

## **Title**

Whole body kinematics and knee moments that occur during an overhead catch and landing task in sport

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## **Abstract**

*Background:* Athletes suffering an anterior cruciate ligament injury tend to exhibit similar body postures that in sidestep cutting are associated with increased knee moments. This relationship, however, has not been investigated in landing. Catching a ball in different overhead positions may affect landing postures and knee joint moments. This study investigated these possible relationships. It was anticipated that some joint postures would be associated increased knee loads during the landing task.

*Methods:* Twenty-five healthy male team sports athletes performed four variations of a landing task. Full body kinematics were identified at initial contact. Peak flexion, valgus and internal rotation moments at the knee, measured during early landing, were normalized to mass and height and statistically compared. Intra-participant correlations were performed between all kinematics and each moment. Mean slopes for each correlation were used to identify the existence of relationships between full body kinematics and knee joint moments.

*Findings:* Landing after an overhead catch when the ball moved towards a player's support leg resulted in increased peak valgus moments. These increased valgus moments were correlated with increased knee flexion, hip flexion, and torso lean, as well as torso rotation towards the support leg, and foot and knee external rotation. Increased internal rotation moments were correlated with reduced hip abduction and external rotation, increased ankle inversion, knee external rotation and torso lean away from the support leg.

*Interpretation:* Learning to land with techniques that do not reflect postures associated with high knee moments may reduce an athlete's risk of non-contact anterior cruciate ligament injury.

## **Key Words**

Anterior Cruciate Ligament; Injury; Injury Prevention; Biomechanics

## Introduction

The two primary sporting maneuvers observed during non-contact anterior cruciate ligament (ACL) injuries are sidestep cutting and landing following a jump (Cochrane et al., 2007). As such, there has been extensive research attempting to better understand what characteristics of these tasks are associated with non-contact ACL injuries. Even though there are many factors implicated in ACL injury, ultimately, when the ACL loading become higher than its strength, ligament damage occurs. Therefor to best prevent non-contact ACL injuries it is important to understand the mechanism behind high ACL loading. Due to this, much of the previous research has been directed towards understanding the cause of and support for external knee moments that results in high ACL loads in sporting tasks (Besier et al., 2001, McLean et al., 2007, Besier et al., 2003a). Moments are used as it has been shown that high valgus and internal rotation moments at the knee coupled with anterior draw, caused by quadriceps extension, can highly load the ACL in particularly in extended knee postures (Markolf et al., 1995, Fleming et al., 2001). So the question arises; what particular facets of sidestep cutting and landing result in critical high knee loading?

Specific sidestepping and landing techniques have been associated with non-contact ACL injuries. These relationships have been derived from both the visual analysis of videos of actual injuries (Hewett et al., 2009, Olsen et al., 2004, Cochrane et al., 2007), and in laboratory studies that have linked specific sidestep cutting techniques to increased knee load (Dempsey et al., 2007, McLean et al., 2005). Specifically, it appears that body postures that have an extended and internally rotated lower limb, with an abducted hip, are associated with increased peak valgus moments (Dempsey et al., 2007, McLean et al., 2005). This would suggest that these postures have an increased risk of ACL injury. From an upper body perspective, increased torso rotation and lateral flexion away from the stance leg have been linked to higher knee internal rotation and valgus moments respectively (Dempsey et al.,

2007). Although there have been numerous studies investigating lower limb kinematics during landing (Lephart et al., 2002, Decker et al., 2003, Onate et al., 2003, Hewett et al., 2005), there is yet to be a study investigating the relationship between full body kinematics and knee moments during landing tasks.

An understanding of how full body kinematics affects knee moments is important, particularly in team sports that involve carrying or catching a ball. Indeed, it has been shown that team sport athletes that suffer a non-contact ACL injury have had some interaction with the ball, often performing a task in rapid response to some game situation (Olsen et al., 2004). This ball interaction probably affects the full body kinematics and knee moments. For example, while performing a sidestep cut the act of carrying a ball on different sides of the body has been shown to modify both technique and knee valgus loading (Chaudhari et al., 2005). Furthermore, Cowling and Steele (2001) found that requiring an athlete to catch a ball during flight altered hip and trunk sagittal plane kinematics during single leg landing. Therefore, a landing task that has anticipatory and ball handling components may better reflect the game scenarios related to ACL injury and high knee loading. Such a task may also provide sufficient kinematic variation to enable the investigation of the relationship between whole body kinematics and knee joint loading.

The first aim of this study was to investigate how knee moments generated during landing were affected by variations in an overhead ball catching and landing tasks that occurs in Australian Football. It was hypothesized that different ball movement directions would result in changed knee joint moments. The second aim was to identify joint postures associated with increased knee valgus and internal rotation moments, as these moments have been associated with high ACL loading. It was anticipated that joint postures associated with increased load would be similar to those observed during actual non-contact ACL injury.

