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Evaluation of an accelerometer to assess knee mechanics during a drop landing

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ABSTRACT

Non-contact anterior cruciate ligament (ACL) injuries account for 70% of all ACL injuries, and can lead to missed time from activity for athletes and a predisposition for knee osteoarthritis. Prior research has shown that athletes who land in a stiff manner, with larger internal knee adduction and extension moments, are at greater risk for an ACL injury. A three-dimensional accelerometer placed at the tibial tuberosity may prove to be a low-cost means of assessing these risk factors. The primary purpose of this study was to compare tibial accelerations during drop landings with kinematic and kinetic risk factors for ACL injury measured with three-dimensional motion capture. The secondary purpose of this study was to compare these measures between soft and stiff landings. Participants were instructed to land bilaterally in preferred, soft, and stiff manners. Peak knee flexion decreased significantly from soft to stiff landings. Peak internal knee extension moment, peak anterior/posterior knee acceleration, and peak medial knee acceleration all increased significantly from soft to stiff landings. No associations were found between landing condition and either frontal plane knee angle at maximum vertical ground reaction force or peak internal knee adduction moment. Significant positive associations between kinetics and accelerations were found only in the sagittal plane. As such, while a three-dimensional accelerometer could discern between soft and stiff landings in both planes, it may be better suited to predict kinetic risk factors in the sagittal plane.

1. Introduction

Non-contact anterior cruciate ligament (ACL) injuries account for around 70% of all ACL injuries annually and are a common reason for missed time from activity for athletes (Agel et al., 2005; Griffin et al., 2000). Women have been shown to be two to eight times more likely than men to experience a non-contact ACL injury, although screening for ACL injuries is not gender specific (Agel et al., 2005; Padua et al., 2015). Regardless of gender, ACL injuries can predispose athletes to long-term problems such as osteoarthritis (Ajuied et al., 2014; Gillquist and Messner, 1999). Prior studies have examined factors that contribute to the risk for ACL injuries and have suggested that there are several categories of risk factors. These categories include neuromechanical, anatomical, genetic, and hormonal factors (Shultz et al., 2015). Regarding neuromechanical risk factors, it has been suggested that frontal plane angles and moments during landing may contribute to injury risk (Hewett et al., 2005; Ireland, 1999; Myer et al., 2015; Oh et al., 2012; Withrow et al., 2006). Prior literature has found greater knee abduction angles and internal knee adduction moments in athletes who had an ACL injury (Hewett et al., 2005). Normally, dangerous values for these risk factors occur during a rapid deceleration while landing, jumping, or performing a sideways cut, and typically during the early portion of ground contact (Kiapour et al., 2014). In addition to frontal plane risk factors for developing an ACL injury, decreased knee flexion angles, increased posterior ground reaction force, and increased internal knee extension moments have been established as sagittal plane knee mechanics that are likely lead to increased ACL strain (Dai et al., 2015; Taylor et al., 2011; Utturkar et al., 2013; Yu et al., 2006). Furthermore, it has been shown in female athletes that increases in both knee abduction angle and internal knee adduction moment occur with decreases in knee flexion angle during bilateral landings (Pollard et al., 2009). It has been suggested that this may be due to a reliance on ligaments in the knee to slow down movement upon landing rather than controlling the deceleration of the body via sagittal plane movement (Pollard et al., 2009).

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Some screening protocols have been established for identifying risk factors such that they can be later targeted for modification in order to prevent injury (Padua et al., 2015). Risk factors that have been examined in great detail with regards to these protocols include increased knee abduction and internal knee rotation angles, and decreased knee flexion angle. Furthermore, studies have built upon these screening measures to develop methods for providing real-time biofeedback for athletes to decrease these risk factors (Dowling et al., 2012b; Ericksen et al., 2016; Ford et al., 2015). Many of these biofeedback protocols involve the use of a motion analysis lab to provide subjects with kinematic or kinetic data either during or after a task. The challenge in widespread use of such protocols is the availability of the technology, and wearable devices may allow for greater opportunity for the application of these protocols more broadly.

