Three-dimensional finite element analysis of the human ACL

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Abstract: - The anterior cruciate ligament (ACL) is the most commonly injured ligament of the body, especially during sport and vehicle accidents, and therefore the biomechanics of the ACL is of interest. Some different kinds of surgery for its relief and reconstruction have been introduced. Diversity of the surgical procedures and therefore selection of appropriate graft, comprise vast topics in orthopedics surgeries. In this study, we present a three-dimensional FE model of the healthy human anterior cruciate ligament in passive full extension. ACL was considered incompressible, isotropic and hyperelastic. Initial strain on the ligament was also considered. Our main goal was to assess the stress distribution within the ligament. This model was validated and compared with numerical results obtained by other authors. The results show with applying the physiological initial strain and external anterior and posterior loads to femur and tibia, the maximal principal stress becomes much higher especially located in the tibial and femoral insertion sites respectively.

Key-Words: - ACL, 3-D finite element model, Incompressible, Hyperelastic, Isotropic, Stress distribution

1 Introduction
Knee ligaments provide stability and restrain knee motion in more than one degree of freedom, while the overall joint stability depends on the contributions of the individual ligaments and their interactions. The fully understanding of the role of each individual ligament in motion restraining is essential for the development of an adequate diagnostic and assessment on surgical procedures. A proper understanding of knee joint biomechanics is therefore essential to improve the prevention and treatment of its disorders and injuries [1]. The anterior cruciate ligament (ACL) limits the hyperextension of knee and forbids the slide of the femur to back on the smooth surface of the tibia [2]. ACL rupture is the widespread phenomenon in sport and vehicle accidents and since its relief and reconstruction for the patient who is usually an athlete, to desire situation and rehabilitation for doing sport have been considered for medics and athletic physic.

Direct measurement of the stress or strain distribution within the ACL is difficult and various techniques have been used in the past: strain gauges (Henning et al., 1985), displacements (Markolf et al., 1990; Renstrom et al., 1986), buckle transducers (Barry and Ahmed, 1986) or optical methods such as Roentgen stereophotogrammetry (Meijer et al., 1989; van Dijk et al., 1979). There are a number of limitations with the existing experimental methods. Experimental sensors tend to alter the natural geometry of the ligament and thus cause errors in the measurements. Also, in the majority of cases, the measurements are taken at discrete locations, rather than continuously over the entire surface of the ligament. The fibrous nature of ligaments means that deformation patterns in the ACL are not uniform and vary significantly according to the location of the measurement. Finally, various studies have tested the ACL in isolation in uniaxial tension, but such tests are not representative of the complex strain patterns that occur during flexion-extension (Amis and Dawkins, 1991) [3]. Although experimental studies measuring the mechanical properties of the ACL were extensive (Woo et al., 1983, 1991; Butler et al., 1986, 1992), the anisotropic mechanical properties of the ACL were unavailable because the complex structure of the ACL made the measurement of the mechanical properties of the ACL difficult [1]. In recent decades wide experimental studies have been done in vivo and in vitro to determine the quantity of forces and strains in ACL.

In general finite element analysis makes to analyze the 3D construction and understands the stress-strain distribution in each part of model. Validation is the most challenging aspect of the FE modeling of ligament mechanics, as it requires accurate
experimental measurements of quantities that are difficult to measure. Finally there are some limited 3D FE models of ACL [1-10].

2 Materials and methods

2.1 Generation of the FEM of the ACL

The acquisition of accurate geometry for the ligament(s) and possibly the bones is a fundamental requirement for the construction of three-dimensional FE models of ligaments. Both magnetic resonance imaging (MRI) and computed tomography (CT) have been used to acquire ligament geometry. MRI can provide detailed images of soft tissue structure in diarthrodial joints. CT provides excellent images of the bones around the joint. Soft tissue is visible in standard CT images, but there is little difference in the signal between soft tissues, and thus, it can be difficult to distinguish the boundaries of a specific ligament in an intact joint [11].

We take the MRI images from a young intact female knee in full extension and then we made an exact 3D model of ligament. The manual segmentation had an accuracy of 3mm distance and 0mm thickness of each segments with high resolution. In the second stage via the modeling software Mimics v10.0 the section and geometrical boundary of each segment had been identified and the 3D model reconstructed (Fig.1).

![Fig.1. 3D model of ACL](image)

2.2 Material Properties

Studies measuring material properties of the ligament are extensive. However, when the ligament is studied in isolation, the exact loading conditions and the ligament geometry in situ remain unknown. Mommersteeg et al. (1995) have shown that the ligament's structural properties vary considerably as a consequence of changes in the configuration of the ligament [12].

