Stress Distribution within the Anteromedial and Posterolateral Bundles of ACL under Anterior Tibial Load

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INTRODUCTION: The anterior cruciate ligament (ACL) plays an essential role in maintaining knee stability, and is one of the most commonly injured ligaments of the knee. Understanding the stress distribution within the anteromedial (AM) and posterolateral (PL) bundles of the human ACL will be helpful in designing reconstruction procedures, optimizing rehabilitation protocols and understanding the environment of biological remodeling. The objective of this study is to develop a three-dimensional finite element (FE) model of a human knee to calculate the stress distribution within the AM and PL bundles of the ACL in response to an anterior tibial load. This model was also validated by comparing the computational predictions with the *in situ* force in the ACL obtained experimentally using the robotic/UFS testing system under the same loading condition [1].

METHODS: Geometry of the femur and tibia of a human knee were reconstructed from MR images (Sigma, Horizon; GE, Milwaukee, WI). The insertion sites of the ACL were obtained using the Microscribe digitizing platform (Immersion Corp., San Jose, CA) [2], and based on the anatomy reported in previous studies [3]. MARC finite element software (MSC. Software Corporation, Los Angels, CA) was used to implement the 3-D FE model of the knee. The anteromedial (AM) and posterolateral (PL) bundles of the ACL were represented by an incompressible hyperelastic material, and assumed to be homogeneous and isotropic as the anisotropic, mechanical properties of the ACL were unavailable. The mechanical properties of the ACL in this study were taken from the literature [4]. The AM and PL bundles were assumed to have an initial *in situ* strain of 3% [5]. The contact and friction between the AM and PL bundles and the contact and friction between the ligaments and bone were also included in this study. The direct constraint method for the solution of contact problems and the Coulomb friction model for the solution of the friction problem were implemented in MARC software. The 3-D FE knee model (Figure 1) was analyzed with the knee at full extension, i.e. with the femur fixed in space and the tibia free to move in 5-DOF. The robotic/UFS testing system was used to obtain the experimental kinematics data of the knee and *in situ* force of the ACL under incremental anterior tibial loads (0-134N) applied to the intact knee at full extension. This kinematics data was used as input into the knee model to calculate the force of the ACL. Computational force of the ACL is the vector summation of the force of the AM, PL bundles and the contact force. The computational force of the ACL was then compared with the experimentally obtained *in situ* force of the ACL to validate the FE model, which then served as a tool to analyze the stress distribution within the ACL.

RESULTS: The computational *in situ* force of the ACL from the FE model under incremental anterior tibial load (0-134N) at full extension varied from the experimental results by 5.0 ± 5.4N (Figure 2). The difference between the computational and experimental *in situ* force of the ACL was less than 10 percent when the anterior tibial load was 134N. Under 134 N anterior tibial load, the computational force of the AM bundle is 51N and that of the PL bundle is 65N. The analytical results show that the stress distribution was non-uniform between the AM and PL bundles (Figure 3). For both the AM and PL bundles, the stress is mainly localized near the femoral insertion site. The maximum stress in the AM bundle is 35 MPa, and the maximum stress in the PL bundle is 28 MPa. The stress near the tibial insertion site was comparatively lower than that near the femoral insertion site.

DISCUSSION: Comparison of the *in situ* force in the ACL in response to an anterior tibial load as calculated by the ACL model with experimental results at full knee extension shows that the computational results agree well with the experimental data. The computational results indicate that the PL bundle endures more force than the AM bundle at full extension, confirming published experimental results [6]. Higher stresses near the femoral insertion site correspond to clinical reports that injury of the ACL occurs more often near the femoral insertion site. Although the ACL is modeled as an isotropic hyperelastic material in this study, a more realistic representation of the mechanical properties of the material that incorporate the characteristic that the ACL does not support significant compressive and bending loads in the direction of the collagen fibers will incorporated in the future. An improved knee model will allow for more accurate analysis of the stress distribution of the ACL.

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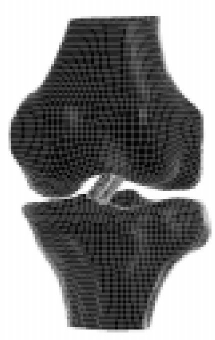

Fig. 1 3-D FE model of a human knee

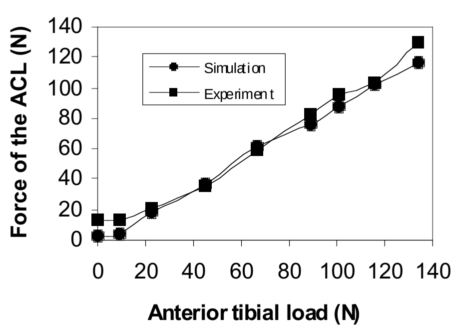

Fig. 2 Validation of the knee model with the experimental data

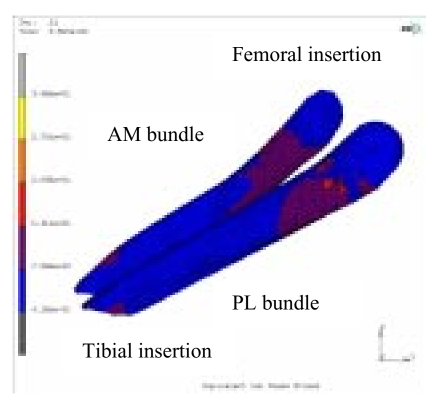

Fig. 3. Stress distribution within the AM and PL bundles under 134N anterior tibial load with the knee at full extension.