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The effect of isolated valgus moments on ACL strain during single-leg landing: A simulation study

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ABSTRACT

Valgus moments on the knee joint during single-leg landing have been suggested as a risk factor for anterior cruciate ligament (ACL) injury. The purpose of this study was to test the influence of isolated valgus moment on ACL strain during single-leg landing. Physiologic levels of valgus moments from an *in vivo* study of single-leg landing were applied to a three-dimensional dynamic knee model, previously developed and tested for ACL strain measurement during simulated landing. The ACL strain, knee valgus angle, tibial rotation, and medial collateral ligament (MCL) strain were calculated and analyzed. The study shows that the peak ACL strain increased nonlinearly with increasing peak valgus moment. Subjects with naturally high valgus moments showed greater sensitivity for increased ACL strain with increased valgus moment, but ACL strain plateaus below reported ACL failure levels when the applied isolated valgus moment rises above the maximum values observed during normal cutting activities. In addition, the tibia was observed to rotate externally as the peak valgus moment increased due to bony and soft-tissue constraints. In conclusion, knee valgus moment increases peak ACL strain during single-leg landing. However, valgus moment alone may not be sufficient to induce an isolated ACL tear without concomitant damage to the MCL, because coupled tibial external rotation and increasing strain in the MCL prevent proportional increases in ACL strain at higher levels of valgus moment. Training that reduces the external valgus moment, however, can reduce the ACL strain and thus may help athletes reduce their overall ACL injury risk.

1. Introduction

Injuries to the anterior cruciate ligament (ACL) frequently occur during the deceleration phase of landing or in preparation for a change of direction (Boden et al., 2000; Griffin et al., 2000). Females who participate in sports that include jumping and cutting often suffer from ACL injuries significantly higher than males (Agel et al., 2005; Gwinn et al., 2000). Recent studies have suggested that the gender difference in dynamic frontal-plane motion during landing may be associated with higher ACL injury rates in females: women often land with more valgus frontal-plane alignment than men and this valgus alignment caused larger valgus moments to the knee joint (Chaudhari et al., 2003; Kernozek et al., 2005; McLean et al., 2005). A prospective study has shown that female athletes who subsequently ruptured the ACL performed jump landing tasks with significantly higher valgus moments than athletes who did not rupture their ACL (Hewett et al., 2005). However, it remains unknown how much these observed gender differences in dynamic valgus alignment and valgus moments increase ACL strain during landing.

Several studies have shown that valgus loading at the knee joint can increase ACL force (Fukuda et al., 2003; Hollis et al., 1991; Markolf et al., 1995). In contrast, some other studies have not observed significant ACL strain under valgus loading (Bendjaballah et al., 1997; Fleming et al., 2001). Another study observed no significant ACL strain until the medial collateral ligament (MCL) was torn by valgus loading (Mazzocca et al., 2003). However, these previous studies were often performed while constraining other degrees of freedom (DOFs) or by applying valgus loading without shear forces that occur during landing. Further, cadaver studies which predict static characteristics of the knee joint under low levels of loading may not predict ACL rupture under the large loading magnitudes and loading rates experienced during sports activities. *In vivo* studies of ACL strain during injury-causing events are not feasible for human subjects, either. Dynamic three-dimensional simulation studies offer an attractive alternative, because they can include more joint

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complexity, allow unconstrained motion, and permit physiologic loads to be applied. However, predictions of ACL strain during dynamic landing have not been previously studied using a model validated for the estimation of ACL strain. The purpose of this study was to test the influence of isolated valgus moment on ACL strain during single-leg landing using a dynamic three-dimensional simulation model driven by *in vivo* human loading data.

2. Methods

We developed a three-dimensional dynamic specimen-specific force-driven knee model, tested for ACL strain measurement against previous cadaveric experiment. In addition to vertical dynamic impact load, which simulates loads caused by landing, physiologic levels of valgus moments from a previous *in vivo* study were scaled and applied to this knee model to investigate the influence of valgus moments on the ACL.

The development and validation test of this knee model to predict ACL strain during landing under a vertical impact force (without additional externally applied torques) has been described in detail by Shin et al. (2007), thus a brief description is given here. The knee model was constructed from sagittal MRI (GE 3D-Spoiled-Gradient-Recalled-Echo, 1.5T, $140\times140\,\mathrm{mm}^2$, 256×256, thickness 1.5 mm) of a cadaveric knee (Fig. 1). MR images were segmented and imported into dynamic rigid-body motion simulation software (MSC.ADAMS, MSC.Software, Santa Ana, CA). This knee model includes ligament bundles (the ACL, posterior cruciate ligament (PCL), MCL, LCL, posterior capsules, and patellar ligament) formulated as non-linear elastic springs (Shin et al., 2007) with properties adapted from published data (Abdel-Rahman and Hefzy, 1998; Cooper et al., 1993; Noyes et al., 1984; Shelburne and Pandy, 1997; Woo et al., 1991; Yu et al., 2001). The origins and insertions of the ACL/PCL were determined based on segmented MRI. Two functional bundles of the ACL/PCL were identified as previously quantified (Harner et al., 1999), and the centroids of each region were estimated to be the insertion points. The MCL, LCL, and posterior capsule were placed over the appropriate bony landmarks (Garg and Walker, 1990; Reicher, 1993) using the same method and bundle orientation as described by Yu et al. (2001). The patellar ligament was modeled by medial/lateral bundles and placed based between the patellar apex and tibial tuberosity. The contact forces at the tibiofemoral and patellofemoral articulation were defined using a penalty regulation of normal contact force constraints (Lötstedt, 1982) with previously reported properties (Nam et al., 2004; Oni and Morrison, 1998). The passive characteristics of the knee model were tested under various conditions, including valgus rotational stiffness and passive flexion movement to ensure proper tibiofemoral behavior. The femoral rollback and screw-home motion were properly simulated with the model (Shin, 2006; Shin et al., 2007).

