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The Biomedical Study of the Subsequent Injury Induced by MCL Rapture

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Abstract—**The rupture of the medial collateral ligament (MCL) occurs frequently during high valgus impact or sporting activities and commonly results in subsequent injuries. Experimental and clinical studies have been widely performed, but were limited to obtaining stress distribution and conducting parametric studies. The purpose of this study was to develop a validated three-dimensional finite element model of knee joint, and to analyze the kinematics and stress distribution of MCL deficient knee in response to typical loading and boundary conditions. The model was developed from magnetic resonance images and validated by the experimental data in the literature. The validated model was applied to analyze the kinematics and stress distribution of the MCL deficient knee exposed to three different loading cases: a valgus moment of 10Nm, an internal rotation of 15Nm and a posterior-anterior load of 200N. Valgus loading was found to be an important factor to abnormal stress distribution in MCL deficient knee. Instead, posterior-anterior loads exhibited few influences on the biomechanical behavior in MCL deficient knee. Under the internal rotation load, remarkable increase of stress in ACL only occurred when the knee flexed. The stability decreased with increasing flexion angle under all loading conditions. This study could help to understand the various subsequent injuries led by MCL injuries, and to predict the potential risks from external loads that should be avoided in rehabilitation.**

I. Introduction

THE rupture of medial collateral ligament (MCL) occurs frequently during high valgus impact or sporting activities [1], [2]. Although clinical studies revealed that MCL could heal spontaneously, its biomechanical properties remain inferior to those of normal ligaments [1]. The decline of the MCL may gradually damage the structures such as anterior cruciate ligament (ACL) and medial meniscus and finally lead to the knee arthritis [1], [3]. Various preventive measures have been studied but limited efficacy was obtained [4]. The outcomes of surgery as well as rehabilitation were also variable [2].

Extensive research has been performed to understand the biomechanics of MCL, which is vital to protect the ligament from injury and to improve the measures of treatment and rehabilitation. The MCL is considered to be the primary static stabilizer in the medial side of the knee. It can resist valgus loads and prevent excessive translation of medial meniscus [5]. The forces and strains in the MCL under typical loading and boundary condition have been obtained experimentally [6], [7]. Femoral or tibial attachment was found to be most susceptible to MCL rupture. However, it is still in dispute about which one is more risky [2]. Subsequent changes of kinematics and in situ force of involved structures following MCL injuries have also been measured [3]. The finite element (FE) models of knee joint with experimental validation can be useful in predicting the stress distribution and conducting parametric studies, which are difficult to investigate experimentally. Preliminary studies represented ligaments as one dimensional (1D) springs and focused on the joint kinematics and ligament forces. Other researchers developed 3D models of parts of knee joint so that the stress distribution as well as the contact and the friction in the ligaments could be calculated [9]. More re-cent studies have constructed models of entire knee in order to simulate complicated practical condition [10]. The validation of these models based on experimental data, however, was limited. Furthermore, little computational analysis has been focused on the biomechanical behavior of the MCL deficient knee under typical loading and boundary conditions.

The purpose of this paper was to determine the biomechanical behavior of the MCL deficient knee under typical loading and boundary conditions using a validated 3D FE model of knee joint. The model was validated by the experimental data in the literature including geometric measurement and kinematics of the knee as well as the in situ forces in anterior cruciate ligament (ACL) under the anterior tibial loads. The validated model was applied to analyze kinematics and stress distribution of knee when the MCL is completely ruptured. Three load cases were applied: a valgus moment of 10Nm, an internal rotation of 15Nm and a posterior-anterior load of 200N. A compression load of 500N (the body weight) was applied for all three cases. Both 0 and 30 degrees of knee flexion were considered.

