive, small and can sample at high rates (>1,000 Hz), making this instrument more convenient and sensitive than video analysis to assess the quick impacts associated with jump landings. Substantial differences in tibial accelerations have been found between bone and skin mounted accelerometers (Lafortune et al., 1995). However, skin movement error is also present in marker based video analysis techniques which are commonly used to quantify tibial accelerations. Additionally, methods for attaching skin mounted accelerometers to the segments have been under review (Kavanagh & Hylton, 2008).

For convenience and non-invasiveness, skin mounted accelerometers may be a more practical approach in assessing tibial impact accelerations, however their validity compared to videography has yet to be assessed. Therefore, the purpose of this study was to compare the tibial axial accelerations, during jump landings, obtained via video analysis and accelerometry.

METHODS: Twenty two recreationally active adults (12 female and 10 male; mean ± SD; age = 21.1 ± 0.94 years; height = 170.7 ± 9.7 cm; body mass = 73.0 ± 17.8 kg) volunteered to serve as subjects for the study. Inclusion criteria included subjects who were 18-24 years old, participated in high school or college sports, and were without orthopedic lower limb pathology that restricts functioning or known cardiovascular pathology. All subjects provided informed written consent and the study was approved by the institutional review board.

Subjects performed 5 minutes of low intensity warm up on a bicycle ergometer followed by dynamic warm up. After the warm up, subjects rested for 5 minutes prior to the testing trials. The test consisted of three trials of the countermovement jump and land performed on a force platform (BP6001200, AMTI, Watertown, MA, USA) sampled at 600 Hz. One minute rest was provided between each jump.

Video analysis of the jump landing was obtained at 600 Hz (Exilim EX-F1,Casio Computer Co. LTD, Tokyo, Japan) from the sagittal plane using 2 cm diameter markers placed on the lateral greater trochanter, knee joint line, and estimated tibia center of mass which was 57% of the distance from lateral malleolus to femoral condyle (Winter, 1990). Markers were digitized using automatic digitizing software (MaxTRAQ 2D, Innovision Systems Inc, Columbiaville, MI, USA). Video data were transformed from the global to the tibial coordinate system for valid comparison to the acceleromter (Winter, 1990).The frequency content of the kinematic signals expands upon impact, thus a fractional Fourier Transform filter with a time-varying “triangular” cutoff frequency was implemented to determine tibial accelerations from the position data. This type of filter has been shown to accurately estimate tibial impact accelerations during jump landings (Georgakis & Subramaniam, 2009).

A ±50 g accelerometer (TSD109, Biopac Systems, CA, USA) was adhered to the anterior aspect of and parrallel to the tibia shaft, at the estimated center of mass. The accelerometer was adhered to the tibia with double sided tape and then LightPlast® Pro athletic tape (BSN medical, Inc. Charlotte, NC, USA) was applied, with tightness to the subject’s tolerance to ensure proper preloading. The tibial axial, or vertical, accelerations were sampled at 600 Hz. The video and accelerometer were temporally synchronized using an external trigger that elicited a square wave and LED signal.

Root mean square error analysis (RMSE) was conducted for each trial to quantify the differences between video and accelerometer tibia accelerations during the entire impact phase. The onset of the impact phase was defined when the ground reaction force rose above 5 N and ended at the lowest point of the greater trochanter marker. The reliability of the acceleration RMSE across the three trials was estimated using a two-way mixed Intra-Class Correlation (ICC) and Repeated Measures ANOVA (ReANOVA) to test for differences between the three trials. Coefficient of Variation (CV) was also calculated and equal to [(Standard deviation of the trials/Mean of the trials) * 100]. A Two Way ReANOVA (trial X method) was performed for the peak acceleration and time at peak acceleration for both the first (toe) and second (heel) peaks during the impact (see Figure 1). All statistical analyses were done using SPSS v. 18. Significance was set at $\alpha = 0.05$ for all statistical tests and follow-up pair-wise comparisons were performed with Bonferroni’s correction when significant differences were found in the ReANOVA.

