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Femoral enthesal shape and attachment angle as potential risk factors for anterior cruciate ligament injury

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Abstract

Although non-contact human ACL tears are a common knee injury, little is known about why they usually fail near the femoral enthesis. Recent histological studies have identified a range of characteristic femoral enthesis tidemark profiles and ligament attachment angles. We tested the effect of the tidemark profile and attachment angle on the distribution of strain across the enthesis, under a ligament stretch of 1.1. We employed a 2D analytical model followed by 3D finite element models using three constitutive forms and solved with ABAQUS/Standard. The results show that the maximum equivalent strain was located in the most distal region of the ACL femoral enthesis. It is noteworthy that this strain was markedly increased by a concave (with respect to bone) enthesal profile in that region as well as by a smaller attachment angle, both of which are features more commonly found in females. Although the magnitude of the maximum equivalent strain predicted was not consistent among the constitutive models used, it did not affect the relationship observed between enthesal shape and maximum equivalent strain. We conclude that a concave tidemark profile and acute attachment angle at the femoral ACL enthesis increase the risk for ACL failure, and that failure is most likely to begin in the most distal region of that enthesis.

Keywords

ACL; injury; risk factor; enthesis; attachment angle; mechanics

1. Introduction

Anterior cruciate ligament (ACL) tears are the most common knee ligament injury, occurring more than 250,000 times per year in the United States [9, 29]. Complete tears of the ACL often require surgical reconstruction and increase the susceptibility to knee osteoarthritis within 10 years of the injury [19, 16]. These injuries are especially common in

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female athletes, who are two to five times more likely to sustain an ACL tear than their male counterparts [31, 13].

There has been considerable interest in determining anatomical features that increase an athlete's risk of ACL injury. Several morphological characteristics have been correlated with ACL injury, such as steeper posterior tibial slope in the lateral tibial plateau [28, 10, 6, 30] and smaller intercondylar notch width [28, 30, 32]. It has also been proposed that smaller cross-sectional area of the ACL is to blame for the increased injury rate of females compared to males [8, 1]. However, to the authors' knowledge, no statistically significant correlation has been found between cross-sectional area and injury risk. Additionally, while these correlations might prove useful, they lack a mechanical analysis that supports their direct causation of ACL injury.

Clinically, the most common location for an ACL tear is at or near the femoral insertion [33]. The reasons for this region's susceptibility are not yet fully understood. Nevertheless, *in vitro* experimental studies have demonstrated that the ACL is particularly prone to failure at the femoral enthesis, especially in the posterolateral (PL) bundle [18, 4, 24].

Recently, Beaulieu et al. [5] identified six main categories of human femoral entheses by the shape of their tidemarks on standardized histological sections (see Fig. 1). Beaulieu et al. [3] also quantified the angle of attachment of the ACL as it arises from lateral femoral epicondyle. The data from that study indicated that, at 15° of knee flexion, male specimens, on average, have a larger attachment angle than their female counterparts; the average male attachment angle was roughly 13° while the average female attachment angle was just 7°. At the present time, the extent to which the femoral enthesal shape and attachment angle affect ACL stress and strain concentration are unknown.

Therefore, the goals of this study were: (1) to use data from the histological studies performed by Beaulieu and colleagues to inform the development of biomechanical models of the ACL femoral attachment, and (2) to examine the differences in strain distribution among the characteristic tidemark profiles in order to determine whether particular profiles may be more prone to injury than others. A simplified 2D analytical model was constructed, followed by a 3D finite model with similar geometry. Three constitutive models were fit to longitudinal and transverse tensile test data from the literature. Results from all models suggest that a concave enthesis and smaller (more acute) attachment angle increase the strain concentration near the distal edge of the femoral ACL attachment, increasing injury risk. Additional analysis demonstrates that the macroscopic force-extension relationship of the structure is dependent both on the enthesis geometry as well as the constitutive form.