A three-dimensional accelerometer has potential as a screening and biofeedback device given their cost and ease of application to the body. A single 3-D accelerometer placed on the proximal tibia may be able to detect risky sagittal and frontal plane mechanics at the knee. There have been efforts to use inertial measurement units (IMUs) for this purpose. Dowling and colleagues utilized a system of multiple IMU sensors to examine thigh frontal plane angular velocity and knee flexion angle as part of a biofeedback protocol (Dowling et al., 2012b). They demonstrated success in assessing frontal plane mechanics. However, their protocol was based on using multiple IMUs, and these devices can be costlier than a simple accelerometer. More importantly, the system used previously appeared more cumbersome for the user, thereby limiting applicability outside a laboratory setting. As such, there remains a need for an inexpensive and user-friendly form of field-based biofeedback to decrease ACL injury risk. Wearable accelerometers have been used previously to screen and act as an intervention for injury risk factors in runners, and this same paradigm could be applied to the risk for ACL injuries (Crowell et al., 2010).

In evaluating the viability for using an accelerometer to assess injury risk, there are three issues to consider; (1) Does acceleration data relate to commonly accepted mechanical markers of injury risk, (2) Can an accelerometer detect changes in an individual's landing mechanics, and (3) Can such data differentiate landing mechanics between individuals? As such, the purpose of this study was to examine tibial accelerations collected via an inexpensive wearable accelerometer in the frontal and sagittal planes during drop-landing maneuvers of different stiffnesses and to compare these measures to knee joint kinematics and kinetics that are risk factors for developing an ACL injury that are collected via 3-D motion capture. Altering landing stiffness was chosen as an intervention because it is easily implemented and also directly relates to common knee injury prevention training approaches (Padua et al., 2015). It was hypothesized that lower extremity tibial accelerations in the sagittal and frontal planes will demonstrate a positive relationship with internal knee extension moment and internal knee adduction moment, respectively. Furthermore, it was hypothesized that tibial accelerations and the aforementioned kinematic and kinetic parameters will significantly differ across landing stiffness conditions.

2. Methods

Subjects: Eighteen subjects (6 males and 12 females, 23.7 ± 3.9 years, 1.68 ± 0.11 m, 65.9 ± 12.9 kg) were recruited from the general student population of the University of Wisconsin-Milwaukee to participate in this study. Written informed consent was obtained from all subjects prior to participation in accordance with the Institutional Review Board of the University of Wisconsin-Milwaukee. Subjects completed a screening questionnaire prior to participation to (1) determine leg dominance, which was assessed by asking which leg one would kick a soccer ball with, (2) ensure that they did not have any medical conditions and were not taking medications that might have impaired the ability to jump and land, and (3) ensure that they did not currently have lower extremity pain or injury, or previously have surgery to the lower extremity that may have changed their landing pattern from normal.

Instrumentation: Three-dimensional kinematic data were collected for the right leg using a Motion Analysis Eagle motion capture system (Motion Analysis, Inc., Santa Rosa, CA) at 200 Hz. For tracking purposes, reflective markers were placed on the right and left anterior and posterior superior iliac spines and four-marker rigid clusters were placed at the thigh, shank, and heel of the dominant leg. Additional markers used for a standing calibration trial were placed on the right and left iliac crests and greater trochanters, on the lateral and medial femoral epicondyles and malleoli, and on the first and fifth metatarsal heads. All markers were placed by one researcher for all subjects. Kinetic data were collected with a Bertec force plate (Bertec, Inc., Columbus, OH) at 1000 Hz. Accelerations at the tibial tuberosity of the dominant leg were collected using a Noraxon DTS 3D Accelerometer (Noraxon, Scottsdale, AZ) at 1000 Hz. The accelerometer was placed on the tibial tuberosity by one researcher for all subjects. A Velcro strap was wrapped around the shank at the level of the tibial tuberosity to ensure that the accelerometer would remain stable underneath. Positive accelerations indicated acceleration of the shank anteriorly and medially within the shank reference frame (Fig. 1). Standard Saucony Jazz shoes were provided for this study (Saucony, Lexington, MA).