To simulate the motion of a single-leg landing, a simulated landing apparatus was created with the same geometric configuration using the same cadaver knee specimen (Fig. 2) as in a previously described cadaver experiment (Withrow et al., 2006). Three musculotendinous groups (the quadriceps, medial/lateral hamstrings, and medial/lateral gastrocnemius) were modeled as linear tension-springs with the same pretension and stiffness used in the experiment to provide the appropriate tension to hold the initial knee flexion angle at 25° before impact load and to simulate eccentric contraction in the quadriceps as done in the physical experiment (Withrow et al., 2006). The proximal femur was connected to the mounting apparatus through a spherical joint and linear slide, allowing free rotation analogous to a hip joint and vertical motion. The distal tibia was

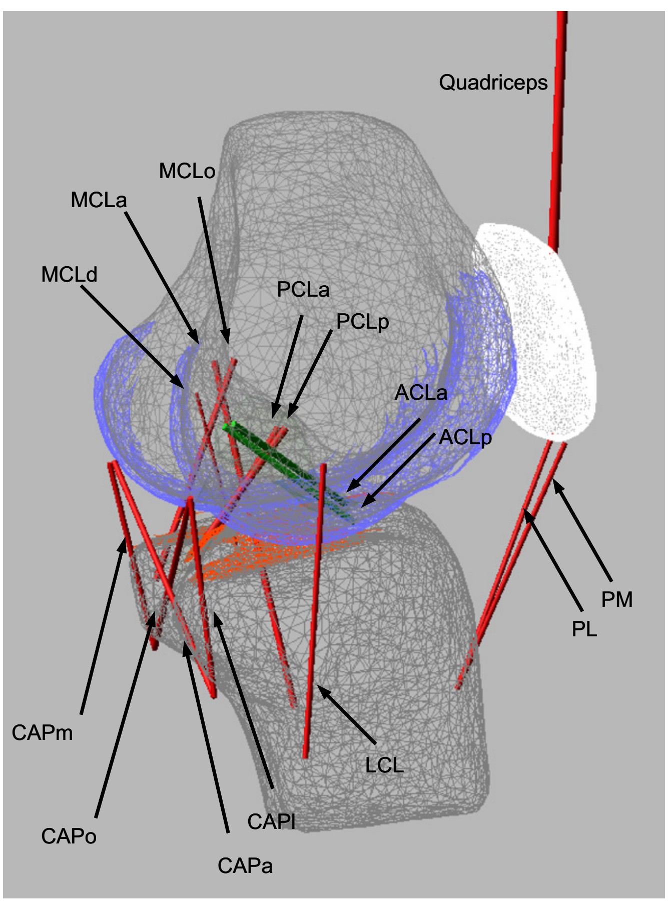

Fig. 1. Schematic of the knee model showing the locations and numbers of bundles of the modeled ligaments: the anterior and posterior bundle of the anterior cruciate ligament (ACLa and ACLp, green); the anterior and posterior bundle of the posterior cruciate ligament (PCLa and PCLp); the lateral collateral ligament (LCL); the anterior, oblique and deep bundle of the medial collateral ligament (MCLa, MCLo, and MCLd); the medial, lateral, oblique popliteal and arcuate popliteal bundle of the posterior capsules (CAPm, CAPl, CAPo, and CAPa); and the medial and lateral patellar ligament (PM and PL). (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

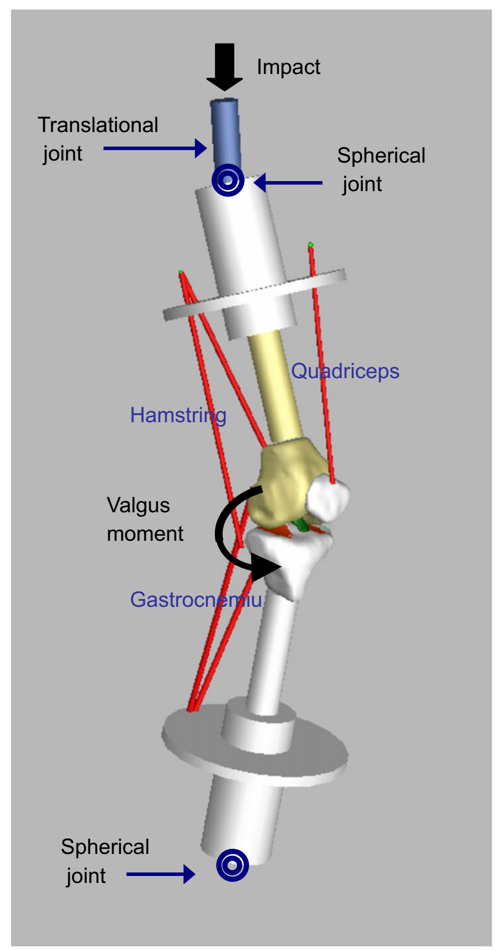

Fig. 2. An illustration of the knee model with the simulated dynamic landing apparatus showing five musculotendinous bundles (the quadriceps, the hamstrings, and the gastrocnemius). The segmented ACL is shown in green. The tibial cartilage is shown in orange. The femoral cartilage is not shown to enhance the inside view of the tibiofemoral joint. Before external loading is applied, the muscles of the knee joint are pretensioned to hold 25° of flexion. The impact force was applied at the top of the femoral axis of the upper limb and the valgus moment was applied at the tibiofemoral knee joint. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)