II. Materials and Methods

A. Knee joint geometry

The 3D FE model of in vivo knee joint was reconstructed using magnetic resonance (MR). The knee (male, 28 years old, without any abnormalities) was positioned at full

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extension and subjected to a MR image scan in the sagittal plane using a fat suppressed gradient echo sequence and 1.5T magnets. A series of slices separated in 2mm intervals with the pixel size of 0.469×0.469 mm2 were obtained. Under the guidance of surgeon, the geometric information of relevant tissue was extracted from MRI using MIMICS (Materialise, Inc., Belgium). Four-node tetrahedral elements were used to mesh the bones, cartilages, menisci and ligaments (except the superficial MCL) in consideration of the irregular shape of the articular tissues. Four-node shell elements were used to mesh superficial MCL since it's thin. Fig. 1 shows the complete 3D FE model of knee joint containing femur, tibia, fibula, cartilages, medial meniscus, lateral meniscus, anterior crucial ligament (ACL), posterior crucial ligament (PCL), lateral collateral ligament (LCL), deep MCL and superficial MCL.

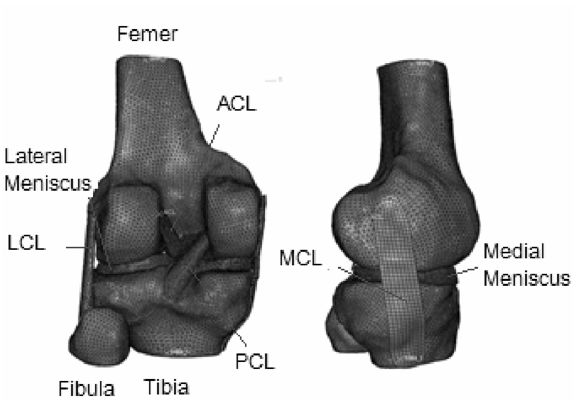

Fig. 1. FE model of the knee joint

To verify the accuracy of the geometry of the model reconstructed, critical dimensions including length and insertions of the ligaments, as well as depth, width and cross-sectional geometric parameters of the meniscus were compared with the published data measured on the cadaveric knees and the MRI of the knees [11], [12]. The deviation of each parameter was found to be less than 3%, which showed that the geometry of the model has high accuracy.

B. Material of tissues

Since the deformation of the bone was far less than that of the soft tissues, the bones were assumed to be rigid. The viscoelastic effect of the soft tissue was not significant under the quasi-static loads, therefore ligaments were considered to behave as a hyperelastic material described by the strain-energy function of Mooney-Rivlin, as shown in Equation (1), which fits the experimental results of the uniaxial tension well.

$$W=C_{10}(I_1^c-3)+C_{01}(I_2^c-3)+(J-1)^2/d \qquad (1)$$

Where W is the strain energy per unit of reference volume. I_1^c and I_2^c are the first and second invariants of Cauchy strain tensor. J is the determinant of Deformation gradient tensor. C10, C01 and d are the material constants.

The material constants (C10, C01 and d) were determined through a least-squares-fit procedure, as shown in Equation (2).

$$E = \sum_{i=1}^{n} (1 - \sigma_i^{th} / \sigma_i^{test})^2 \qquad (1)$$

Where σ_i^{test} is a stress value from the test data, and σ_i^{th} is the nominal stress derived from the material model. The constants of equation (1) are determined when the E reached the minimum value using the Lagrange Method. The material model of each ligament was developed respectively according to the published experimental data (due to lack of data, material of LCL was assumed to be equal to that of MCL) [13], [14]. Cartilage was considered to behave as a single-phase linear elastic and isotropic material with the elastic modulus of 5MPa and the Poisson's ratio of 0.35 because of the little deformation [15]. For the same reason, meniscus was assumed to be a single-phase linear elastic and isotropic material with an elastic modulus of 59MPa and a Poisson's ratio of 0.49 [16].

C. Loading and boundary conditions

Finite sliding contact without friction was established between bones, meniscus and bones as well as ligaments and bones [10]. Meniscus and ligaments were attached to bones by setting up the constraint of the proper nodes at the attachments. The deep MCL was attached to the external periphery of the medial meniscus identically [9].