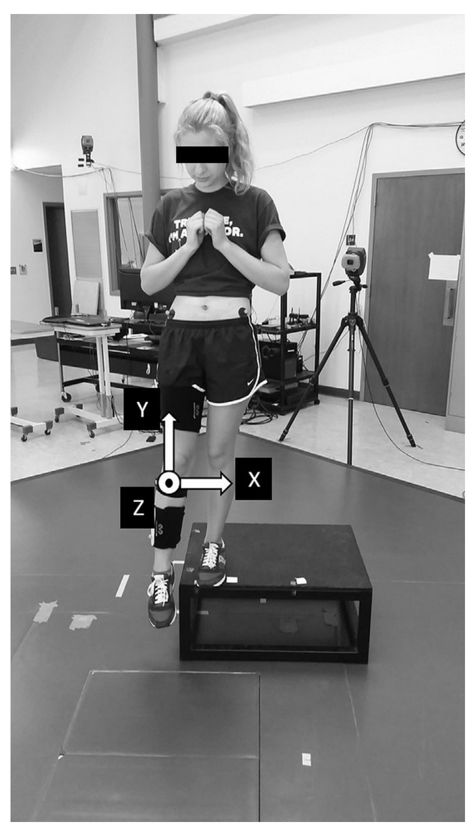

Fig. 1. Bilateral drop landing task initial position. Tri-axial accelerometer was located at the tibial tuberosity and stabilized with Velcro strap. Arrows indicate positive acceleration values for each of the three axes.

Data Collection: After collecting data for the standing calibration trial, subjects were instructed to stand on a 30-cm tall box located next to the force plate. Subjects were instructed to perform a bilateral drop landing task in which the individual stepped off the box and landed on the ground. Subjects were instructed to land with the foot of the dominant leg on the force plate, with the opposite foot off the force plate. Trials were rejected if the subject did not land completely on the force plate or if the subject did not land with two feet. Three landing conditions were implemented with ten drop landing trials collected for each condition. Subjects were instructed to land in preferred, soft, and stiff manners. Preferred landings were described as occurring in a manner that felt normal to the subject. Soft landings were described as occurring with knees flexed. Stiff landings were described as occurring with knees extended. Subjects performed at least one practice trial for each condition prior to collecting drop landing trials. Preferred landing trials were performed first, followed by soft landing trials, and lastly stiff landing trials. The time between each trial was approximately thirty seconds, and subjects were allowed two to three minutes of rest between each set of ten trials.

Data Analysis: Three-dimensional kinematic and kinetic data were processed using Visual3D (v6.00.15, C-Motion, Inc., Rockville, MD). Kinematic motion capture data were low-pass filtered at 12 Hz, and kinetic data were low-pass filtered at 50 Hz. Accelerometer data were not filtered. The hip joint center was calculated as twenty-five percent of the linear distance between the greater trochanter markers (Weinhandl and O'Connor, 2010). The knee joint center was determined as fifty percent of the linear distance between the lateral and medial femoral epicondyles. The ankle joint center was determined as fifty percent of the linear distance between the lateral and medial malleoli. Joint moments were calculated via an inverse dynamics approach and were normalized to body mass (Bresler and Frankel, 1950). Knee angle at initial contact in the frontal and sagittal planes, knee angle in the frontal plane at the time of maximum vertical ground reaction force, peak internal knee extension moment, peak internal knee adduction moment, and maximum knee flexion angle across the three conditions were all reported. Knee flexion angle, knee abduction angle, internal knee extension moment and internal knee adduction moment were all denoted as positive. The dependent variable of peak internal knee extension moment was determined to be the first peak after initial contact. Peak tibial accelerations following initial contact anteriorly, posteriorly, and medially, were also reported across all three conditions.

Statistical Analysis: A one-way ANOVA followed by a Tukey post-hoc analysis was used to compare kinematics and kinetics between conditions. Pearson product-moment correlation coefficients were computed within each subject from ten preferred landing trials to assess the linear relationship between peak internal knee extension moment and peak accelerations of the knee in the sagittal plane and peak internal knee adduction moment and peak medial acceleration in the frontal plane during the preferred condition. Group mean correlation coefficients were computed to assess the relationship of peak anterior and posterior accelerations with peak internal knee extension moments and peak medial accelerations with peak internal knee adduction moments within subjects and across conditions. A significance level of ($p < 0.05$) was applied for all statistical analyses. All statistical analyses were performed using SPSS (v19.0 SPSS Inc., Chicago, IL).