RESULTS: The reliability of the acceleration RMSE across the three trials was: ICC_{ave} = 0.872 (95% Confidence Interval = 0.740 to 0.943). There were no differences across the trials (Mean (g) ± SD: Trial 1 = 4.15 ± 2.17; Trial 2 = 4.75 ± 3.27; Trial 3 = 4.20 ± 1.81), while the CV was 26.6%.

As shown in Table 1 there were no differences between the main effects of trial or method for peak acceleration during the first or second peaks ($p > 0.05$). In addition, there was also no interaction for either of the peaks ($p > 0.05$).

The time at peak acceleration exhibited a significant difference ($p < 0.001$) between the methods of assessment for both peaks, but there was no difference ($p > 0.05$) across the trials or significant interaction of method and trial (see Table 2).

Table 1
Mean and ±SD peak accelerations (g) determined via accelerometer and video across three trials (n = 22).

	Peak 1		Peak 2	
	Accelerometer	Video	Accelerometer	Video
Trial 1	8.35 ± 3.4	8.75 ± 3.8	13.20 ± 9.2	13.35 ± 8.8
Trial 2	8.65 ± 4.6	8.62 ± 4.7	11.43 ± 8.9	13.12 ± 9.7
Trial 3	8.80 ± 4.5	9.39 ± 4.2	15.35 ± 11.1	14.22 ± 6.7

Table 2
Mean and ±SD time at peak acceleration (ms) determined via accelerometer and video analysis across three trials (n = 22).

	Peak 1		Peak 2	
	Accelerometer	Video [a]	Accelerometer	Video [a]
Trial 1	34.4 ± 5.3	25.4 ± 7.4	73.1 ± 18.3	67.2 ± 17.7
Trial 2	35.3 ± 7.5	26.1 ± 7.9	73.0 ± 19.3	69.8 ± 15.3
Trial 3	35.8 ± 7.1	25.2 ± 7.4	72.5 ± 16.2	69.6 ± 12.9

[a] Significantly different from the Acceleration Method ($p < 0.001$)

DISCUSSION: This study compared tibial axial accelerations during the impact of a countermovement jump using video analysis and accelerometry. The average overall magnitude of difference between the two assessment methods during the entire impact phase, assessed by RMSE, was 4.4 g at each data point. This is a relatively large overall difference; however the magnitude of the two peak accelerations (toe and heel strike) were not different between the video and accelerometer data. There were, however, temporal differences between the assessment methods (see Figure 1), which likely contributed to the high average RMSE.

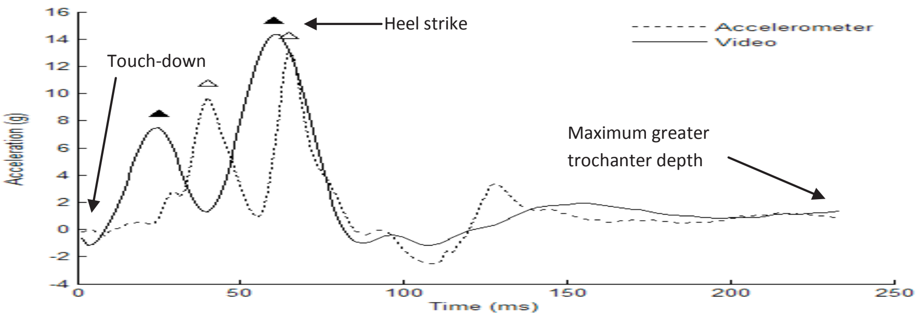

Figure 1: Sample tibial acceleration curve assessed via accelerometer (dashed) and video (solid). The triangles indicate the peaks (open = accelerometer; solid = video).

Possible explanations for the temporal differences and high RMSE may include skin movement, the choice of video filter parameter values and phase shift. Since the accelerometer (anterior) and video marker (medial), were attached to the skin in different locations, both may have undergone unique accelerations. Specifically, the additional weight of the accelerometer (17 grams) may have caused a greater amount of skin movement resulting in acceleration differences between methods. The filtering technique implemented required parameters such as: highest frequency induced during the landing, impact duration and time of maximum acceleration. The improper determination of impact duration and time of maximum acceleration could place the “triangular” time-varying cutoff threshold in an erroneous location leading to the present temporal differences. Despite temporal difference, the error between the two methods was reliable.