The published experimental configuration was simulated, with the femur fixed and the tibia free to move in five degrees of freedom (DOF) [8], [9]. The tibia was flexed at 0° and 30° respectively by simulation, and the incremental anterior tibial loads (20, 40, 60, 80 and 100N) were applied for the tibia. The calculated forces in the ACL were compared to the corresponding forces of the experiment published in order to validate the model. The verified FE model was then used to analyze the kinematics and the stress distribution of the knee.

In order to determine subsequent changes of the kinematics and stress distribution resulted by the complete rupture of the MCL, Three load cases was considered: the tibia was flexed at 0° and 30° by simulation, a valgus moment of 10Nm, a internal rotation of 15Nm and a posterior-anterior load of 200N, and were applied respectively. A compression load of 500N (the body weight) was applied for the three cases.

The finite element calculation was performed with ABAQUS (Simulia Inc., USA). An optimized mesh size was determined while the differences in peak stresses between this particular mesh and its double-densed mesh were less than 10%.

III. RESULTS

A. Anterior tibial load

Under the incremental anterior tibial loads (20, 40, 60, 80N and 100N) at full extension, the posterior tibial translation was calculated to be 0.65, 1.3, 1.9, 2.5 and 3mm respectively, which were about 0.5mm higher than the published experimental data (Fig. 2(A)). The computational forces in the ACL were 12, 25, 37.5, 52 and 66N respectively for the same loads. The differences between the calculated forces and the published experimental data rose gradually with the increasing loads and reached the peak of 12% under 80N of the load (Fig. 2(B)).

Under the same loading condition at 30 degrees of flexion, the posterior tibial translations resulted to be 0.8, 1.7, 2.7, 3.4 and 4.3mm, and the corresponding forces in ACL resulted to be 16, 32, 49, 68 and 86N. Fig. 2(C) shows that the computational tibial translations matched the published experimental data in most of the loading range except for the load of 80N, at which the difference increased slightly. This difference might be caused by the exclusion of anatomic features such as muscle and fat and the material assumption inapplicable with the increasing load. Fig. 2(D) shows that the calculated forces matched well with the published data.

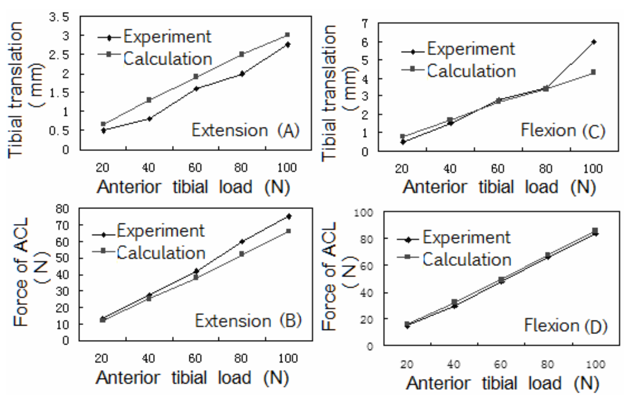

Fig. 2 Calculated tibial translations ((A), (C)) and Forces in ACL ((B), (D)) compared with the published data (0 and 30 degrees of flexion)

B. Combined loads in MCL deficient Knee

Under the combined loads of valgus moment and compression when the knee was at full extension, a remarkable deformation caused by MCL rapture was found to be in the ACL. As show in Fig. 3, the stress concentration happened near the femoral and tibial insertion. The maximal Von Mises stress increased obviously by 123%. The lateral meniscus suffered more load as a result of the MCL lesion. Its highest contact pressure occurred at the anterior site of the meniscus, with a maximum value of 4.9MPa. The stress in structures such as PCL and LCL also rose at different levels. The valgus rotation of tibia was calculated to be 1.26 degrees, which is 57% higher than that of the normal knee. At 30 degrees of flexion, valgus rotation increased to 3.4 degrees. The stress concentration was also observed at the region of the tibial insertion in PCL.