3. Results

Average values for knee flexion angle at initial contact, maximum knee flexion angle, peak internal knee extension moment, and peak anterior and peak posterior accelerations of the knee within each of the three landing conditions are presented in Table 1. Both knee flexion angle at initial contact and maximum knee flexion angle significantly decreased from the soft landing to the stiff landing condition (Table 1). Peak internal knee extension moment, peak anterior knee acceleration and peak posterior knee acceleration significantly increased from the soft landing to the stiff landing condition (Table 1). Peak internal knee extension moment, peak anterior acceleration and peak posterior acceleration all occurred within the first 30 ms following initial contact (Fig. 2). Positive associations were found between peak internal knee extension moment and both peak anterior knee acceleration (mean $R^2 = 0.67 \pm 0.13$) and peak posterior knee acceleration (mean $R^2 = 0.78 \pm 0.05$) when compared across conditions within subjects, averaged across all subjects (Fig. 3). Peak positive acceleration also displayed a significant correlation with peak internal knee extension moment during preferred landings across subjects ($R^2 = 0.524$).

Frontal plane knee angle at initial contact, frontal plane knee angle at the time of maximum vertical ground reaction force, peak internal adduction moment, and peak medial knee acceleration average values within each of the three landing conditions are displayed in Table 2. Knee abduction angle at initial contact significantly increased from the soft landing to the stiff landing condition (Table 2). Peak medial knee accelerations significantly decreased from the stiff landing to both the preferred and soft landing conditions (Table 2). No significant differences were displayed between landing conditions for peak internal adduction moment or frontal plane knee angle at the time of maximum vertical ground reaction force. Peak internal knee adduction moments occurred within the first 20 ms after initial contact and were directly followed by medially directed knee accelerations (Fig. 4). No association was found between peak medial acceleration and internal knee adduction moment (mean $R^2 = 0.017 \pm 0.76$) when compared across conditions within subjects, averaged across all subjects (Fig. 5). Furthermore, no significant correlation was found between peak medial acceleration and peak internal knee adduction moment during preferred landings ($R^2 = 0.001$).

4. Discussion

The first main objective of the present study was to compare sagittal and frontal plane tibial accelerations captured via an accelerometer to kinematic and kinetic risk factors for ACL injury commonly captured via a three-dimensional motion capture system. Risk factors included knee flexion and abduction angles, and knee internal extension and internal adduction moments (Dai et al., 2015; Hewett et al., 2005; Myer et al., 2015; Oh et al., 2012; Yu et al., 2006). The findings show a moderately positive correlation between peak internal knee extension moment and peak anterior acceleration, and a strong positive correlation between peak internal knee extension moment and peak posterior acceler-

Table 1
Group mean (±SD) sagittal plane knee kinematics and kinetics for each landing condition.

	Soft	Preferred	Stiff
Flexion at initial contact (°)*	22.2 ± 6.8[a]	14.5 ± 5.8[b]	12.2 ± 5.2[b]
Maximum flexion (°)*	94.5 ± 16.2[a]	70.8 ± 17.8[b]	56.1 ± 6.9[c]
Peak internal extension moment (Nm/kg)*	1.4 ± 0.4[a]	1.7 ± 0.6[a,b]	2.0 ± 0.5[b]
Peak anterior acceleration (m/s^2)*	27.1 ± 18.1[a]	54.4 ± 26.8[b]	80.9 ± 40.5[c]
Peak posterior acceleration (m/s^2)*	38.3 ± 11.3[a]	56.5 ± 14.8[b]	68.1 ± 12.7[c]

* Indicates significant main effect for condition ($p < 0.05$); Letters (a, b and c) that are alike indicate no significant difference betweeen conditions ($p < 0.05$).