2. Methods

2.1. Analytical model formulation

The ACL femoral enthesis was first modeled as a 2D trapezoidal body of width w rigidly attached to a fixed curve, $y = A(x)$. This curve characterized the enthesal shape at the junction of the calcified and uncalcified fibrocartilage, and it had a mean slope of a/w , such that the insertion angle of the enthesis (ϕ) was 13° or 7°, the average attachment angle for

males and females, respectively. Enthesal profiles were constructed from histological slices following the grouping scheme shown in Fig. 1. The opposite edge of the body ($y = L$) represented the ligament proper, which was assumed to undergo a uniform displacement δ . Fig. 2 depicts the variables used in creating the model.

Assuming homogeneity and no Poisson effect, the displacement field in the ligament can be approximated by

$$u_y(x, y) = \frac{\delta}{L - A(x)}(y - A(x)). \quad (1)$$

From the representative histological slices shown in Fig. 1, analytic expressions for enthesal profiles, $y = A(x)$, were constructed. Each of the enthesal profiles, described by $y = A(x)$, was assumed to have one of the following forms based on its corresponding category in Fig. 1:

$$A_{A,B}(x) = \frac{a}{w}x \pm (x^2 - wx), \quad (2)$$

$$A_{C,D}(x) = \frac{a}{w}x \pm (2x^2 - 3wx^2 + w^2x), \quad (3)$$

or

$$A_{E,F}(x) = \frac{a}{w}x \pm (4x^4 - 8wx^3 + 5w^2x^2 - w^3x). \quad (4)$$

The nonlinear terms in parentheses were added to give concave enthesal shapes and subtracted to yield convex shapes (with respect to the bone). Additionally, a linear enthesal profile,

$$A_{linear}(x) = \frac{a}{w}x, \quad (5)$$

was considered. Analytic forms for the Lagrange strain tensor and equivalent (von Mises) strain were found using the deformation gradient tensor for each profile's unique displacement field:

$$\mathbf{F} = \begin{bmatrix} 1 & 0 \\ \frac{\partial u_y}{\partial x} & \left(\frac{\partial u_y}{\partial y} + 1 \right) \end{bmatrix} \quad (6)$$

$$\mathbf{E} = \frac{1}{2}(\mathbf{F}^T \mathbf{F} - \mathbf{I}) \quad (7)$$

$$\mathbf{E}' = \mathbf{E} - \frac{1}{3}Tr(\mathbf{E})\mathbf{I} \quad (8)$$

$$E_{eq} = \sqrt{\frac{2}{3}\mathbf{E}':\mathbf{E}'} \quad (9)$$

where $\mathbf{F}$ is the deformation gradient, $\mathbf{E}$ is the Lagrange strain, $\mathbf{E}'$ is the deviatoric part of $\mathbf{E}$, and E_{eq} is the equivalent strain.

Because this study seeks to gain understanding about which enthesal shapes may be more prone to injury, a failure criterion must be considered. Equivalent strain is not an accurate failure criterion for anisotropic materials such as ligaments; however, no such failure criterion currently exists. Furthermore, Luetkemeyer et al. [20] showed that while the two ACL bundles have different longitudinal stress-strain relationships when stretched in the mean fiber direction, their equivalent strains are remarkably consistent. This suggests that the ligaments may deform in ways that minimize their total deviatoric strain energy, and that there may exist some suitable failure criterion similar to the von Mises criterion. Regardless of what that failure criterion is, it seems likely that shear strain is part of it, and thus, that injury is likely the result of both high longitudinal and shear strains. Therefore, equivalent strain is used in this study to approximate the injurious potential that the combined longitudinal and shear loading present in unidirectional loading of a ligament attached to bone at an acute angle have on the ligament.

2.2. Finite element model formulation

A 3D finite element (FE) model representing a single ACL bundle was also considered to supplement the findings of the 2D analytical model. The FE model was generated and analyzed in ABAQUS v6.14 (SIMULA, Providence, Rhode Island, United States), a commercial finite element solver. The ACL models, shown in Fig. 3, each have a diameter of 10 mm and net attachment angle of 13°. In agreement with the 2D formulations, convex, linear, and concave ACL enthesal geometries were constructed (shown in Figs. 3a-c, respectively).