## Methods

### Participants

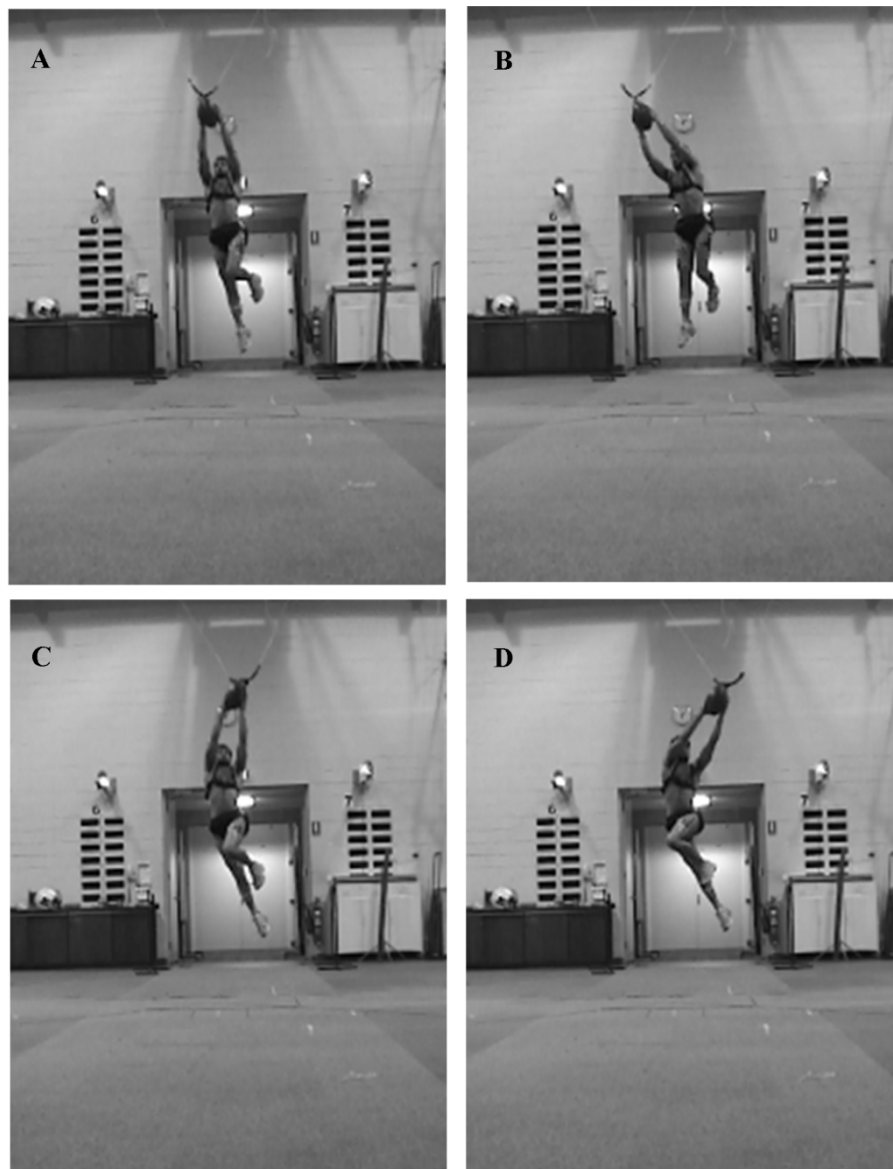
Twenty five healthy male team sports athletes were recruited to participate in this study (height: mean 181.8 (SD 7.1) cm, mass: mean 78.0 (SD 12.1) kg). All participants were experienced in performing landing tasks through their respective team sport. Participants were excluded if they had a history of major lower limb injury. Ethics approval was obtained from The University of Western Australia (UWA) Human Research Ethics committee and written informed consent was obtained from all participants prior to data collection. Subject numbers were based upon a power analysis performed using effect sizes reported by Dempsey et al. (2007) and a power of 0.8 and  $p = 0.05$ .

### Experimental Design

All testing was undertaken in the UWA Sports Biomechanics Laboratory with the movement of retro-reflective markers affixed to the participants' segments recorded using a 12 camera VICON MX motion analysis system sampling at 250 Hz (VICON, Oxford, UK). Ground reaction forces from a 1.2 m x 1.2 m force plate (Advanced Mechanical Technology Inc., Watertown, USA) were synchronously recorded with the motion analysis data at a data acquisition rate of 2000 Hz. Before commencing the trials participants selected their preferred support leg; i.e. the leg used for both takeoff and landing.

A ball movement rig was constructed, which was attached to a gantry above the force plate (Supplementary Material). This rig allowed the same trained experimenter to release a ball that then fell, under gravity, either towards or away from the participant's support leg. The initial height of the ball was set to that reached from each participant's maximal vertical jump off their preferred support leg. During the testing session participants performed three trials of four landing tasks (Figure 1): 1) ball moving toward the support leg early in approach

(TE), 2) ball moving toward the support leg late in approach (TL), 3) ball moving away from the support leg early in approach (AE), and 4) ball moving away from the support leg late in approach (AL). During TE and AE trials the ball approached a peak lateral movement of 0.6 m, whereas the TL and AL trials displayed reduced lateral movement. At the commencement of each trial the participants were unaware of which direction the ball would fall, and all trial tasks were carried in random order.



**Figure 1** Ball catch position in the landing task. A – Towards Late (TL) ; B – Towards Early (TE); C – Away Late (AL); and D – Away Early (AE).

Participants performed as many tasks as necessary (15.6 (SD 2.5) tasks) to record three successful trials of each of landing task. To counter for the possible effect of fatigue on the results, participants were given adequate rest between repetitions. During this time they walked back to the start marker and the landing rig was reset. On average participants performed one trial every 1.5 to 2 minutes, limiting the accumulated fatigue. The tasks were presented in a random order, accounting for any effect of any fatigue when the tasks are averaged across the individual. We have successfully used this approach previously in our sidestepping analysis (Besier et al., 2001, Cochrane et al., 2010, Dempsey et al., 2009, Stoffel et al., 2010). A successful trial involved participants taking off and successfully taking possession of the dropping ball and landing on their preferred foot on the force platform. No restrictions were placed on landing technique following initial foot contact with the platform.

### **Data Collection and Analysis**

Participants were fitted with retro-reflective markers as per the UWA Full Body Model (Dempsey et al., 2007), a combination of the UWA Upper (Reid et al., 2010) and Lower Body Models (Besier et al., 2003b). This consisted of 50 markers placed on either bony landmarks or as part of three-marker clusters. Single markers were placed on the left and right forehead, left and right rear head, left and right acromion process, the sternal notch, spinous process of C7 and T10, the xiphoid process, left and right anterior superior iliac spines, left and right posterior superior iliac spines, left and right head of the first and fifth metatarsal, left and right head of the third metacarpal, and the left and right calcaneus.