Ligaments are highly anisotropic due to their fibrous structure. The degree of anisotropy can vary substantially between different types of ligaments and the fiber orientation generally represents an adaptation to the mechanical environment. For instance, the collagen fibers in the cruciate ligaments of the knee are highly aligned with the long axis. Although often assumed to be incompressible due to their high water content, experimental evidence suggests ligaments undergo some volume change during deformations.

The loading and unloading curves of ligaments under tension do not follow the same path. Rather, a hysteresis loop is observed during cyclic tensile testing due to internal energy loss. Creep, an increase in deformation over time under a constant load, and stress relaxation, a reduction in stress over time under a constant deformation can both be observed in ligaments. The effects of conditions, such as temperature and hydration level on the viscoelastic behavior of ligaments, have also been investigated.

The variation of ligament stress–strain behavior with strain rate is another indicator of the viscoelastic nature of the tissue. Woo et al. (1990) compared the material properties of rabbit medial collateral ligaments (MCLs) tested at five different strain rates. Results showed that changes in strain rate of over four orders of magnitude had relatively small effects on ligament material properties. Tensile strength and ultimate strain increased slightly with increasing strain rate while tangent modulus remained essentially unchanged [11]. The ligaments undergo severe strain and finite deformation and thus it is necessary to treat them as hyperelastic material for analysis.

2.3 Constitutive law and mechanical properties

Constitutive equations are used to describe the mechanical behavior of ideal materials through specification of the dependence of stress on variables, such as the deformation gradient, rate of deformation, temperature, and pressure. The accurate description and prediction of the three-dimensional mechanical behavior of ligaments by constitutive equations remains one of the challenges for computational modeling. The development and application of these constitutive models relies on an understanding of ligament structure and function, and knowledge of available experimental data [11].
Ligaments are dense connective tissues consisting of mainly parallel-fibre collagenous tissues embedded in a highly compliant solid matrix of proteoglycans (Fung, 1981). The preferred orientation of the collagen fibres induces the transversely isotropic symmetry of the ligament. Due to their natural composite structure, ligaments can be described by a continuum theory of fibre-reinforced composites at finite strains.

The existence of a strain energy function \( C \) that depends on \( I_1, I_2 \) and \( I_3 \), the first three principal invariants of the right Cauchy-Green deformation tensor \( C \) is postulated. The strain invariants \( I_1, I_2 \) and \( I_3 \) are represented in terms of principle stretches \( \lambda_1, \lambda_2 \) and \( \lambda_3 \) of strain tensor.

\[
\begin{align*}
I_1 &= \lambda_1^2 + \lambda_2^2 + \lambda_3^2 \\
I_2 &= \lambda_1^2 \lambda_2^2 + \lambda_2^2 \lambda_3^2 + \lambda_3^2 \lambda_1^2 \\
I_3 &= \lambda_1^2 \lambda_2^2 \lambda_3^2 = J^2
\end{align*}
\]

(1)

It is assumed that the strain energy function characterizing the mechanical behavior of the ACL and this function represents the mechanical response of the ground substance \( \psi_m \). Where \( \psi_m \) is chosen as being a Neo-Hookean incompressible isotropic hyperelastic potential that is a simple extension of the classical linear isotropic elasticity for the finite strain regime.

The neo-Hookean model has been shown to represent well the elastic behavior of the ground substance of connective tissues (Weiss et al., 1996; Limbert, 2001). For an incompressible material, \( \psi_m \) also exhibits the property of convexity which assures stability of the material.

\[
\psi_m(I_1, I_2, I_3) = C_1(I_1 - 3) + \frac{1}{D}(J - 1)^2
\]

(2)

Where \( C_1 \) is the Neo-Hookean constant and equal 1.95, \( I_1 \) strain invariant and \( D \) the inverse of the bulk modulus \( k = 1/D \) which was chosen for all the ligaments as \( k/C_1 = 1000 \) [1].

### 2.4 Boundary conditions

The solid volume representing the ACL was meshed with 8-noded hexahedral elements using Ansys v11.0. Special care was taken in order to optimize the performance of the mesh for the large displacement and large strain analysis. This element assumed to have the hyperelasticity, large strain and simulation the deformation of incompressible hyperelastic properties. (Fig.2)

Ligaments were attached to bone by establishing the final row of elements at their proximal and distal ends to be composed of the same material than the nearby bone. We assume the nodes of the tibial and femoral insertion areas were considered as rigidly fixed. Only in the direction which applies the external loading, ligament can move.