Three constitutive models were implemented for each ACL geometry. First, in an effort to directly extend the 2D analyses to this idealized geometry, an isotropic (nonlinear) neo-Hookean material description was used. Then, to more accurately describe the material mechanics of ligaments, two transversely isotropic constitutive models were considered: the single fiber family Holzapfel-Gasser-Odgen (HGO) model [14] and a transversely isotropic form of a freely-jointed eight-chain model (FJC) [7]. In models with material directionality,

the preferred material direction—or fiber direction—was oriented in a Cartesian sense along the direction of prescribed displacement (Fig. 3e).

Material parameters were determined based on the optimization framework presented in Marchi et al. [22]. The mechanical behavior of the ACL along its preferred material direction was assumed to be an average of uniaxial responses from each of its constituent bundles [23]. Due to the absence of data describing the mechanical behavior of the ACL normal to its preferred material direction, the transverse behavior of the ACL was assumed to be identical to that of the medial collateral ligament (MCL). Problematically, there is large variability in the observed macroscopic response of the MCL; therefore, the transverse response of the ACL was modeled with data from both Quapp and Weiss [26] and Henninger et al. [12]. Neo-Hookean ACL models were assumed to have the same material parameters as the matrix (isotropic) phase of the HGO model. Experimental data, both along and normal to the preferred material direction, and constitutive model best fits are shown in Fig. 4. The large variation in transverse data can be seen by comparing the magnitudes of data in Fig. 4b and Fig. 4c. All material properties were implemented using user-defined subroutines [22].

Each FE geometry was subjected to an applied macroscopic axial stretch of 1.1. This value was chosen because it is far from small strain but within the physiologic range, in addition to being less than half the failure strain, as measured grip-to-grip [8]. The entheses were completely constrained, while displacements were applied to the surface located on the top of each model in Fig. 3. During the displacement step, nodes on the displacement surface were kinematically coupled with respect to a local cylindrical coordinate system attached to a point located in the center of the surface; displacement boundary conditions were applied directly to this central point. This allowed for axial displacements to be prescribed, while simultaneously leaving radial displacements free. Each ACL model shown in Fig. 3a-c contained 70,350 fully-integrated hexahedral elements and was solved implicitly using ABAQUS/Standard. The maximum longitudinal and equivalent strains were evaluated for each geometry.

2.3. Macroscopic stress-stretch simulation

The material models implemented in the FE models were built assuming the datasets used were the result of uniaxial tension experiments. However, recent experiments have made full-field deformation measurements of the ACL bundles under tension, and both full-volume [20] and full-surface [21] strain fields show that homogeneous uniaxial tension was not achieved, as both shear and transverse strains were large. The extent to which this assumption affects the construction of the material model is currently unknown. Thus, as a secondary aim, the macroscopic stress-stretch relationship of each FE model was compared to the experimental data to which the constitutive models were fit. Little difference between the FE model predictions and the experimental data would suggest that standard uniaxial tension assumption is a reasonable approximation.

3. Results

3.1. 2D analytic approach

Compared to the linear and concave shapes, the parabolic convex (Fig. 1A) entheses profile increased longitudinal strain throughout most of the ligament, but simultaneously reduced shear strain (Fig. 5h-n), thus reducing the equivalent strain (Fig. 5o-u). Both the maximum longitudinal strain and maximum shear strain were found on or near the most distal margin of the profiles; however, the magnitude of the maximum shear strain varies greatly with entheses shape and attachment angle (see Figs. 5 and 6). Thus, both of these geometrical features were found to have significant effects on the maximum equivalent strain.

The most critical feature of entheses shape with respect to maximum equivalent strain was the convexity near the most distal attachment point of the ACL. If this part of the profile is convex (as in Figs. 1A, 1C, and 1E), equivalent strain is reduced. In contrast, concavity in this region (as in Figs. 1B, 1D, and 1F) increased the maximum equivalent strain. Maximum equivalent strain moved toward that of the linear entheses shape with increasing polynomial order (Fig. 6).