Three-marker clusters were placed on the upper arm, forearm, thigh and leg and a two-marker cluster on the dorsal surface of the hand. In addition, the ankle, wrist and shoulder joint centers were respectively defined using markers on the left and right medial and lateral malleoli, left and right radial and ulnar styloid processes and left and right anterior and posterior shoulder. These markers were removed during the dynamic trials. A six-marker

pointer was used to identify 3D location of the medial and lateral humeral epicondyles of both elbows, and medial and lateral femoral epicondyles of both legs (Besier et al., 2003b). Functional knee and hip tasks were carried out to identify the knee joint flexion/extension axis and hip joint center, respectively (Besier et al., 2003b). Each subject also stood on a foot calibration rig, where foot abduction/adduction and rear foot inversion/eversion angles were measured, with the resulting data used to establish the foot alignment and coordinate system (Besier et al., 2003b). These procedures are described in detail in Besier et al. (2003b).

Subsequently, kinematic and inverse dynamic calculations were performed in VICON Workstation (VICON, Oxford, UK), using the UWA Model, which employs custom code written in MATLAB (Mathworks, Natick, USA) and VICON BodyBuilder (VICON, Oxford, UK). Prior to modeling, both the ground reaction force and position data were filtered using a 4<sup>th</sup> order 18 Hz zero-lag low-pass Butterworth filter, with the filter frequency selected from residual analysis and visual inspection of the data. Inverse dynamics were used to calculate external joint moments, using the body segment parameters reported by de Leva (1996).

Our modeling procedures have been shown to have high repeatability. This is for both lower limb kinematics and kinetics (Besier et al., 2003b) and upper body kinematics (Reid et al., 2010). For example, we reported test-retest coefficients of multiple determination greater than 0.7 for varus/valgus moments and internal/external rotation moments (Besier et al., 2003b). The UWA model has also been shown to be able to be accurate to within 1° in controlled dynamic tasks (Elliott et al., 2007).

A landing phase was identified based upon the vertical ground reaction force curve. Previously *in vivo* studies have shown that the peak ACL load occurs close to the peak vertical force (Cerulli et al., 2003). Therefore, similar to the approach used by Decker and colleagues (2003), we defined the landing phase from initial foot contact to a time point that



was double the time from initial foot contact to the peak vertical ground reaction force. This process was performed using a custom MATLAB program that identified the start and end time points of the landing phase.

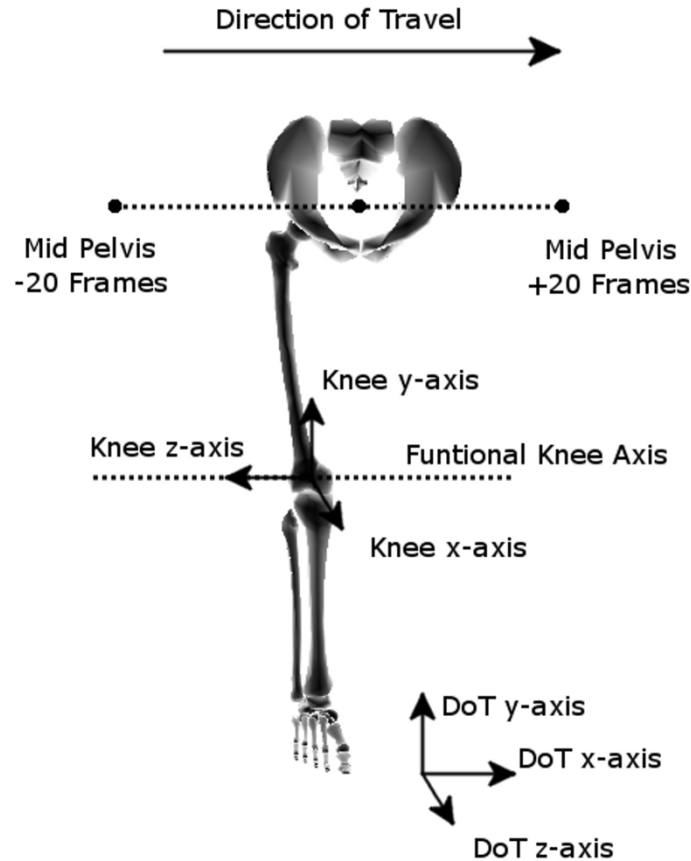
The largest peak valgus moment during the entire landing phase was selected for analysis because this was deemed to reflect the instant where this moment places an athlete at greatest risk of injury (Hewett et al., 2005, Markolf et al., 1995, Fleming et al., 2001). The moments were normalized to each subject's height (m) multiplied by their mass (kg) (Dempsey et al., 2009, Stoffel et al., 2010, Hewett et al., 2005, Yu et al., 2006, McLean et al., 2007, Chaudhari et al., 2005, Cochrane et al., 2010).

To characterize the landing body posture the following kinematic variables were determined at initial foot contact of the support limb: foot rotation, ankle plantar/dorsi flexion, ankle inversion/eversion, knee flexion/extension, hip flexion/extension, hip abduction/adduction, hip internal/external rotation, torso flexion/extension, torso lateral flexion and torso rotation. Initial foot contact was selected for two reasons. Studies undertaking video analysis of actual ACL injuries consistently report joint posture at initial foot contact (Hewett et al., 2009, Olsen et al., 2004, Cochrane et al., 2007). As we are hypothesizing that postures related to high knee joint moments will be similar to those seen in actual injury, we needed to select the same time point. Secondly, previous studies that have examined the relationship between sporting technique and non-contact ACL injuries have used initial foot contact (Dempsey et al., 2007, McLean et al., 2005).

