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A Knee-Specific Finite Element Analysis of the Human Anterior Cruciate Ligament Impingement against the Femoral Intercondylar Notch

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Abstract

This work presents a finite element analysis of Anterior Cruciate Ligament (ACL) impingement against the intercondylar notch during tibial external rotation and abduction, as a mechanism of noncontact ACL injuries. Experimentally, ACL impingement was measured in a cadaveric knee in terms of impingement contact pressure and six degrees-of-freedom tibiofemoral kinematics. Three-dimensional geometries of the ACL, femur and tibia were incorporated into the finite element model of the individual knee specimen. A fiber-reinforced model was adopted, which accounts for the anisotropy, large deformation, nonlinearity and incompressibility of the ACL. With boundary conditions specified based on the experimental tibiofemoral kinematics, the finite element analysis showed that impingement between the ligament and the lateral wall of intercondylar notch could occur when the knee 45° was externally rotated at 29.1° and abducted at 10.0°. Strong contact pressure and tensile stress occurred at the impinging and nonimpinging sides of the ligament, respectively. The impingement force and contact area estimated from the model matched their counterparts from the corresponding cadaver experiment. The modeling and experimental approach provides a useful tool to characterize potential ACL impingement on a knee-specific basis, which may help elucidate the ACL injury mechanism and develop more effective treatments.

Keywords

Anterior Cruciate Ligament (ACL); Noncontact ACL Injury; ACL Impingement; Finite Element Analysis

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I. Introduction

The anterior cruciate ligament (ACL) is the most frequently injured knee ligament and contributes to knee stability in multiple directions (Arms et al., 1984; Beynnon et al., 1992; Daniel et al., 1994; Fetto and Marshall, 1980; Griffin et al., 2000; Li et al., 1999). Although external tibial rotation and knee valgus may not load the ACL (Markolf et al., 1995; Miyasaka et al., 2002), the ACL may be injured via its impingement against the lateral wall of the intercondylar notch (Fung et al., 2007; Fung and Zhang, 2003; LaPrade and Burnett, 1994).

Various computational models of the ACL have been developed to evaluate ligament behavior in health, disease and injury (Hirokawa and Tsuruno, 2000; Song et al., 2004). In particular, the finite element (FE) method has been shown to be an effective approach characterizing stress distributions in the ACL in response to loading and tibiofemoral movements.

The purpose of the current study is to evaluate an FE ACL-impingement model built from a cadaveric knee specimen with the ACL characterized *heterogeneously* using longitudinally aligned fibers and ground substance. The model was validated based on the corresponding experimental impingement force and tibiofemoral kinematic data collected from the cadaveric specimen subjected to joint movements associated with impingement. We sought to evaluate stress distributions within the ACL and contact pressure on its surface, thereby, identifying specific regions in the ligament that are particularly susceptible to tissue injury resulting from impingement of the ligament against the intercondylar notch.

II. Methods

A. ACL-Impingement Experiment

A fresh-frozen cadaver knee with a relatively narrow notch which showed representative impingement was used in this study (Fung and Zhang, 2003; Zhang et al., 2003). The tibia was moved strenuously by hand to about 30 degree tibial external rotation and 10 degree abduction with the knee flexion angle at about 45 degrees.

Six degree-of-freedom (DOF) knee joint kinematics was measured by tracing three infrared Optotrak™ markers attached to the tibia and three to the femur (Optotrak™ 3020, NDI, with 0.1mm accuracy in horizontal plane, 0.15mm in vertical axis, and 0.04~0.06 degree of angular accuracy). The knee flexion/extension, abduction/adduction and the internal/external rotation angles as defined in the Joint Coordinate System (Grood and Suntay, 1983) were derived from the marker recordings. At the same time, the contact pressure distribution between the ligament and the lateral notch wall was measured with a thin, flexible pressure transducer (6900 series, Tekscan, Inc., South Boston, MA).