3.2. 3D FE approach

Similar differences between ACL attachment geometries were observed with the 3D FE models. Maximum tensile and equivalent strains (calculated using Eq. 9) for each ACL entheses geometry and material model combination are shown in Fig. 7. In all cases, for a given material model, the maximum equivalent strains were highest and lowest in ACLs with concave and convex entheses, respectively (Figs. 7b and 7d). Deformation in neo-Hookean ACL models manifested similarly for equivalent entheses shapes independent of the experimental data used in their construction. A similar trend was not observed when directional material models were used. While maximum equivalent strains all followed the same pattern (concave > linear > convex), the differences between maximum equivalent strain values notably varied with material model and experimental data. In particular, large differences in the maximum equivalent strains of convex vs. linear vs. concave entheses shapes were observed with the FJC model assuming transverse data of the form presented by Henninger et al. [12] (Fig. 7d); however, only small differences between maximum equivalent strains were observed with transverse data from Quapp and Weiss [26] (Fig. 7b). There was no consistent trend in the effect of entheses geometry on maximum tensile strains across material model forms or the experimental data used.

3.3. Macroscopic stress-stretch analysis

As a secondary aim, the validity of the assumption of homogeneous uniaxial tension for constitutive modeling purposes was assessed. If uniaxial tension is a valid assumption, the macroscopic stress-stretch relationship of the FE model (an idealized version of the experimental geometry) should match the experimental data used in the construction of the implemented material model. Conversely, the constitutive form, transverse experimental data, and entheses geometry all affected the macroscopic stress-stretch behavior of the idealized ACL. The stress-stretch responses for each transversely isotropic ACL model along the preferred material direction are shown in Fig. 8. There was large variation in

predicted macroscopic behavior for a given set of transverse experimental data, as well as with the assumed constitutive form. For each transverse dataset (data from either Quapp and Weiss [26] or Henninger et al. [12]), the response of FJC models was stiffer compared to HGO models—comparing (a) to (b) and (c) to (d) in Fig. 8. Models built assuming transverse data of the form presented in Quapp and Weiss [26] were stiffer compared to their equivalent Henninger et al. [12] based model; this feature can be seen by comparing (a) to (b) (HGO model) and (b) to (d) (FJC model) in Fig. 8. Only FJC based models with transverse data from Quapp and Weiss [26] approached the expected macroscopic preferred material direction response (Fig. 8). Small variation between enthesal geometries was also detected. For every material model and transverse data combination except the FJC model with Quapp and Weiss [26] transverse data, convex enthesal attachments produced the stiffest macroscopic stress-stretch relationship. In the case of the FJC model with Quapp and Weiss [26] transverse data, the concave enthesal yielded the stiffest response (Fig. 8).

4. Discussion

The present results lend insight into why ACL failures often occur at the femoral enthesal. The acute attachment angle of the ACL at the femoral enthesal prompts a concentration in shear strain near the most distal point of attachment. This strain concentration is exacerbated by a concavity in the tidemark in that region, but mitigated by convexity, because concavity locally minimizes while convexity locally maximizes the attachment angle in the distal margin of the femoral enthesal.

While the shear strain in the ACL is sensitive to the convexity of the profile, attachment angle is more influential in the concentration of shear and equivalent strain. The increase in strain with decreasing insertion angle may explain the ACL injury risk associated with small notch width [28, 30, 32], as smaller notch width should require a more acute attachment angle. If so, this provides a mechanistic understanding of the correlation found between notch width and injury risk.

Entesal shape and insertion angle may help explain the gender disparity seen in ACL injuries. In Beaulieu et al. [5], the mean femoral enthesal attachment angle of female specimens was found to be nearly half that of their male counterparts. A second look at the study's raw data showed that more than 81% of the male tidemark sections were convex near the distal corner of the femoral enthesal, while only about 57% of female enthesal sections exhibited one of these more advantageous shapes [5].