In order to identify the orientation of the knee relative to the person's direction of travel, the knee-path rotation angle was calculated in VICON Bodybuilder. This was defined as the rotation of a knee coordinate system around the y-axis of the person's direction-of-travel coordinate system (Figure 2). The knee coordinate system was defined with the origin at the

knee joint center; the z-axis was the functional knee flexion/extension axis previously calculated (Besier et al., 2003b), with positive being left to right; the y-axis was a unit vector along the plane defined by the knee joint center and hip joint center orthogonal to the z-axis, with positive being superior; and the x-axis was a unit vector orthogonal to both the z-axis and y-axis, with positive being anterior. The direction-of-travel coordinate system was defined with the origin being the global origin; the x-axis being the vector running from the x and z position of the mid-pelvis virtual marker (midpoint between the left and right anterior superior iliac spines, left and right posterior superior iliac spines markers) 20 frames prior to the current frame to the x and z components of the mid-pelvis point 20 frames after the current frame, with positive being the direction of travel; the y-axis was the unit vector of the global y-axis, with positive being superior; and the z axis was orthogonal to x-axis and y-axis, with positive going left to right. A standard joint coordinate system approach was used to calculate the knee-path rotation angle.(Grood and Suntay, 1983) The knee coordinate system was used as the parent segment and the direction-of-travel coordinate system the child. As such the third term from this calculation was the rotation of the knee joint coordinate system around the y-axis of the direction-of-travel coordinate system. Outputs were adjusted such that a negative knee-path rotation angle indicates the knee was externally rotated to the person's direction of travel.

In order to identify the differences in the knee moments between the four landing tasks we used a one way (landing task) repeated measures ANOVA for each moment (SPSS 15.0 - SPSS Inc., Chicago, USA). When there were significant main effects within each ANOVA, a *post hoc* test was performed using a Sidak correction. Significance was set at  $p < 0.05$  in all these statistical tests.



**Figure 2** Schematic of the knee-path rotation angle calculation. This figure reflects an knee with is externally rotated by approximately 90° to the direction of travel. DoT – Direction of travel coordinate system. The skeleton in this figure was generated usin Schematic of the knee-path rotation angle calculation. This figure reflects an knee with is externally rotated by approximately 90° to the direction of travel. DoT – Direction of travel coordinate system. The skeleton in this figure was generated using OpenSIM (Delp et al., 2007).

Relationships between full body kinematics and knee moments for the four landing conditions were identified using data pooled within participants. Intra-participant correlations were performed for all positional data at initial foot contact and the normalized peak valgus and peak internal rotation moments. Based on the approach by McLean and colleagues (2005) for sidestep cutting tasks, a one sample *t*-test was used to identify whether the mean correlation slope values were significantly different from zero ( $p < 0.05$ ). In recognition of the exploratory nature of the study and to improve functional relevance, effect sizes were also calculated using G\*Power (Faul et al., 2007). Variables approaching significance that had a moderate ( $d = 0.41$ ) or higher effect size (Thomas et al., 1991) were

considered to be relevant in terms of reducing knee loads and, therefore, potentially reducing ACL injury risk.

## Results

There was no significant difference between any of the four landing tasks for either the peak flexion moment ( $p = 0.270$ ) or peak internal rotation moment ( $p = 0.441$ ) of the support limb. However, there was a significant difference between tasks for the peak valgus moment ( $p = 0.001$ ). The *post hoc* test showed that the TE condition ( $0.43$  (SD  $0.24$ )  $\text{Nm}\cdot\text{kg}^{-1}\cdot\text{m}^{-1}$ ) had a significantly greater valgus moment than both the AE ( $0.23$  (SD  $0.17$ )  $\text{Nm}\cdot\text{kg}^{-1}\cdot\text{m}^{-1}$ ,  $p = 0.001$ ) and AL conditions ( $0.31$  (SD  $0.16$ )  $\text{Nm}\cdot\text{kg}^{-1}\cdot\text{m}^{-1}$ ,  $p = 0.005$ ), whereas TL ( $0.36\pm 0.21$   $\text{Nm}\cdot\text{kg}^{-1}\cdot\text{m}^{-1}$ ) was significantly higher than AE ( $p = 0.001$ ).

The whole body kinematics that were significantly correlated to the peak valgus moment (i.e. slopes different to zero) were: knee flexion/extension, torso lateral flexion, torso rotation and foot rotation (Table 1). Effect sizes for these relationships ranged from  $0.65$  to  $1.2$ . Hip flexion/extension was not significantly correlated to the peak valgus angle but displayed a medium effect size. Specifically, increased valgus moments were associated with; increased knee flexion, increased hip flexion, increased lean of the torso over the support leg, increased torso rotation towards the support leg, and an externally rotated foot (Figure 3). An increasing valgus moment was also significantly correlated with more externally rotated relative to the direction of travel ( $p = 0.004$ ,  $d = 1.0$ ) (Table 1). The strongest correlations were for knee-path rotation ( $r^2 = 0.27 \pm 0.20$ ), torso rotation ( $r^2 = 0.27 \pm 0.20$ ), and foot internal/external rotation ( $r^2 = 0.26 \pm 0.20$ ).