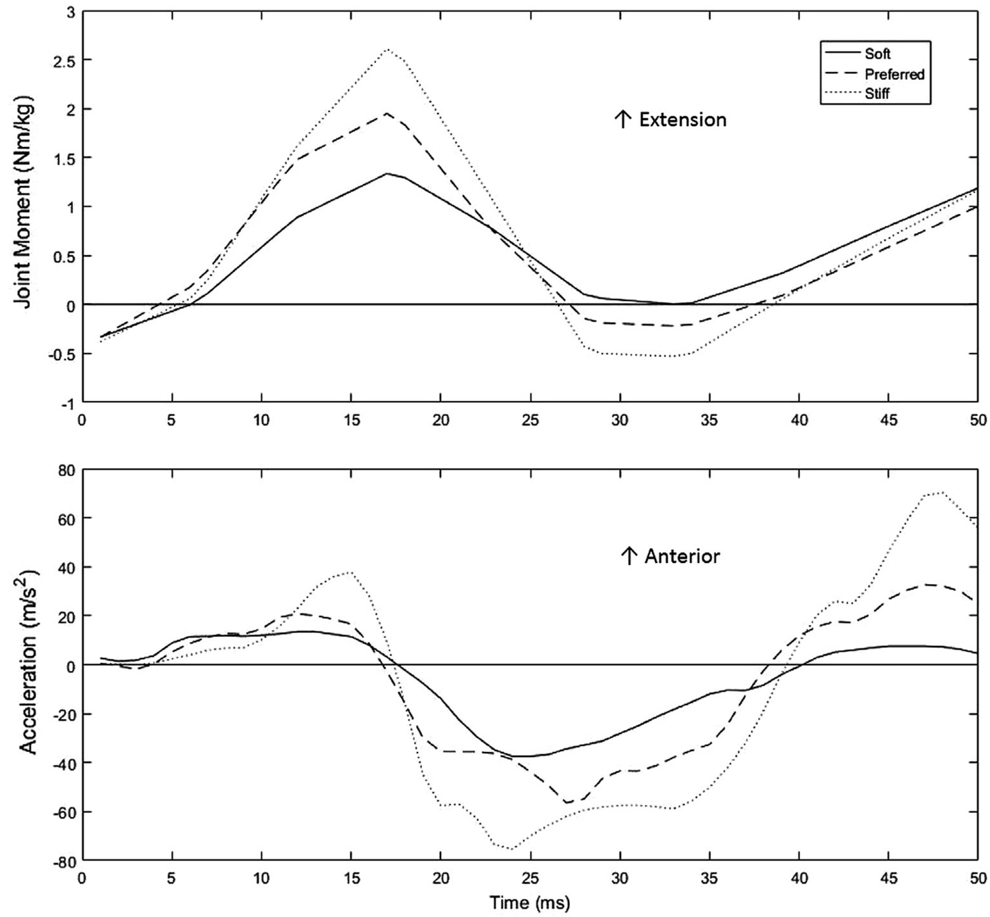

Fig. 2. Exemplar trials displaying knee acceleration and joint moment in the sagittal plane in one subject for 50 ms following initial contact. Peak anterior and posterior knee accelerations and knee extension moment occur within the first 30 ms and increase with stiff landing patterns.

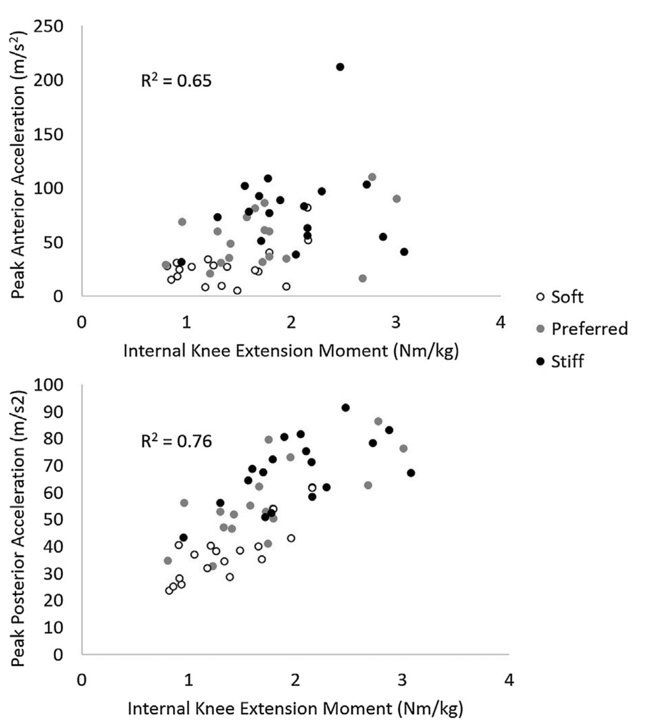

Fig. 3. Peak internal knee extension moment displayed significant associations with both peak anterior and peak posterior knee accelerations when compared across conditions and averaged across all subjects.