These results may also explain why the PL bundle of the ACL is more susceptible to fatigue failure than the anteromedial (AM) bundle [18, 4]. In the histological study by Beaulieu et al. [5], all parabolic concave profiles were seen in the posterior sections while all but one of the convex profiles were found in anterior sections. Because the PL bundle femoral attachment site is more posterior and distal than the AM bundle attachment, this suggests that the PL bundle may possess a less advantageous concave profile as well as be located in the region of highest shear and equivalent strain concentration.

While other anatomical risk factors are largely non-modifiable, it is possible that the shape of the enthesis may be able to change in response to particular types of loading. Milella et al. [25] found that enthesal robusticity scores were significantly greater in right versus left upper limb entheses, indicating significant right hand dominance. Although debated, enthesal morphology has long been used as a marker of activity level to study ancient populations in bioarcheology [27]. The fact that almost 70% of all profiles categorized by Beaulieu et al. [5] had a convex shape in the inferior margin suggests that enthesal convexity is favored over concavity with respect to the bone in that region.

Worthy of further investigation is the notable dependence of the maximum equivalent strain values predicted on the constitutive model form and transverse dataset used, as shown in Fig. 7. Most computational biomechanics studies on ligaments have ignored the material response orthogonal to the fiber direction. This study used both of the two sets of transverse ligament mechanical data published to date. Even when the same constitutive form and axial response dataset was used, the two model fits provided disparate results. This shows that the transverse and/or shear response of ligaments is a necessary consideration for accurate constitutive modeling of ligament tissue.

When the same datasets were used in the construction of the two transversely isotropic constitutive models considered, they produced dissimilar results as well. This is not necessarily unexpected, as differences in aspects of the model physics, like shear coupling, would lead to differences in the predicted equivalent strain fields. Future ligament computational studies may want to consider more than one constitutive form in their model, since the sparsity of experimental data in multiple deformation states makes it difficult to know which material models can accurately describe the real material physics. Moreover, the lack of data in several deformation states means it is impossible to ensure that all model parameters are identifiable.

Also pertinent to this discussion is the wide array of macroscopic stress values predicted for a given macroscopic stretch illustrated in Fig. 8. This applies both to different constitutive models fit to the same datasets as well as to the same constitutive model fit with different transverse data. This has implications not only in modeling, but also in experimental characterization. The fact that these global stress-stretch curves do not align with the curve to which the constitutive models were fit implies that boundary effects prevent the computational models from achieving a state of uniaxial tension, which was assumed for material model fitting. This indicates that current ligament constitutive models are actually a description of both the structural stiffness and material stiffness. FE models require constitutive models that describe the true material response; thus, material and structural properties need to be decoupled. This is principally a result of the large characteristic decay length (the length over which boundary effects are significant) associated with highly transversely isotropic materials [2, 15]. Still, traditionally, as well as in the current study, ligament constitutive modeling has not yet accounted for the effects that specimen geometry and boundary conditions have on the measured macroscopic mechanical behavior. For this reason, we recently measured full-volume displacement fields for the AM and PL bundles of the ovine ACL [20]. Full-field inverse methods will be used to fit constitutive models to this data, accounting for the irregular geometry and resulting strain field inhomogeneity.

Similarly, it is commonly assumed that the mechanical response of ligaments transverse to the fibers makes little difference in the predicted response of constitutive models of ligaments. As Fig. 8 reinforces, both constitutive form and transverse mechanical behavior play invaluable roles in the construction of computational models of the ACL.

Limitations of this study include the use of the von Mises criterion for an anisotropic material, simplified constitutive modeling assumptions, and the possibility that a double bundle model would have been a more accurate representation of the ACL than the single bundle models used. Additionally, the femoral attachment angles used to create the 2D and 3D models were based on data collected at 15° knee flexion. However, it should be noted that this is a common knee flexion angle at which ACL injuries occur [17]. Finally, the cross-sectional shape of the ACL was idealized as a circle and the applied deformation did not include bending, torsion, or displacement in any direction other than axial. More complex loading may yield different results.