Biological soft tissues are usually exposed to a complex distribution of ‘‘in vivo’’ residual stresses as a consequence of the continuous growth, remodeling, damage and viscoelastic strains that they suffer along their whole life. Selective cutting of the living tissue and removal of its internal constraints can relieve these stresses. Their most important aim is to homogenize the final stress distribution at different stages of tissue deformation [1].

The moderate (1% s^-1) strain rate was applied that corresponds to physiological strain rate. We can define this strain rate, as a pre-load which if we cut the ligament from each point its length will be shorter. The strain rate during injury is very important regarding the magnitude of the lesion. The strain rate during non-physiological loads is determinant in the risk of damage in ligaments. Non-physiological movements under low strain rates do not usually provoke ligament damage that under high strain rates occurs [10].

![Fig.2. Finite element model of the ACL](image)

### 2.5 FE Analysis

On modeling ligaments, two important assumptions were made. First, no difference in the material behavior between the ligament body and its insertion were considered. Second, material characteristics depending on time, such as viscoelasticity, creep and relaxation were neglected.

Using the boundary conditions described in Section 2.3 and the material properties described in Section 2.2, we assumed therefore an incompressible
isotropic hyperelastic model which four series of analyses were performed. A first FE analysis was carried out with the anterior tibial load 134N and the second FE analysis, using anterior femoral load 134N, the third by using posterior tibial load 134N and the fourth by using the posterior femoral load 134N at full knee extension. For analysis of this model we used Neo-Hookan equation of Ansys v11.0. to assess the maximum stress and stress distribution in the model.

3 Results

3.1 Anterior tibial load
Under the proposed anterior tibial load of 134N and initial strain 1% , as it is shown in Fig.3 the maximum principle stress located in the femoral insertion site. The first principle stress of 4.9 MPa, the second principle stress of 4.88 MPa and the third principle stress of 4.85 MPa obtained in the femoral insertion. As it is shown, the femoral insertion site is the highest stress distribution part of the ligament, which is similar to the results of literature [1].

3.2 Posterior tibial load
Under the proposed posterior tibial load of 134N and initial strain 1% , as it is shown in Fig.4 the maximum principle stress located in the femoral insertion site. The first principle stress of 7.52 MPa, the second principle stress of 7.43 MPa and the third principle stress of 7.36 MPa obtained in the femoral insertion. As it is shown, the femoral insertion site of the ligament is introduced as the highest stress distribution part of the ligament.

3.3 Anterior femoral load
Under the proposed anterior femoral load of 134N and initial strain 1% , as it is shown in Fig.5 the maximum principle stress located in the tibial insertion site. The first principle stress of 7.12 MPa, the second principle stress of 7.08 MPa and the third principle stress of 7.07 MPa obtained in the tibial insertion. As it is shown, the tibial insertion site is the highest stress distribution part of the ligament, which is similar to the results of literature [10].
3.4 Posterior femoral load
Under the proposed posterior femoral load of 134N and initial strain 1%, as it is shown in Fig.6 the maximum principle stress located in the tibial insertion site. The first principle stress of 7.8 MPa, the second principle stress of 7.9 MPa and the third principle stress of 7.7 MPa obtained in the tibial insertion. As it is shown, the tibial insertion site of ligament is introduced as the highest stress distribution part of ligament.

4 Conclusion
In this paper, we present a complete 3D model of the healthy human anterior cruciate ligament. The connection of ligament to femur and tibia are assumed rigidly fix. Ligament was modeled as incompressible hyperelastic isotropic material. Initial strain 1% in the ligament was considered, and then results were compared by those reported in the literature [1,10].

As it was shown the bone–ligament insertion sites are introduced as high stress distribution parts of ligament. By applying anterior/posterior tibial load, the femoral insertion site was introduced as the highest stress distribution site of anterior cruciate ligament and the stress in the middle of ligament was higher than that near the tibial insertion site. Clinical reports indicate that most ACL tears occur near the femoral insertion site and in the mid-substance (Sherman et al., 1991). The stress distribution within the ACL corresponds to those clinical reports and offers additional information for understanding the mechanism of ACL injury. Moreover, by applying anterior/posterior femoral load, the maximal principal stress becomes much higher especially located in the tibial insertion site of the ligament. It is well known that damage in mature ligaments occurs usually at high load rates; on the contrary, bone–ligament insertions are usually ruptured under lower load rates. It is necessary to mention that clinical wise, the bone-ligament insertion is identified as the most injured part of rupture.

With the methodology developed in this study, the FE 3D model of healthy ligament and the reconstructed ligament in one person in each leg will be performed in future to compare the stress distribution in 2 cases. Furthermore prepare the model of a complete knee include of the bones and ligaments to study the parameters that lead to defect of ACL.
References:


