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MODELING OF PATIENT-SPECIFIC KNEE KINEMATICS AND LIGAMENT BEHAVIOR USING FORCE-DEPENDENT KINEMATICS

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SUMMARY

Patient-specific musculoskeletal knee models are promising tools to assist clinical decision making. However, in order to use these model-based predictions, validation of the models is of undeniable high importance. Since ligaments are highly important structures for guiding and stabilizing knee motion, this study investigated the behavior of these structures as a first validation method. Experimental data was collected to validate the model, namely a CT scan of a cadaveric leg to reconstruct three-dimensionally the bones, and kinematical data obtained through recording bone motion using optical reflective markers that were rigidly attached to femur and tibia. The distance between ligament insertion points, each determined on the reconstructed three-dimensional bone model, was used to deduce ligament lengths. These lengths were compared to the lengths computed by a subject-specific knee model which uses the force-dependent kinematics method. Furthermore, knee kinematics as model output were compared with experimental kinematic data. Using subject-specific information such as contact geometry and ligament insertions, the model could provide a good estimate of the measured ligament lengths and knee kinematics. These results show that accurate predictions of ligament length changes can be made using a patient-specific knee model.

INTRODUCTION

Ligaments are important structures for stabilizing the knee joint. When developing a musculoskeletal knee model, these structures can strongly influence the outcome of the computed knee kinematics. Adding patient-specific information, such as ligament insertion points, is therefore essential to correctly predict the movements. In this study, a patient-specific knee model is validated by comparing experimental data of knee kinematics and ligament lengths to the model output while performing a squat motion.

METHODS

Data were collected as described by Victor et al. [1]. A CT scan was performed on a cadaveric leg with attached passive optical reflective markers. This CT data was used to automatically segment the bones using Mimics® (Materialise N.V.). After the three-dimensional reconstruction of femur and tibia, relevant surface landmarks for identifying the insertions of the ligaments were located. The localization of these ligament insertion points was done using the quantitative morphologic description of LaPrade et al. [2]. The distance between the insertion points on femur and tibia (superficial MCL (sMCL) proximal and distal) and on femur and fibula (lateral collateral ligament (LCL)) was used as the length of these ligaments at any given position in the flexion arc of the knee joint.
The specimen was mounted onto a mechanical knee rig, with femur and tibia rigidly fixed in containers. Recordings were made, where the knee was passively flexed from full extension to 120 degrees and subsequently the knee was extended again. This movement was repeated several times while recording the markers to compute the relative position of all points on femur and tibia, permitting the measurement of ligament lengths throughout the knee flexion motion.

The patient-specific geometry and ligament insertions were implemented into the AnyBody Modeling System v. 5.3.1 (AnyBody Technology A/S, Denmark). The model consists of a standard model of a whole leg with patient-specific tibio-femoral contact geometry and four ligaments (sMCL proximal and distal part, LCL, anterior cruciate ligament (ACL) posterolateral and anteromedial part, and posterior cruciate ligament (PCL) anterolateral and posteromedial part). Ligament properties (stiffness and slack length) were based upon parameters given by Blankevoort et al. [3]. Cartilage and menisci geometry of a different subject is also implemented into the model, using available STL-files of Guess et al. [4]. As the meniscus and cartilage are not visible in the CT scan, a scaled version from this other data was taken instead.

The simulation with the AnyBody model was done using a new analysis method, named force-dependent kinematics (FDK) [6]. Instead of simplifying the knee joint to an idealized revolute joint, the FDK method allows internal tibio-femoral motions, assuming quasi-static force equilibrium between all the acting forces in these directions. Thus, the internal motions will be dependent on the subject-specific contact geometry and the surrounding soft-tissue structures, such as ligaments and muscles. In this model, which is grounded at the hip joint, the five internal degrees-of-freedom of the tibio-femoral joint were computed using FDK, while flexion-extension was driven to produce 120 degrees knee flexion over a time period of 120 seconds.

Simulating a passive squat motion, the model ligament lengths were calculated by determining the distance between the insertion points at each degree of flexion. Additionally, translational kinematics were studied by projecting the anterior-posterior (AP) translation of the medial and lateral femoral condyle centers (MFT and LFT) onto the tibial horizontal plane [5].

RESULTS AND DISCUSSION

Figure 1 shows the preliminary result of the subject-specific model length predictions of sMCL proximal, sMCL distal and LCL, compared with their experimentally measured length. The computed length changes correspond well with the experimental length changes during the squat movement until 90 degrees of knee flexion. For deep knee flexion, model predictions and experimental data start to differ. One explanation for the noticeable length offset is the fact that ligaments attach to a whole region on the bone, not on one point, which is assumed in the model. More importantly, there is a good correspondence between the main shape of the experimental and modeled curves.

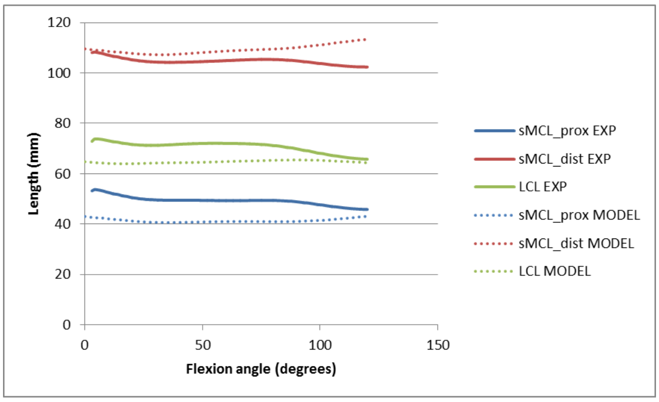

Figure 1: Comparison of experimental (solid line) and computed (dotted line) length of sMCL proximal part (blue), sMCL distal part (red) and LCL (green).

Figure 2 shows preliminary results of MFT and LFT computed in the model compared with experimentally measured MFT and LFT. The model overpredicts posterior translation, both for MFT and LFT and this effect is attributed to the way the menisci were modeled. In the model, the menisci are modeled as rigid bodies, not permitting any deformation. During deep knee flexion, the femoral condyles start rolling over these rigid bodies instead of deforming them, thus causing the larger translation.

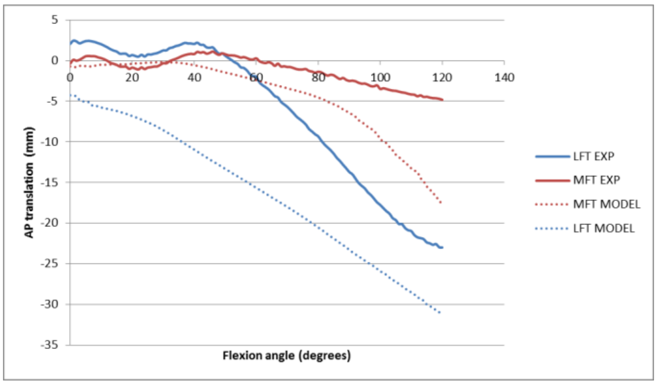

Figure 2: Comparison of experimental (solid line) and computed (dotted line) AP translation of medial (red) and lateral (blue) condyle center (MFT and LFT).

CONCLUSIONS

This work presents preliminary results showing that by using patient-specific geometry and ligament insertion points, a good estimation can be given of the length change of the ligaments. Introducing deformable menisci and comparing additional experimental and modeled knee kinematics can further validate this knee model and this will be done in future work.

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