Elastographic Imaging of Strain Distribution within the Anterior Cruciate Ligament and at the ACL-Bone Insertions

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Abstract—The anterior cruciate ligament (ACL) functions as a mechanical stabilizer in the tibiofemoral joint. Over 250,000 Americans each year suffer from ACL ruptures and tears, making the ACL the most commonly injured knee ligament. A long term goal of our research program is to promote graft-bone integration via the regeneration of the native ligament-bone interface. To this end, we have focused on the design of biomimetic scaffolds combined with tissue engineering to induce organized interface regeneration. An understanding of mechanical properties of the ligament-bone interface is critical for biomimetic scaffold design and clinical evaluation. To date, experimental determination has been difficult due to the small length scale (< 1 mm) involved at the ACL insertions. This study utilizes ultrasound elastography to characterize the functional properties of the ACL and the ACL-bone interface under applied loading. Specifically, the tibiofemoral joints were mounted on a biomechanics material testing system and loaded in tension while RF data was collected. Axial elastograms between successive RF frames were generated using cross-correlation and recorrelation techniques. Elastography analyses revealed that when the joint is in tension, complex strains with both compressive and tensile components were found at the tibial insertion. These results are in agreement with those of prior finite element analysis (FEA) model predictions. In addition, the magnitude of displacement was found to be the highest at the ACL proper and decreased in value from ligament to bone. Our results indicate that elastography is a novel and useful method in understanding the mechanical properties of the ligament itself and the ligament-bone interface.

I. INTRODUCTION

The anterior cruciate ligament (ACL) is the primary joint stabilizer and the most frequently injured ligament of the knee [1], with over 75,000 reconstruction procedures performed annually in the United States [2]. Currently, the widely used semitendinosus or hamstring tendon-based ACL reconstruction grafts are mechanically anchored, and their clinical success is limited by the lack of biological graft integration with the subchondral bone. The native ACL inserts into bone through a fibrocartilage interface [3], which functions to minimize stress concentrations and facilitate load transfer between two distinct types of tissue. This interface, however, is not re-established after tendon graft-based ACL reconstruction. Without a stable interface, the graft-bone insertion site represents the weakest point mechanically during the early healing period [4-6].

Our long term research objective is to promote graft-bone integration via the regeneration of the native ligament-bone interface. The success of this approach will depend on an understanding of the mechanical properties of both the ACL and the ACL-bone interface. This understanding is critical in the design of biomimetic scaffolds with controlled structural and mechanical properties emulating those of the insertion zones [7]. While the mechanical properties of the ACL mid-substance have been investigated extensively [8,9], those of the ACL insertion are not well understood. The strain distribution at the ACL-bone junction is not known. One of the major challenges associated with measuring the mechanical properties of the ACL-bone interface has been the physical scale of the interface, which spans only 100-300 µm in humans [3].

The objective of this study is to determine the feasibility of utilizing elasticity imaging techniques towards mechanical characterization of the ACL and ACL-bone interface. Elasticity imaging, or methods for the mapping of mechanical responses or properties using ultrasound or MRI images acquired before and after a mechanical excitation, was initially developed in the early 1990s as an alternative to palpation [10,11]. It has since then been utilized in intravascular and cardiovascular applications in vivo, as well as in the guidance of thermal therapy procedures and robotic surgery.

One of the most interesting aspects of recent elastography applications is in the area of tissue biomechanics. The complexities of tissue mechanics (both elastic and viscoelastic properties) can be measured using this novel method [12]. Time-dependent changes in mechanical properties can be monitored in real-time and characterized. The majority of the reported elastographic studies and related theoretical models of articular joint tissues have so far focused on articular cartilage [13-15]. For example, poroelastography [13], or elastographic imaging of poroelastic tissues, has been used to estimate and image the in-plane strain ratio of the simulated cartilage during a sustained compression. The technique is capable of identifying and depicting the permeable interfaces of the tissue, the permeability of the solid to the fluid (or the type of fluid contained in the solid matrix) and the Poisson’s ratio...
of the solid matrix at equilibrium, i.e., at no further variation of the strain ratio. The theoretical model has been verified in phantoms [16-17] while similar in vitro tissue results have been reported [14-15].

Ultrasound Elastography thus represents a novel, practical and reliable technique for the characterization of the mechanical behavior of the ACL and the insertion sites. In this study, we explore the use of ultrasound elastography to determine strain distributions at the ACL proper and at the insertions (Fig. 1a). We will also image the mechanical response of the ACL and ACL-bone insertion region under applied tensile loading (Fig. 1b). Specifically, the incremental displacement and strain distributions at the ligament and between ligament and bone will be determined at several strain rates.

II. METHODS AND RESULTS

Elastography of bovine ACL and insertions in vitro during applied tensile loading

The femur-ACL-tibia complex (FATC) was mounted on an MTS 858 Bionix Testing System (Fig. 1b). The femur and tibia were aligned along the tensile axis and the sample was submerged in saline during testing. The FATC was loaded at different strain rates and tested to failure while RF data was collected at 5 MHz using a Terason ultrasound scanner (Teratech, Inc) and processed offline. Two-dimensional RF data using a 3 MHz linear array were acquired continuously during the tensile loading for a period of 3 sec. Sequences of ultrasound RF images were acquired at 54 frames/s (128 RF lines, sampling frequency at 10 MHz). The axial displacement between successive frames was estimated and imaged using cross-correlation and recorrelation techniques [19] with a window size of 3 mm and a window overlap of 75%.

The elastographic convention for the sign of the strain estimate imaged is kept here, i.e., positive strain denotes compression while negative strain denotes tension in the axial direction. Axial displacements are estimated from two consecutive RF frames, using a 1D cross-correlation algorithm [11,19]. In this algorithm, time-shifts between two consecutive backscattered signals are determined by the cross-correlation of small sliding windows over the entire 2D ultrasound image. At high decorrelation noise, recorrelation techniques were employed [19]. Finally, the strain distribution is computed by differentiating the displacement map along the axial direction. For the numerical differentiation, a least-squares regression method was used.

In all specimens tested, the ACL and the interface between the ACL and the femoral or tibial bone were clearly identifiable on the ultrasound images (Fig. 2), as well as in the displacement maps and elastograms (Figs. 3, 5, 6). In addition, within the ACL bundles, a narrow band of high strain in the middle and along the length of ACL was noted that also corresponded to a highly echogenic area on the B-scan images. This may reflect the parallel bundle-organization of the ACL.

Two different cases of displacement and strain estimation were studied: (a) using successively acquired ultrasound data (incremental) and (b) using a reference ultrasound frame for the strain estimation (non-incremental). Displacement and strain in (a) were estimated between successive RF data frames and with a period of 21 msec. The purpose was to obtain the temporal profile of the motion and deformation variation at the ligament and the insertion. Displacement and strain in (b) were estimated relative to the reference frame, e.g., at the beginning of the tensile stretching. The objective was to obtain a map of the cumulative deformation at the ligament and the insertion. In both cases, yellow-red areas of the elastograms correspond to compressive strain.

![Figure 1](image1.png)

**Figure 1:** a) The anatomy of a human knee joint (anterior view) [18] and b) its positioning in the MTS device for testing (posterior view).

![Figure 2](image2.png)

**Figure 2:** Sonogram of ACL and insertions (posterior scan), with corresponding displacement images and elastograms in Figure 3. The ACL is highly hypoechoic and its entire length is not clearly seen on the sonogram.

It was found that when the joint was tested in tension in the tibial orientation, the largest displacement occurred in the ACL proper and at the femoral insertion (Fig. 3a). In addition, the elastographic analysis revealed that the strain profile at the tibial insertion was highly complex, with both compressive and tensile strains found at the tibial insertion.
site, indicated by the yellow-red and green-blue regions, respectively, on the elastograms (Fig. 3b). Compressive strains were found in the ACL itself, most likely because in this experiment, the ultrasound transducer was aligned with respect to the insertion (region of interest) during loading.

The distribution of tensile and compressive strain varied as a function of strain rate. With increasing strain, the temporal variation in stress revealed higher displacement and a corresponding increase in compressive strain at the ACL-bone interface (Fig. 4). In addition, the magnitude of displacement was found to be the highest within the ACL proper and decreased in value in a gradual transition from ligament to bone (Fig. 5). Results from an additional sample are shown in Fig. 6. In this case, deformation in the sample was again induced with applied tensile loading, but the images were acquired at a different configuration with respect to the sample in Fig. 5. Here, the transducer was aligned with respect to the principal direction of the ACL. It is important to note that in this sample, compressive strain (red region, Fig. 6b) is again observed at the interface while the joint is loaded in tension. In addition, tensile strains (blue regions) are found at the same interface. These results are consistent with those presented in Fig. 4 and they collectively demonstrate the complexity of the strain profile at the tibial insertion.

III. DISCUSSION

This study represents the first reported effort at applying ultrasound elastography to determine the mechanical properties of the ACL and ACL-bone interface. The results of this study demonstrate the feasibility of applying elasticity imaging for the in vitro mechanical characterization of the bovine ACL and ACL-bone interface. The data presented here is the first experimental determination of the complex strain distribution at the insertion sites.

Strain elastograms indicate that the strain distribution at the insertions is highly complex, with both tensile and compressive strain components localized at the tibial insertion site. The presence of both tensile and compressive strain found at the ACL insertion is in agreement with finite-element analysis performed by Matyas et al., which showed...
that when the medial collateral ligament (MCL) is loaded in tension, the principal stress component at the femoral insertion is compressive [20]. While the angle of insertion differs between MCL and ACL, fibrocartilage tissue is the dominant tissue type seen at the ACL-bone and MCL-bone interfaces. Moreover, fibrocartilage tissue is generally found in regions where compressive loading is dominant. It is possible that collagen fibers in tension may transmit shear and compressive stresses through the fibrocartilage interface between the ACL and bone [3].

In this study, the largest displacement occurred in the ACL proper and at the femoral insertion, which may offer an explanation for the clinical observation that injuries to the ACL occur more frequently at the ligament proper and at the femoral insertion site. The magnitude of displacement was found to be the highest at the ACL proper and decreased in value from ligament to bone, which suggest a tissue-type-dependent increase in stiffness progressing from ligament to interface and then to bone.

Future studies will focus on validation of the elastographic findings through independent mechanical measurements and the use of higher resolution scanners in order to conduct in-depth evaluation of the zonal-dependent changes in mechanical properties found at the ACL-to-bone insertion. It is believed that this highly organized multi-tissue transition permits the transfer of mechanical loads between soft tissue and bone. Results of this study will assist in our efforts to understand the structure-function relationship inherent at the ACL-bone interface, and will be important in future studies for the in situ monitoring of graft mechanical properties during healing.

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V. REFERENCES