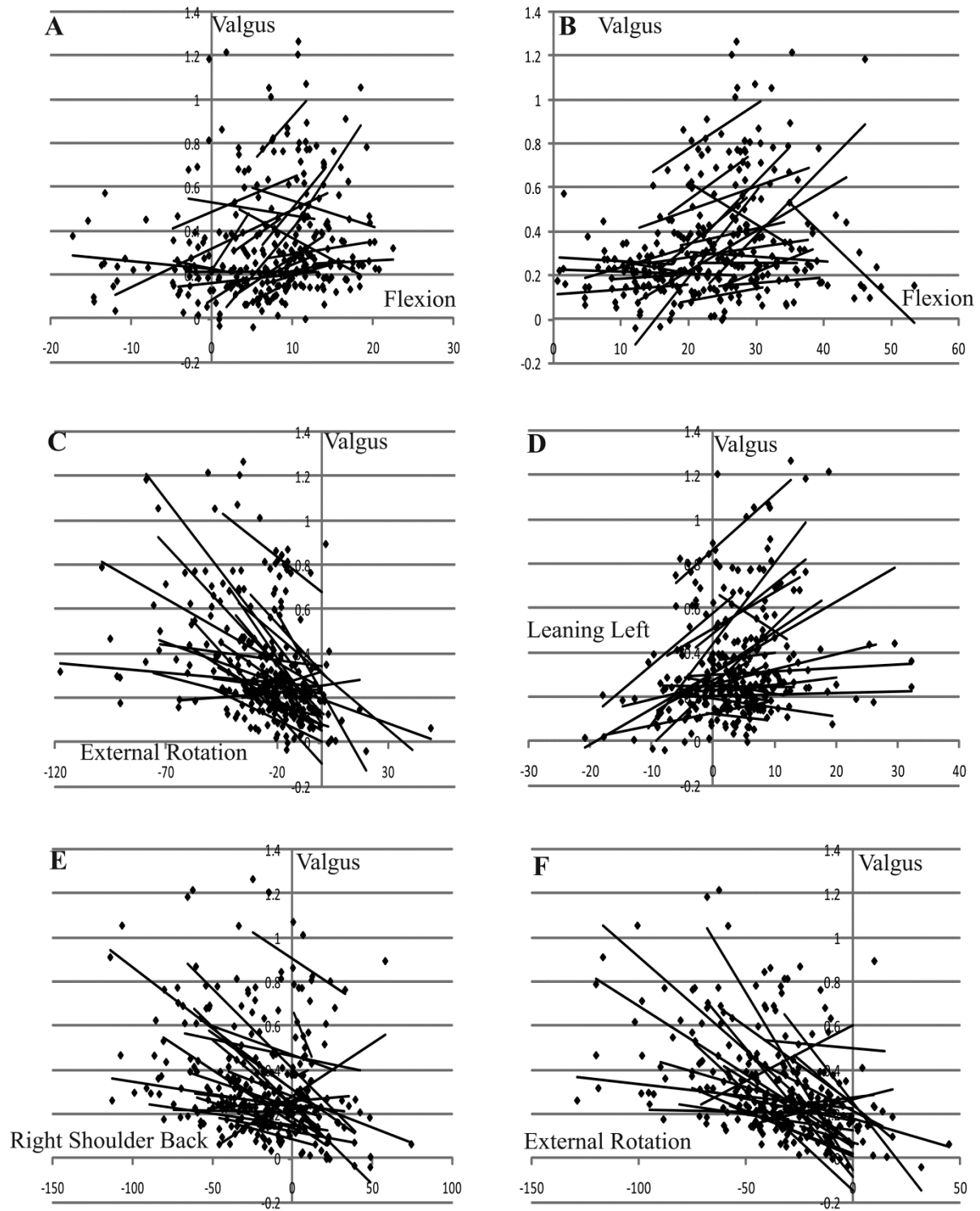
**Table 1** Mean (standard deviation) of the slopes between position data at initial contact and the peak valgus moment with the associated p value, effect size (d) and r<sup>2</sup> value (bolded values indicate significance difference for the slope).

	Slope	<i>p</i>	<i>d</i>	<i>r</i> <sup>2</sup>
Knee Flexion/Extension	<b>0.015 (0.023)</b>	<b>0.005</b>	<b>0.65</b>	<b>0.14 (0.11)</b>
Hip Flexion/Extension	0.007 (0.015)	0.053	0.46	0.15 (0.16)
Hip Abduction/Adduction	-0.003 (0.014)	0.411	0.21	0.11 (0.11)
Hip Internal/External Rotation	0.004 (0.012)	0.148	0.33	0.14 (0.13)
Ankle Plantar/Dorsi Flexion	-0.007 (0.019)	0.088	0.36	0.13 (0.16)
Ankle Inversion/Eversion	0.002 (0.037)	0.832	0.05	0.17 (0.16)
Foot Internal/External Rotation	<b>-0.006 (0.005)</b>	<b>0.001</b>	<b>1.20</b>	<b>0.26 (0.18)</b>
Torso Flexion/Extension	-0.001 (0.008)	0.289	0.12	0.10 (0.13)
Torso Lateral Flexion	<b>0.008 (0.010)</b>	<b>0.001</b>	<b>0.80</b>	<b>0.16 (0.16)</b>
Torso Rotation	<b>-0.002 (0.003)</b>	<b>0.001</b>	<b>0.66</b>	<b>0.27 (0.20)</b>
Knee Rotation relative to Path	<b>-0.004 (0.004)</b>	<b>0.004</b>	<b>1.00</b>	<b>0.27 (0.20)</b>