B. Building a Geometric Model of the Knee

After the experiment, the knee was dissected and the surface geometry of the ACL was captured by digitizing the ligament surface using an Optotrak™ digitizing probe. The knee was fixed at 30° of flexion (neutral in other DOFs) and positioned to reach slight tautness in the ligament fibers, and this ligament conformation was used as the stress-free initial condition in the FE analysis. The orientations of the fibers on the surface of the ACL were digitized as well, so that these can be interpolated to define orientations of the internal fibers in the ACL model assuming that the orientation of fibers in the ACL vary continuously. The surface geometries of the femur and the tibia were attained using a laser scanner (LPX-250, Roland Corp., rotary and planar resolution: 0.2° and 0.2mm).

C. Finite Element Analysis

The ACL was modeled as a composite structure consisting of pseudo fibers embedded in an isotropic hyperelastic ground substance matrix, modeled as hybrid linear hexahedral elements, with incompressible neo-Hookean material properties (Hirokawa and Hasezaki, 2010; Limbert et al., 2004; Reese et al., 2010). The ground substance matrix is described by the following energy equation:

$$\psi_m = C_1(I_1 - 3) \quad (1)$$

where ψ_m was the strain energy, I_1 the first invariant of the right Cauchy-Green strain and C_1 a material constant. The fibers in the ACL are represented by nonlinear spring elements that coincided with the vertices of the hexahedral elements. These spring elements are characterized by the empirically obtained nonlinear force-stretch relationship in Equations (2) and (3). The equations were modifications of the transversely isotropic constitutive law, applicable in fiber-reinforced matrix models (Weiss et al., 1996).

$$\sigma = \begin{cases} 0 & \text{if } \lambda < 1 \\ C_2(e^{C_3(\lambda-1)} - 1) & \text{if } 1 \leq \lambda < \lambda^* \\ C_4\lambda + C_5 & \text{if } \lambda^* \leq \lambda \end{cases} \quad (2)$$

and

$$C_5 = C_2(e^{C_3(\lambda^*-1)} - 1) - C_4\lambda^* \quad (3)$$

where σ was the tensile stress in the pseudo fibers, λ the stretch along the fiber, and C_1 through C_5 and λ^* were material constants. C_1 was set to 2.9 MPa, and C_2 through C_5 were determined with custom material testing performed in our laboratory (Ren et al., 2010). A flat and uniform strip of the ACL specimen was loaded along two directions (along the fiber and orthogonal to the fiber). As a result, $C_2=1.133$ MPa, $C_3=47.75$, $C_4=349.8$ MPa, $C_5=-357.28$ MPa, $\lambda^*=1.039$. This fiber-reinforced model allowed separate characterizations of stresses in the ground matrix and the fibers.

Displacement boundary conditions were imposed on the femur and tibia according to the tibiofemoral kinematics measured in the experiment. The knee model was implemented and simulated in Abaqus/StandardTM.

III. Results

ACL impingement was detected at 4 degree abduction and 15 degree external rotation, and the impingement force increased with increasing abduction and external rotation (Fig. 1).

FE analysis at 29.1° external rotation and 10.0° abduction showed that the ligament underwent bending and stretching, and deformed to the shape of the corresponding contact area on the lateral notch wall during the impingement (Fig. 2). The impingement force and contact area predicted by the FE analysis matched with those measured during the experiments (Fig. 3). At the peak impingement, the measured force reached 36.9 N, while the model predicted 37.4 N. The difference between the model-predicted and measured peak impingement forces averaged over the kinematic trajectory was 1.03 ± 2.54 N (mean $\pm$ SD).

The measured contact area at the peak impingement was 19.7 mm², while the predicted peak area was 19.2 mm² (Mean $\pm$ SD of difference: 0.89 $\pm$ 1.18 mm²).