Table 2
Group mean (±SD) frontal plane knee kinematics and kinetics for each landing condition.

	Soft	Preferred	Stiff
Frontal plane knee angle at initial contact (°)*	−2.0 ± 2.7[a]	0.01 ± 2.8[a,b]	0.4 ± 1.5[b]
Frontal plane knee angle at maximum vertical ground reaction force (°)	1.0 ± 3.6	0.1 ± 3.1	−0.7 ± 1.9
Peak internal adduction moment (Nm/kg)	0.2 ± 0.1	0.3 ± 0.2	0.3 ± 0.3
Peak medial acceleration (m/s^2)*	22.4 ± 10.7[a]	37.7 ± 16.1[a]	63.1 ± 35.7[b]

* Indicates significant main effect for condition ($p < 0.05$); Letters (a and b) that are alike indicate no significant difference between conditions ($p < 0.05$).

ation, when looking across conditions and within subjects. Furthermore, there were no associations between peak internal knee adduction moment and peak acceleration in the medial direction when examining data across conditions and within subjects, or for preferred landings alone. These results suggest that an accelerometer can potentially predict knee joint moments in the sagittal plane during bilateral landing. Additionally, accelerometers may be better suited to predict knee joint moments in the sagittal plane as compared to the frontal plane.

A further purpose of this study was to compare knee accelerations and known risk factors for ACL injury in both the sagittal and frontal planes across soft, preferred, and stiff landings. In the sagittal plane, as expected, knee flexion decreased with an increased landing stiffness. Both maximal knee flexion and knee