In response to a combined load of internal rotation moment and compression, an excessive rotation of 1.4 degree was observed in the MCL deficient knee at 0 degree of flexion, which increased to 2.6 degree at 30 degree of flex-ion. When the knee fully extended, the injury of MCL had no influence on the stress distribution in ACL. However, the maximum Von Mises stress increased remarkably as a result of the MCL rupture at 30 degree of flexion. The stress concentration occurred near the femoral insertion in PCL while slightly changes were observed in LCL and Meniscus at 0 and 30 degrees of flexion.

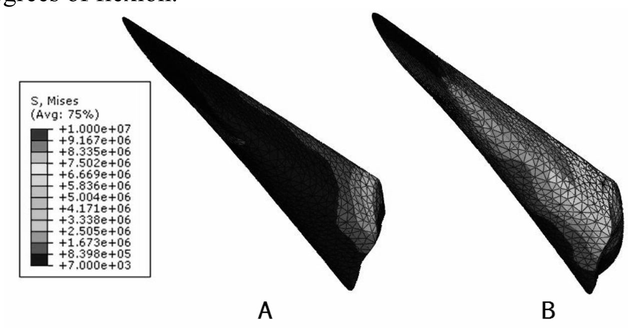

Fig. 3 Stress distribution of ACL. A: ACL in normal knee, B: ACL in MCL deficient knee

Under the load of posterior-anterior load at 0 degree of flexion, the stress concentration was predicted at the posterior site of the medial meniscus. The deformations in other structures, however, remains small even the MCL was ruptured. Similar results were obtained at 30 degree of flexion.

IV. DISCUSSION

In this paper we developed a relatively complete 3D FE model of the healthy human knee joint. The geometry of the model was obtained using magnetic resonance and was verified by the published data. The published experimental configuration was used to validate the model developed. The computational results matched well with the published experimental data. With the validated model, we analyzed the kinematics and stress distribution of MCL deficient knee in response to typical loads. Valgus loading was found to be an important factor to cause abnormal stress distribution in MCL deficient knee. Severe stress concentration happened in most of the structures under such loading and boundary conditions. Instead, Posterior-anterior loads exhibited no influence on the biomechanical behavior in MCL deficient knee. In additional, the stability decreased with increasing flexion angle, especially under the internal rotational loading, the maximum Von Mises stress remained normal at full extension, and increased remarkably as a result of the MCL rupture at 30 degree of flexion.

As shown in the comparative results, the peak stress in the MCL deficient knee increased remarkably when resisting valgus loading. Stress concentration happened near the femoral and tibial attachments in ACL, which may potentially cause the subsequent injuries of the ACL. These findings correspond to the clinical studies that ACL injures commonly in combination with MCL injuries [2]. It reveals that MCL plays an important role in resisting valgus moment. Besides, the valgus loading is one of the most dangerous loadings for the MCL deficient knee. Patients who are suffering MCL injuries or undergoing rehabilitation should avoid such external loads, especially in the condition of knee flexion.

The calculated results matched well with the published data in the most of the loading range. The existent deviation that computational translations were higher than that of experiment is most probably due to the exclusion of the anatomic features such as muscle and fat. Thus, the stiffness of the knee decreased and the translations calculated were higher than the experimental results. Besides, the assumption of isotropic material was no longer applicable when the deformation was very large. This may explain the increment of the differences between calculated and experimental data at higher external loads.

In the future work, more practical conditions will be simulated. Typical loads leading to MCL injuries and risky motion that should be avoided in rehabilitation will be analyzed. Ligament reconstruction can also be mimicked to help to improve the surgery. Therefore, mechanical proper-ties including failure models will be also developed to assure the precision of the results.

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