The FE analysis showed strong contact pressure (Fig. 4(a)). The FE analysis showed high maximum principal stress on the non-impingement side of the ACL during its impingement, which was likely a result from the combined ligament bending and stretching. The tensile stresses in the embedded fibers and the ground matrix reached 8.9 MPa and 3.0 MPa, respectively, on the non-impingement side (Fig. 4(b)). In contrast, fibers on the impingement side were not stretched.

IV. Discussion

An FE model of the knee joint was developed in this study to examine the impingement behavior of the ACL against the intercondylar notch. Simulated movements in the FE model yielded a close match between the experimental and predicted force and area of impingement, in support of the validity of the proposed model.

In a related animal study, cell death was observed in the impinged area of cyclically loaded ACLs where there was significant contact pressure while those loaded without impingement did not show any cell death (Fung et al., 2005). These observations suggested that repeated impingement may cause local, cumulative damage in the ligament, thereby, leading to progressive weakening of the tissue, and ultimately, structural tear.

The study constitutes a part of an effort to elucidate the ACL impingement as an injury mechanism of the ligament. How impingement 'causes' ligamentous injury has to be confirmed first before the injury 'mechanism' can be addressed. It has been documented in previous studies that ACL injuries occur at tibial external rotation and/or knee valgus at moderate knee flexion (Ireland and Ott, 2001; McLean et al., 2004; Olsen et al., 2004) consistent with those simulated in the current study. Earlier studies have provided evidence in cadaveric knees of the occurrence of impingement (Fung et al., 2007; LaPrade and Burnett, 1994; Ren et al., 2008). Our prior investigation of passive knee laxity in human subjects supports the plausibility of experiencing these movements, especially in women (Park et al., 2008). Based on the modeling capabilities provided by the proposed model, contributions to mechanical conditions and injury from ACL biomechanical properties and notch geometry can be evaluated at variable kinematic positions, thereby, improving our ability to characterize potential mechanisms of injury.

A limitation of the current model similar to other FE models (Hirokawa and Tsuruno, 2000; Limbert et al., 2004) was that stress concentration near the ligament attachment sites might be attributed partially to the non-ideal meshing of the ligament geometry. In addition, the measurement error might have caused a volume error in the ACL model (no more than 6%) and in kinematic movements (1%), which might have contributed to errors in the various measures investigated in this study.

Another limitation was that the ACL was assumed to be stress-free with uniform fiber properties at 30 degree knee flexion (Woo and Adams, 1990) and neutral rotation and abduction, with the zero-strain assumption based on our observation in a related study that ACL strain during simulated walking using a cadaver model was about zero at such a position (Zhang et al., 2001). Furthermore, it was reported that the anterior bundle of the ACL had larger load-related material properties than the posterior bundle (Butler et al., 1992), which was not incorporated in this study.

The above study provides a computational means of estimating deformation of the ACL and stress distribution within the ligament during ACL impingement, which is difficult to

measure experimentally. Further studies on the failure criteria are expected in order to have a better appreciation of the likelihood of injury at various joint configurations.

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Fig. 1. Experimental data. Total impingement force and contact area measured at a kinematic sequences: from Configuration A (flexion: 46.3 deg, abduction: 0 deg, external rotation: 0 deg) to Configuration B (flexion: 44.8 deg, abduction: 10.0 deg, external rotation: 29.1 deg)

Fig. 2. Knee joint configurations before and after external rotation and abduction of the tibia relative to the femur as simulated in the finite element analysis.

Fig. 3. Comparison between the FEA-simulated and the experimentally measured results on impingement force and contact area: The total impingement force and contact area were plotted in relation to the tibial external rotation angle and abduction angle with the knee flexion at about 44.8 deg.

Fig. 4.

(a) The contact pressure (in MPa) between the ACL and the femoral intercondylar notch on the impinging side of the ligament. (b) The distribution of the maximum principal stress in the ACL matrix and embedded fibers (in MPa). Up to 3.0 MPa and 8.9 MPa stress was observed in the matrix and the fibers on the non-impinging side, respectively.