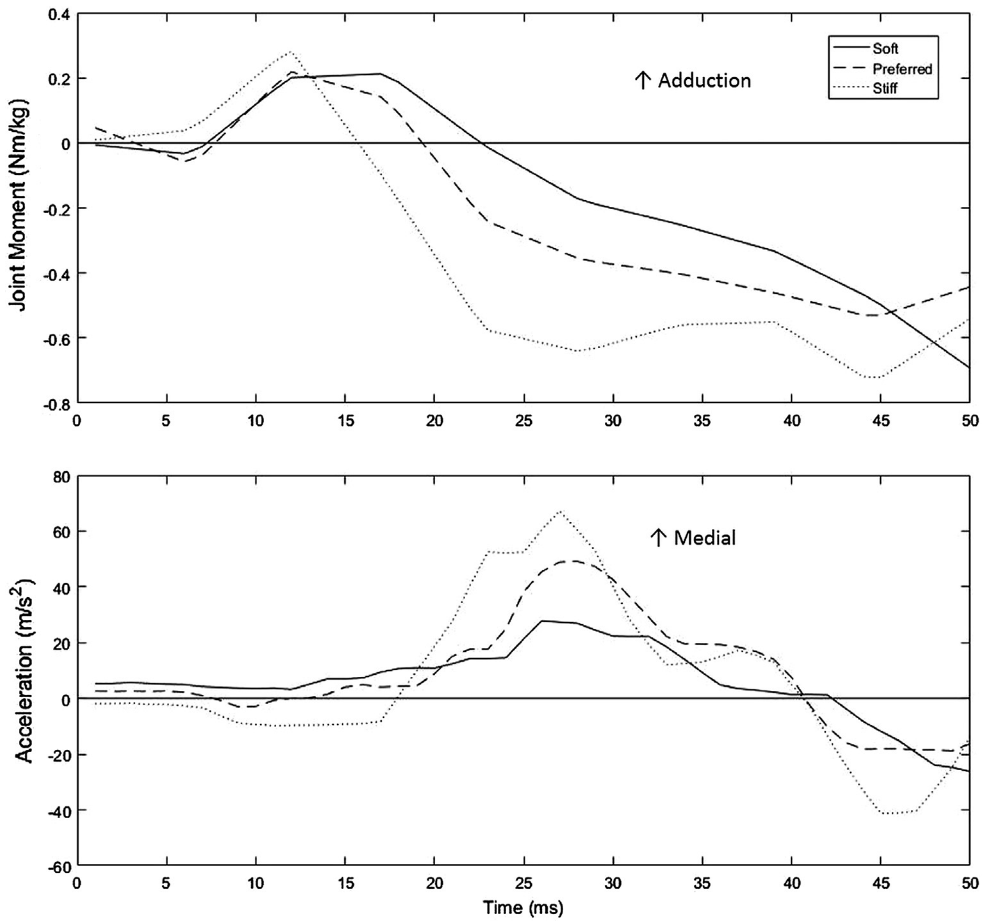

Fig. 4. Exemplar trials displaying knee acceleration and joint moment in the frontal plane in one subject for 50 ms following initial contact. Peak knee adduction moments occurred within the first 20 ms after initial contact and were followed directly by medial knee accelerations. Only medially directed knee accelerations increased with stiff landing patterns.

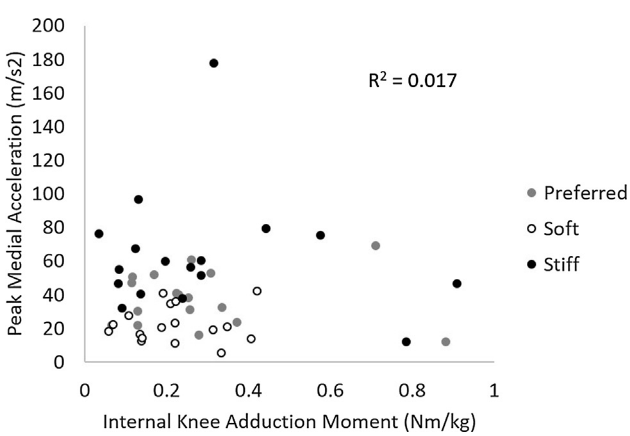

Fig. 5. Peak internal knee adduction moment did not display a significant association with peak medial knee acceleration when compared across conditions and averaged across all subjects.

flexion at initial contact displayed decreasing values from soft to stiff landing conditions. This indicates that subjects were successful in following instructions for performing both soft and stiff landings. However, because knee flexion at initial contact was not significantly different between preferred and stiff landing conditions, this may suggest that the subjects in the present study typically land in a position resembling a stiff landing initially. Peak internal knee extension moments were significantly greater during stiff landing compared to soft landings. Significant differences in knee flexion angle and internal knee extension moment across soft and stiff conditions agree with previous work comparing the same variables across soft and stiff drop landings using biplane fluoroscopy (Myers et al., 2011). In addition, peak accelerations in both the anterior and posterior directions and peak knee internal extension moments increased with an increase in landing stiffness. This indicates that there is a direct relationship between sagittal plane peak accelerations and peak extension moments at the knee for varying levels of landing stiffness.

While changes in sagittal plane kinematics with landing condition agreed with previous studies, frontal plane kinematics differed in this respect. In the frontal plane, knee angle at initial contact did change significantly from 2 degrees of adduction during the soft condition to neutral during the stiff condition. Furthermore, frontal plane knee angle at the time of maximum vertical ground reaction force did not significantly differ for any condition. However, these frontal plane knee angle magnitudes do not agree with the previous work of Pollard and colleagues for bilateral landings as they displayed larger peak abduction angles for both soft and stiff landing conditions (Pollard et al., 2009). Pollard and colleagues designated soft and stiff landing groups based on their subjects' preferred landing pattern, however, which may partially explain this disagreement (Pollard et al., 2009). The preferred condition in the present study elicited a neutral knee angle at both initial contact and maximum vertical ground reaction force and did not significantly differ from either the soft or stiff conditions. As such the subjects from the present study may not naturally use the ligaments to control movement and decelerate the body upon landing as was suggested by Pollard and colleagues (Pollard et al., 2009).