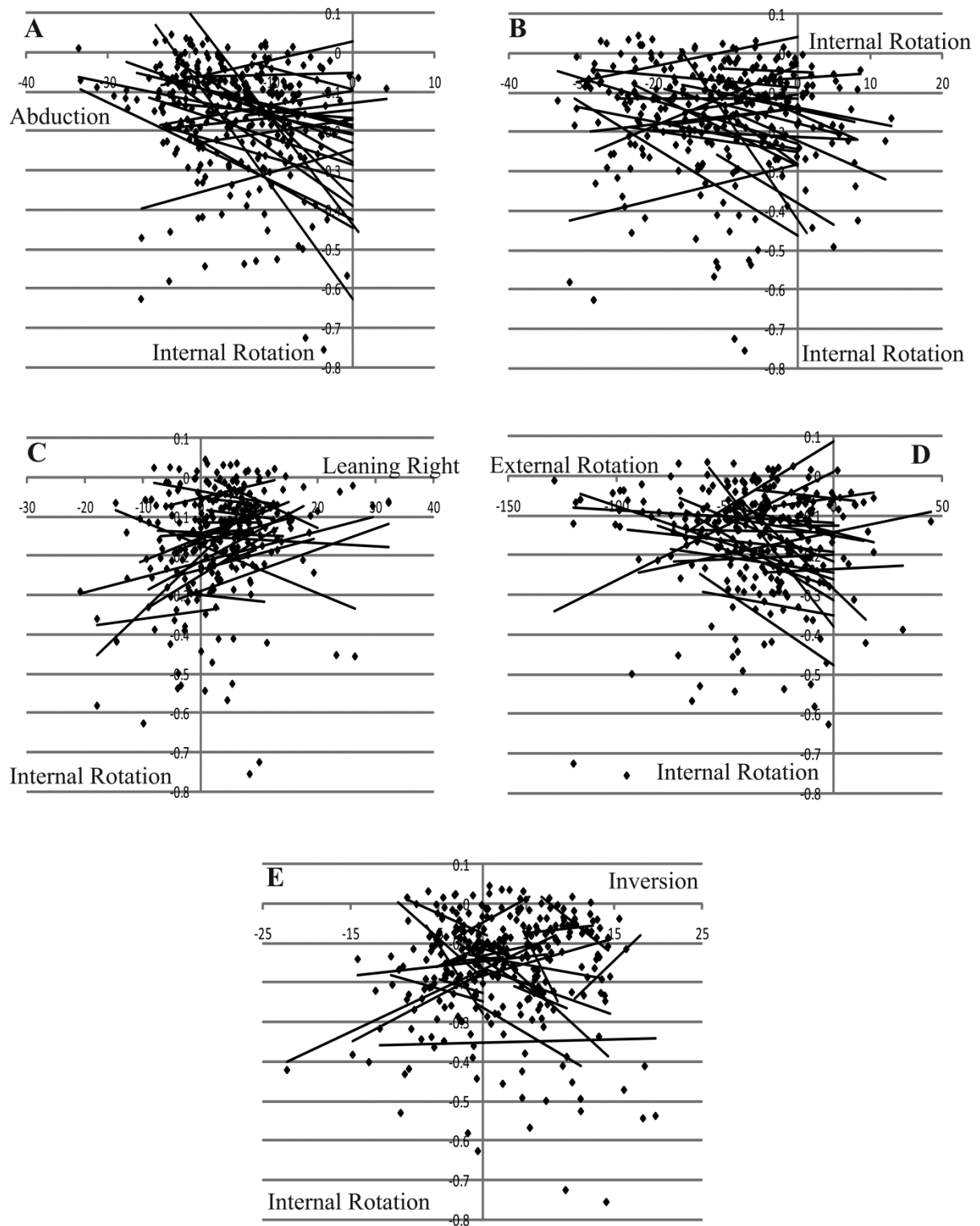
The whole body kinematics that were significantly correlated with the peak internal rotation moment (i.e. slopes different to zero) were: hip abduction/adduction, hip internal/external rotation, ankle inversion/eversion and torso lateral flexion (Table 2). Effect sizes for these relationships ranged from 0.50 to 0.60. More specifically, higher internal rotation moments were associated with less hip abduction, less hip external rotation, more ankle inversion, and greater torso lean away from the support leg (Figure 4). Higher internal rotation moments were also significantly correlated with greater external rotation of the knee relative to the direction of travel, as they returned a slope significantly different to zero (Table 2). When correlated to the internal knee rotation moment, torso lateral flexion had an  $r^2 = 0.15 \pm 0.17$ , whereas hip abduction/adduction, ankle inversion/eversion and knee/path rotation all had an  $r^2 = 0.14$ .

**Table 2** Mean (standard deviation) of the slopes between position data at initial contact and the peak internal rotation moment with the associated p value, effect size (d) and r<sup>2</sup> value (bolded values indicate significance difference for the slope).

	Slope	p	d	r <sup>2</sup>
Knee Flexion/Extension	-0.003 (0.017)	0.456	0.18	0.15 (0.15)
Hip Flexion/Extension	-0.002 (0.008)	0.270	0.25	0.13 (0.16)
Hip Abduction/Adduction	<b>-0.006 (0.010)</b>	<b>0.005</b>	<b>0.60</b>	<b>0.14 (0.17)</b>
Hip Internal/External Rotation	<b>-0.004 (0.007)</b>	<b>0.025</b>	<b>0.57</b>	<b>0.08 (0.08)</b>
Ankle Plantar/Dorsi Flexion	0.007 (0.033)	0.303	0.21	0.14 (0.14)
Ankle Inversion/Eversion	<b>-0.008 (0.014)</b>	<b>0.012</b>	<b>0.57</b>	<b>0.14 (0.18)</b>
Foot Internal/External Rotation	-0.001 (0.003)	0.061	0.33	0.15 (0.14)
Torso Flexion/Extension	-0.003 (0.008)	0.137	0.38	0.14 (0.19)
Torso Lateral Flexion	<b>0.003 (0.005)</b>	<b>0.023</b>	<b>0.60</b>	<b>0.15 (0.17)</b>
Torso Rotation	-0.000 (0.002)	0.680	0.00	0.12 (0.15)
Knee Rotation relative to Path	<b>-0.001 (0.002)</b>	<b>0.034</b>	<b>0.50</b>	<b>0.14 (0.16)</b>



**Figure 3** Scatter plots of position data versus peak valgus rotation moment for position variables with slopes significantly different from zero. Black lines are regression lines for each subject. A – knee flexion/extension; B – hip flexion/extension; C – foot internal/external rotation; D – torso lateral flexion; E – torso rotation; F – knee rotation relative to path. Vertical axis represents the varus/valgus moment in Nm•kg<sup>-1</sup>•m<sup>-1</sup>. Horizontal axis represents the joint angle in degrees.