Despite the large differences in maximum strains predicted by the various models used, the trend seen in maximum equivalent strain was consistent. A concave enthesal shape increased the maximum equivalent strain seen by the ACL in every model used, and thus, would appear to be more prone to injury than other enthesal shapes. Future research on ACL injury risk factors should consider enthesal shape and insertion angle.

5. Conclusions

The primary aim of this work was to model the femoral attachment of the ACL and determine whether some enthesal shapes, as identified by Beaulieu et al. [5], may cause the ACL to be more prone to injury than others, as predicted by the maximum equivalent strain. Based on all of the models tested, a more acute attachment angle and bone concavity in the distal margin of the enthesis are likely more prone to cause ligament failure than larger attachment angles and convex enthesal shapes. This analysis illustrates the effects of two of the variables involved in predicting the mechanical failure of the ACL.

Additionally, this work used the formulated FE models to examine the validity of traditional constitutive modeling assumptions. It was determined that homogeneous uniaxial tension is not a valid assumption for accurate constitutive modeling, and that the transverse response or matrix phase of the constitutive model has a significant impact on both the strain fields and the macroscopic response.

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Figure 1:

Human ACL femoral enthesal profile categories include second order convex (A), second order concave (B), third order convex (C), third order concave (D), fourth order convex (E) and fourth order concave (F) polynomial fits. Note that the shapes are classified as convex or concave with respect to the bone in the distal half. In each panel, the proximal end of the enthesis is shown at left, while the distal end is shown at right. Reproduced from Fig. 3 in Beaulieu et al. [5], used under CC BY 4.0. Lower two rows of panels have been exchanged.

Figure 2:

Diagram depicting the relationships among variables used in the 2D model over an image of an ovine ACL. The red arrow indicates the direction of the displacement (δ) of the boundary $y = L$, which was 10% of L . The red line represents $A(x)$, the femoral enthesis boundary, described by polynomials.

Figure 3:

3D FE models corresponding to convex (a), concave (b), and linear (c,d) enthesal shape profiles. Straight, dotted red lines are drawn between the most proximal (left) and distal (right) points of the entheses to highlight the concavity/convexity. The boundary conditions are shown in (d). Elements on the femoral enthesis boundary were encastered and elements on the opposite edge were assigned a uniform axial displacement.

Figure 4:

Experimental stress-stretch data and constitutive model fits (a) along and (b,c) normal to the preferred material direction. Experimental datasets are from (a) McLean et al. [23], (b) Quapp and Weiss [26] and (c) Henninger et al. [12]. Solid and dashed lines denote constitutive model best fits assuming transverse experimental stress-stretch data from [26] and [12], respectively.

Figure 5:

2D (a-g) Longitudinal displacement, (h-n) in-plane shear strain, and (o-u) equivalent strain fields for each entheseal profile. All contours computed with the average male (13°) attachment angle. Hatched regions indicate bone. These results demonstrate that the equivalent strain is largest near the distal (right) end of the enthesis.

Figure 6:

2D maximum (a) shear and (b) equivalent strains of various enthesal profiles and attachment angles. * denotes an enthesal profile for which the maximum strain value occurs at a location other than the point $\{w, a\}$ (see Fig. 2).

Figure 8:

Predicted macroscopic stress-stretch curves of various geometric and constitutive models.

Simulation results using constitutive model best fits assuming transverse experimental stress-stretch data from Quapp and Weiss [26] and Henninger et al. [12] are shown in (a,b) and (c,d), respectively.

Table 1:

Best fit constitutive parameters of various ACL material models

neo-Hookean	μ (MPa)	B (MPa)		
Quapp and Weiss [26]	4.05	100		
Henninger et al. [11, 12]	0.0736	100		
HGO	μ (MPa)	B (MPa)	k_1 (MPa)	k_2
Quapp and Weiss [26]	4.05	100	3.58	44.2
Henninger et al. [11, 12]	0.0736	100	5.84	33.1
oFJC	C_r (MPa)	B (MPa)	a	$b = c$
Quapp and Weiss [26]	0.0806	100	1.25	1.15
Henninger et al. [11, 12]	0.333	100	2.21	0.325