THE STUDY OF MENISCI EFFECT ON TIBIO-FEMORAL KINEMATICS IN A COMPUTATIONAL KNEE JOINT

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INTRODUCTION

The menisci are important knee structures that are often ignored in computational models of the knee. Previous work on this project has shown that representation of the menisci significantly reduces the contact forces and pressures on the articulating surfaces of the tibia and femur [1]. The purpose of this study was to examine the significance of the menisci on tibio-femoral kinematics. Improvements were made to a previously developed multi-body model of the meniscus by dividing meniscus geometries into smaller pieces in the circumferential direction. The effects of the menisci on knee kinematics were then examined by inserting the menisci models into a validated dynamic 3-D computational model of a knee loaded in a dynamic knee simulator (Kansas Knee Simulator, University of Kansas, Lawrence, KS). The predicted tibia-femur kinematics of the knee model with (w) menisci and without (wo) menisci were compared to experimental data during simulated squat and walking profiles.

METHODS

In the present study, knee geometries (tibia, femur, patella, articular cartilage, medial and lateral meniscus, and ligaments) of the subject specific model were derived from Magnetic Resonance Images (MRI) of a cadaver knee (78 year old female, right knee). 3D Slicer (www.Slicer.org) was used to convert the MR images into three-dimensional geometries. The knee geometries as well as ligament insertion/origin points were aligned in MSC.ADAMS (MSC Software Corporation, Santa Ana, CA) using acquired experimental data.

Six one-dimensional spring elements were used to simulate the knee ligaments including the anterior cruciate ligament (anterior and posterior bundles), posterior cruciate ligament (anterior and posterior bundles), lateral collateral ligament, and medial collateral ligament. Because the menisci remained intact during the experimental tests, previous studies suggested that representation of the menisci model might improve our prediction of tibio-femoral kinematics [2].

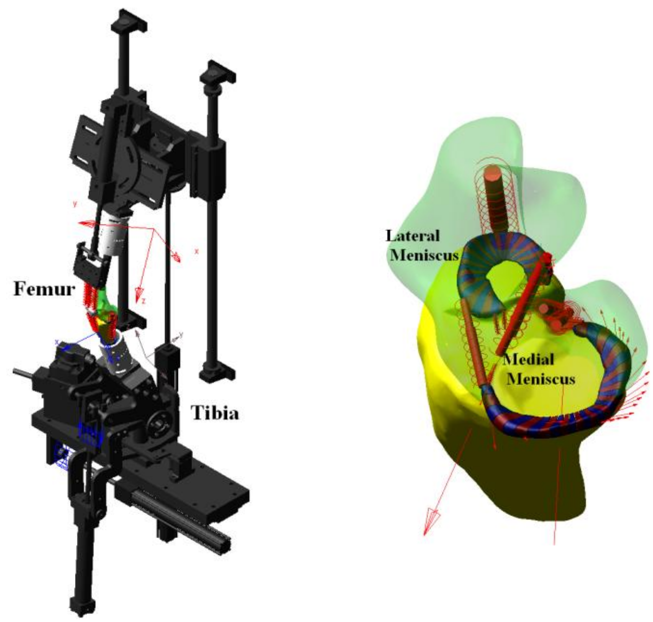

Figure 1: Knee model with menisci in ADAMS/View

Previously, a macro was written to automatically divide the menisci geometries into 4x4 mm discrete rigid elements. Since meniscus fibers are oriented circumferentially, several studies treated the menisci as a linearly elastic and transversely isotropic material [3]. In view of that, we modified the macro to divide menisci geometries in the circumferential direction (Figure 1). This new version of the macro also assigns mass properties, links the geometries, and automatically defines deformable contacts between the menisci, femur, and tibia. Hertzian contact theory was used to determine the parameters of the compliant contact force model based on cartilage material properties and contact geometry [1]. The menisci geometries interact through field elements which are represented by a 6x6 matrix of stiffness coefficients.

Matrix elements were estimated from matching the force-displacement relationship of an individual multi-body model of lateral meniscus to an isotropic FE model of the same geometry of meniscus in ABAQUS (Abaqus Inc., Providence, RI).

The lateral and medial menisci were attached to the tibia plateau with eight linear springs, representing horn attachments of the menisci. The insertions for the horn attachment were located on the anterior and posterior of each meniscus and on the tibia plateau. The transverse ligament was also modeled as a one-dimensional spring element. The resulting stiffness of all ligaments were similar to our previous study [1].

Simulation inputs to the computational model included the measured forces and torques produced by the actuators of the knee simulator. During testing, the femur and tibia position and orientation were measured using an Optotrak 3020 system with respect to the camera coordinate. To facilitate data comparison, all experimental data was transformed to the femur coordinate. Finally, in order to study the effects of the menisci, the root mean square error (RMSE) of predicted tibio-femur kinematics with and without menisci were compared to experimental data of the loaded cadaver knee during a walk and squat cycle.

RESULTS AND DISCUSSION

RMS errors of predicted position and orientation for simulation with and without the menisci are shown in Table 1. The position and orientation of the tibia were represented in the femoral coordinate system as Cartesian XYZ and Euler 123 coordinates. The maximum knee (hip) flexion angle in the squat test was 110 (55) degrees. During the squat a simulated body load was applied at the hip, but no out-of-sagittal -plane forces were applied at the ankle. The walking profile replicates knee loading and motion of ISO specification 14243-1. Results indicate that during the simulated squat exercise, the menisci improve the kinematics error for some orientations, but not all. The result of the walking profile shows that adding the menisci to the knee model generally decreases RMS errors. This may be explained by the fact that the menisci play a more significant role during the walk which has significant non-sagittal plane activity than the simulated sagittal plane profile like the squat.

CONCLUSIONS

In this study, a macro was developed that automated the process of creating a multi-body meniscus model from its MRI derived geometry. In addition, tibio-femoral kinematics were determined from tracking the positions and orientations of the tibia relative to the femur during a simulated walk and squat exercise. The limitation of this study includes the use of Hertzian contact theory to determine the parameters of the deformable contact model. It is believed that the extra lateral shift (x) on tibia kinematics of model with menisci was because of the simplification to define the contact force. Our current results suggest examining the effect of menisci on the tibio-femoral kinematics while each cruciate ligament is transected. In the future we will be considering more complex dynamic activities, such as squatting profiles with transected ACL/PCL. Also a design of experiment approach will be used to improve our calculation of parameters for the deformable contact force model.

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Table 1: RMS errors between model and experimental kinematics for model with (w) and without (wo).

RMSE	Orientation(deg) and position (mm)					
	Body 1	**Body 2**	**Body 3**	**X**	**Y**	**Z**
Squat (wo menisci)	1.47	2.31	9.95	17.19	6.41	6.10
Squat (w menisci)	1.49	3.70	10.90	23.01	6.05	6.40
Walk (wo menisci)	1.33	2.71	6.22	11.98	6.21	2.82
Walk (w menisci)	1.13	2.49	6.72	12.99	4.89	2.73