To the authors' knowledge, this is the first study to examine the reliability of RMSE tibial axial accelerations during jump landings using this accelerometer. Specifically, the mean differences between the two measurement methods were consistent across the three trials. The second acceleration peak of both instruments ranged from 5.5-48.5 g which are typical accelerations for jump heights ranging from 13.0-38.0 cm (Moran & Marshall, 2006; Elvin et al., 2007; Zhang et al., 2008). Jump height and tibial axial accelerations have been shown to be weakly correlated (Elvin et al., 2007). This is likely due to the variability in landing technique (Self & Paine, 2001). Therefore, a composite error (ie. RMSE) should be used to control for movement variability, when assessing instrumentation reliability during landing movements. High reliability was demonstrated with a uniaxial accelerometer with tibial impact accelerations varying from 9.8-20.7g, but attachment of the accelerometer was via strapping (Self & Paine, 2001). The taping method presently used to attach the accelerometer to the tibia seemed to be effective over the course of three jump landing impacts. Future research is needed to examine the inter-session reliability of this specific accelerometer and the reliability and validity of accelerations at other body segments of interest.

CONCLUSION: No differences were found between video analysis and accelerometry assessed peak accelerations. However, the times at peaks were different between the two methods. The accelerometer used in this study and attachment method yielded repeatable peak accelerations during jump landings. Therefore, accelerometry would be an inexpensive, time efficient alternative to video analysis for the assessment of peak tibial impact accelerations if temporal characteristics are not of interest.

REFERENCES:

Bisseling, R.W., and Hof, A.L. (2006). Handling of impact forces in inverse dynamics. *Journal of Biomechanics,* 39, 2438-2444.

Elvin, N.G., Elvin, A.A., and Arnoczky, S.P. (2007). The correlation between ground reaction force and tibial acceleration in vertical jumping. *Journal of Applied Biomechanics,* 23, 180-189.

Georgakis, A., and Subramaniam, S.R. (2009). Estimation of the second derivative of kinematic impact signals using fractional fourier domain filtering. *IEEE Transactions on Biomedical Engineering,* 56(4), 996-1004.

Jensen, R.L., and Ebben, W.P. (2007). Quantifying plyometric intensity via rate of force development, knee joint and ground reaction forces. *Journal of Strength and Conditioning Research,* 21(3), 763-767.

Kavanagh, J.J., and Hylton, M.B. (2008). Accelerometry: A technique for quantifying movement patterns during walking. *Gait and Posture,* 28, 1-15.

Lafortune, M.A., Hennig, E.M., and Valiant, G.A. (1995). Tibial shock measured with bone and skin mounted transducers. *Journal of Biomechanics,* 28, 989-993.

Madigan, M.L., and Pidcoe, P.E. (2003). Changes in landing biomechanics during a fatiguing landing activity. *Journal of Electromyography and Kinesiology,* 13, 491-498.

Moran, K.A., and Marshall, B.M. (2006). Effect of fatigure on tibial impact accelerations and knee kinematics in drop jumps. *Medicine and Science in Sports and Exercise,* 38(10), 1836-1842.

Murphy, D.F., Connolly, D.A., and Beynnin, D.D. (2003). Risk factors for lower extremity injury: a review of the literature. *British Journal of Sports Medicine*, 37, 13-29.

Self, B.P., and Paine, D. (2001). Ankle biomechanics during four landing techniques. *Medicine and Science in Sports and Exercise,* 33(8), 1338-1344.

Winter, D.A. (1990). Biomechanics and Motor Control of Human Movement. John Wiley & Sons, Inc., New York.

Zhang, S.N., Derrick, T.R., Evans, W., and Yu, Y.J. (2008). Shock and impact reduction in moderate and strenuous landing activities. *Sport Biomechanics,* 7(2), 296-309.

Acknowledgement:
Study expenses were funded by a Northern Michigan University Excellence in Education grant. We thank Dr. Apostolos Georgakis and Suba Raman Subramaniam for the MATLAB filtering routine. Furthermore, we thank Timothy Suchomel and McKenzie Fauth for assistance with data acquisition.