**Figure 4** Scatter plots of position data versus peak internal rotation moment for position variables with slopes significantly different from zero. Black lines are regression lines for each subject. A – hip abduction/adduction; B – hip internal/external rotation; C – torso lateral flexion; D – knee rotation relative to path, E- ankle inversion\eversion. Vertical axis represents the internal/external moment in Nm·kg<sup>-1</sup>·m<sup>-1</sup>. Horizontal axis represents the joint angle in degrees



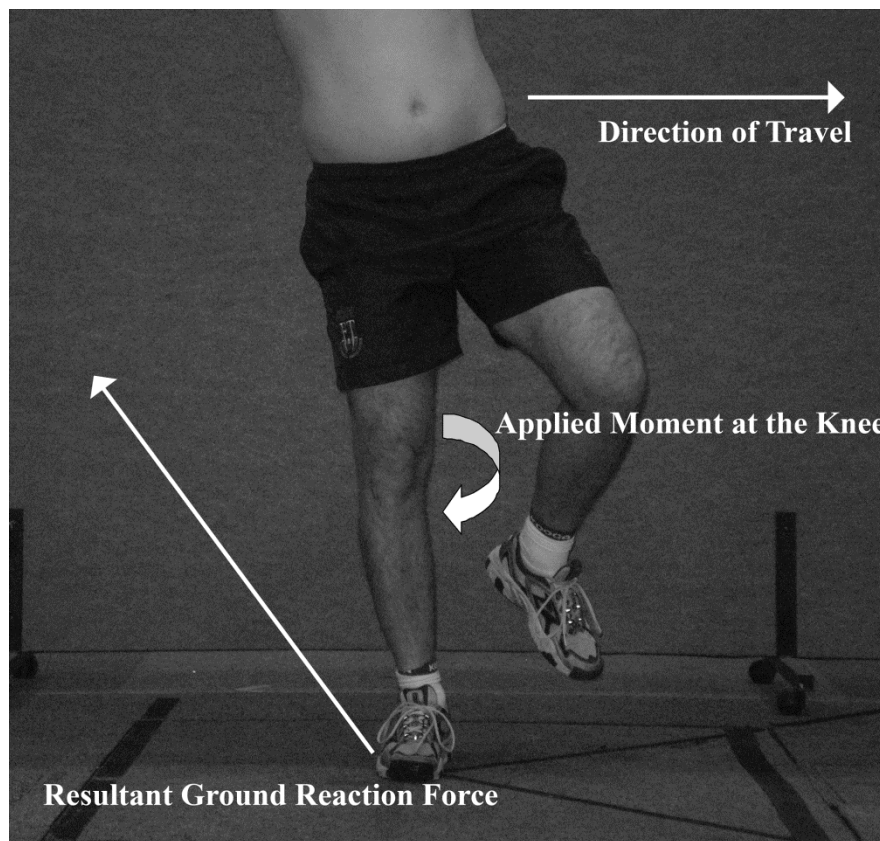
## Discussion

This study first aimed to investigate the impact of different ball movement conditions during an overhead ball catch and landing task on knee moments. We found that catching a ball that was dropping and moving *toward* the participants' preferred landing leg caused larger peak knee valgus moments compared to catching a ball dropping and moving *away* from the participants' preferred landing leg. No significant between-task differences were observed for any of the other knee moments. In the second aim, we investigated the relationships between joint postures at initial foot contact and peak knee valgus and internal rotation moments. Although there were no significant differences in peak internal rotation moments between ball movement conditions, it was decided to continue with the analysis of this moment, as it was thought that the relationship between internal rotation moment and technique may have been independent of ball movement. The variations in body postures induced by altering the landing task allowed us to identify specific joint postures associated with higher knee valgus and internal rotation moments.

Specifically, we found that landing with the knee externally rotated relative to the person's direction of travel was correlated with both increased peak valgus and peak internal rotation moments. An increased valgus moment was also associated with increased knee flexion, increased hip flexion, increased torso lean over the support leg, increased torso rotation towards the support leg, and an externally rotated foot. A higher peak internal rotation moment was linked to less hip abduction, less hip external rotation, more ankle inversion and the torso leaning away from the support leg. Most of these specific joint postures reflect landing postures that have previously been observed during non-contact ACL injuries (Hewett et al., 2009, Olsen et al., 2004, Cochrane et al., 2007).

Although each of the relationships of the joint postures with knee joint moments may have had slopes significantly different from zero, the  $r^2$  values were small, ranging from 0.14 to 0.27 for peak valgus and 0.08 to 0.15 for peak internal rotation moments. However, these  $r^2$  values were of similar magnitude to those reported for the relationships between peak valgus moments with knee varus/valgus angles or hip flexion/extension angles in McLean et al (2005), although their hip internal rotation had  $r^2$  in excess of 0.50. As their results were used to stimulate further studies into technique and injury risk, the postural associations with knee loading found in the current study can be used to inform future studies of landing techniques to reduce risk of ACL injury.

Having the knee externally rotated relative to the direction of travel was associated with both high valgus and high internal rotation moments at the knee. The increase in valgus moment was most likely due to the posteriorly directed braking component of the ground reaction force during the initial landing period having a line of action partly directed from the medial to lateral direction across the knee (Figure 5). We also observed that external foot rotation was correlated with the peak valgus moment, which is consistent with and confirms the “position of no return” described by Ireland (2002), whereby external foot rotation is commonly seen during ACL injury.