The frontal plane knee angles that were close to neutral may, in turn, explain the lack of a significant difference in peak internal knee adduction moments between landing conditions. A neutral frontal plane knee angle may have decreased the peak moment compared to what has typically been observed in prior studies for ACL injured knee kinetics. However, the lack of a significant difference between landing conditions in healthy individuals for peak internal knee adduction moment has been shown previously in the literature (Myers et al., 2011). It is possible that a stiff bilateral drop landing in a laboratory setting may not stress healthy individuals to the point that we begin to observe kinetics similar to those found for athletes who have incurred an ACL injury (Myer et al., 2015). However, bilateral drop landing is a task that has been used in ACL injury prevention studies previously and can be used in future studies to examine the effectiveness of changing landing patterns with biofeedback to reduce the risk for ACL injury (Padua et al., 2015).

While peak internal adduction moments were not significantly different across landing conditions, peak medial knee accelerations were significantly larger during stiff landings as compared to soft landings. As such, while these linear accelerations could not predict any of the frontal plane kinematics or kinetics discussed previously, these accelerations could reasonably differentiate between landing conditions across subjects. Previous work examining the use of an IMU during bilateral drop jump maneuvers has shown weak to moderate correlations between peak internal adduction moment and both thigh and shank angular velocities (Dowling et al., 2012a). However, the present study is likely one of the first studies to examine linear acceleration in the frontal plane in relation to kinematic and kinetic variables related to the knee. While it is possible that segment angular velocity may be a better predictor of peak internal adduction moment than linear acceleration, the difference in task between the two studies may explain the lack of a relationship between knee acceleration and moment in the present study. As such, incorporating the drop jump maneuver, particularly with different landing conditions, may be better at eliciting kinetics similar to those found in individuals at greater risk for ACL injury. In turn, we might find a stronger relationship between peak medial acceleration of the knee and peak internal adduction moment.

Given that the purpose of this study was focused on the applicability of an accelerometer to assess knee mechanics, both males and females were recruited without regard to gender distribution. However, in theory it is possible that the nature of the accelerometer data may differ between genders, as females have been shown to land more stiffly than males (Weinhandl et al., 2010). Therefore, a post-hoc analysis was performed comparing the 12 females to the six males. Given the relatively few males, effect sizes were examined for potential differences between genders. All effect sizes were less than 0.3. Therefore, there were no apparent differences between genders for any of the sagittal or frontal plane variables.

While the kinematics and kinetics of the knee in the sagittal and frontal planes have been widely focused on with regards to ACL injury, accelerations of the knee in either plane have not been examined to the same extent. The present study is novel in that it examines these knee accelerations and displays the effectiveness of an accelerometer in detecting both within- and between-subject variation in sagittal plane moments for different landing conditions. Furthermore, this study shows that frontal plane accelerations can be used to distinguish between different types of landing conditions across subjects. The simplicity of using a single accelerometer placed just below the knee to predict specific risk factors for ACL injury is important when examining more recent literature regarding injury prevention programs. While previous studies have shown the effectiveness of scoring systems via video capture or feedback via three-dimensional motion capture in reducing the risk for ACL injury (Ericksen et al., 2016; Padua et al., 2015), these programs either take time or are more cumbersome and relegated to the laboratory. The previous work performed using IMU's to provide biofeedback, more easily implemented, utilized visual feedback in a laboratory setting (Dowling et al., 2012b). As such, the incorporation of a simple accelerometer measuring only linear accelerations could be used as a screening tool to assess increased accelerations in the sagittal plane. Additionally, the accelerometer could be used to provide real-time biofeedback for individuals at risk for an ACL injury to increase knee flexion during landings and decrease internal knee extension moment. This could be accomplished with a training protocol that could be used outside of a laboratory setting and could be effective in reducing the risk for ACL injury.

It was hypothesized for this study that sagittal and frontal plane knee accelerations would be significant predictors of known kinematic and kinetic risk factors for ACL injury when examined across different landing conditions. Significant differences across landing conditions were found for sagittal plane kinematics and kinetics. However, in the frontal plane, only initial contact angle differed between conditions. Furthermore, significant differences were found for peak accelerations in the sagittal and frontal planes between conditions. However, only peak accelerations in the sagittal plane displayed significant associations with peak knee moments. This suggests that while accelerometers can differentiate between landing conditions in the sagittal and frontal planes, linear accelerations can only predict sagittal plane kinetics for the drop landing maneuver.

Conflict of interest statement

There are no financial and personal relationships with other people or organizations that could inappropriately influence (bias) this work.

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