**Figure 5** How external rotation of the knee relative to the direction of travel may increase knee valgus moment. The action of the posteriorly directed resultant ground reaction force relative to the direction of travel will cause a valgus moment at the knee, when the medial aspect of the knee is facing the direction of travel

It has been argued in the literature that an increase in knee flexion during landing tasks would be beneficial for reducing ACL injury risk (Decker et al., 2003, Ireland, 2002, Kernozek et al., 2008). However, the current study found that an increased knee flexion angle at initial ground contact was correlated with increased peak valgus moments. Furthermore, reduced peak knee flexion during landing tasks has previously been associated with an increased risk of suffering an ACL injury (Hewett et al., 2005). It may be that athletes who make initial contact with a more extended knee joint, and then move into flexion, increase the time over which the force is absorbed and, therefore, reduce peak loading (Schmitz et al., 2007, Norcross et al., 2010). Therefore, there may be a compromise between the best knee flexion angle at initial contact, where the level of flexion in early landing is sufficiently high to

ensure low ACL strain from the applied load (Markolf et al., 1995), while still remaining sufficiently extended to allow sufficient flexion post landing.

The current study found that increased hip flexion was associated with increased peak valgus moments during landing tasks. Although apparently different to the literature (Lephart et al., 2002, Decker et al., 2003, Kernozek et al., 2008) the majority of hip flexion angles identified in this study were similar to those seen during actual injury episodes, particularly those around where the highest valgus moments occurred (Krosshaug et al., 2007). As we proposed for the knee, increasing hip flexion at initial contact may reduce the range the athlete has available to flex the lower limb (Schmitz et al., 2007, Norcross et al., 2010). This will thereby reduce time over which force absorption occurs resulting in higher peak moments.

The correlation of higher internal rotation moments at the knee with less hip abduction and less hip external rotation is again reflective of the “position of no return” described by Ireland (2002). Previously, plyometric-based intervention studies have used a “knee over toe” position as a teaching guide for good landing technique (Olsen et al., 2005). An athlete landing with joint postures associated with higher valgus and internal rotation moments would be in direct contrast to the suggested ideal “knee over toe” posture.

Torso rotations have been shown to impact on both peak valgus and peak internal rotation moments during sidestep cutting tasks (Dempsey et al., 2007). Additionally, the inability to control the trunk, particularly after lateral perturbations, is predictive of future ACL injuries (Zazulak et al., 2007). It is therefore not surprising that torso rotations were associated with high valgus and internal rotation loads during landing tasks. During sidestep cutting, pure technique training focusing on bringing the torso upright, together with bringing the foot closer to the midline, has previously resulted in reduced peak knee valgus moments (Dempsey et al., 2009). This would suggest that a similar technique training focused

approach may be successful in landing tasks. Based upon the results from this study, athletes should be discouraged from landing with their torso laterally flexed, hip internally rotated or their knee or foot externally rotated. Instead, they should strive for an upright and forwards facing torso with their leg rotated such that the knee and foot are pointing in the direction of travel. Although, due to the flow of play in a game, athletes may find it difficult to not lean over and rotate their body to catch a ball located on the same side as their landing leg, they may be taught to move their body following the catch. They should be taught to move to a body posture not associated with high knee moments. Future work should investigate this notion further.

As this is the first study to directly investigate associations between full body kinematics and knee loads in landing tasks, further investigation should be undertaken to confirm the results and clarify the relationship of loading to knee and hip flexion. This future work should take into account the influence of hip and knee flexion angles on support of knee loads provided by the musculature crossing the knee (Buford et al., 2001, Lloyd and Buchanan, 2001). There are several methodological considerations that should also be undertaken when designing subsequent studies. Firstly, the landing task should be reflective of game scenarios. The task utilized in this study was designed to mimic an overhead mark in Australian football. It has been shown that ball positioning affects both the kinematics and kinetics of both landing and sidestep cutting (Chaudhari et al., 2005, Cowling and Steele, 2001). Therefore, tasks designed to replicate other sports may produce different results. Secondly, the methodology of identifying variations needs to be considered. In the current study, variability was introduced by using different ball movements in an overhead catch to induce changes during the same task. McLean et al. (2005) utilized natural variation within sidestep cutting, while Dempsey and colleagues (2007) assessed the impact of imposed techniques within sidestep cutting. Therefore, future landing studies may also use natural

postural variation or direct manipulation of landing posture to examine how knee joint loading is affected by whole body posture.

The main limitation to this study is that we analyzed each joint posture's independent effect on the knee moments. This approach does not take into consideration the combined joint postural effects, as the resultant loads at the knee are a result of whole body positioning. Proposing technique modifications based on pulley on the results of this study also does not address the fact that joint angles are also dependent upon each other. That is, it has previously been shown that altering one joint in a landing task will affect other joints in the same limb (Whitting et al., 2009). Future studies therefore need to consider the interrelatedness of joints, and impact on joint moments, in their study design.

## **Conclusions**

This study showed that when an individual is attempting to catch an overhead pass, movement of the ball towards the support leg before the catch is made affects the subsequent landing posture and increases the knee joint valgus moment during the land. We also identified certain postures at initial foot contact during landing that were associated with increased valgus and internal rotation loads at the knee. Specifically, landing with an externally rotated foot, with the knee also externally rotated to the direction of travel, with an abducted and internally rotated hip and a laterally flexed or rotated torso were associated with higher valgus and internal rotation loads at the knee and, in turn, appear to be associated with an increased risk of ACL injury. Learning to land with techniques that do not reflect these postures that are associated with higher valgus and internal rotation loads at the knee might reduce an athlete's risk of non-contact ACL injury. We found conflicting results for knee and

hip flexion and further work is needed to understand the relationship between these angle and joint loading.

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## **Conflict of Interest**

No authors have a conflict of interest

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**Supplementary Material** Ball movement rig. The ball was suspended from a gantry above the force plate and released by the same experimenter for each participant. Release of the right catch caused the ball to fall to the left.

