VALIDATION OF A SUBJECT SPECIFIC FINITE ELEMENT MODEL OF THE HUMAN KNEE DEVELOPED FOR IN-VIVO TIBIO-FEMORAL CONTACT ANALYSIS

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INTRODUCTION
In vivo estimates of articular stresses within the knee joint would be beneficial for a wide range of studies relating joint mechanics to tissue damage/degeneration (e.g. osteoarthritis). Many finite element models have been proposed, but these models are generic in nature and their application to in vivo studies is limited. We have developed a method combining subject specific finite element (FE) models with high accuracy 3D knee joint kinematics to study tibio-femoral contact. The goal of this study was to validate this experimental/modeling approach for predicting articular contact stresses using a cadaver model loaded in a new knee simulator. Our ultimate goal is to use this validated FE method to predict in vivo tibio-femoral loading during a variety of movement tasks.

METHODS
A single right knee from a cadaveric specimen (male, age 31yrs) was prepared for placement in a load application system (Fig1 b). An arthroscopic procedure was employed so that soft tissues and joint capsule remained intact. Tantalum beads (0.8mm) were placed along the line of anterior cruciate ligament, (n=7), the menisci (n=18), the tibia (n=3) and femur (n=3). We interfaced the specimen with cups, aligned it with the load application system at 15 (°) flexion and placed it within the biplane radiography system (RSA) (fig 1a). The in-situ 3D kinematics history of all tautament beads using the RSA was collected at 250 frames/s (dynamic accuracy ±0.1mm). Knee motion was assessed during a series of uniaxial compressive loading protocols (0-1000N at a rate of 10mm/s). We then fixed surgically to the tibia plateau the K-scan pressure sensor (Tescan) using an arthroscopic procedure. After appropriate sensor calibration the loading protocol was repeated to evaluate the pressure distribution profile in the articular surfaces. This allowed for the evaluation of the effect of the sensor on the kinematics and further calibration of the model by comparing experimental and model results.

To construct the model a CT system was used to scan the knee in 0.5 mm increments at 0 degrees of flexion. Two plastic tubes filled with solution of Cupric Sulfate with paramagnetic properties were fixed in the femur and tibia. This along with the tantalum beads allowed for co-registration of bony geometry with soft tissue geometry. The slices were manually digitized for both bone geometry and cortical shell thickness. MRI measurements were performed using a GE 3 T system. We acquired standard anatomic scans which include: coronal T2, a sagittal proton density weighted imaging using a fast spin echo sequence and sagittal spoiled gradient echo imaging. Combination of the above imaging protocols allowed for geometry description of meniscal, collagenous tissue and articular cartilage. The CT and MRI volumetric data were reconstructed using surface maps as described by Treece et al. (2000). The model included both the cortical and trabecular bone of the femur and tibia, articular cartilage of the femoral condyles and tibial plateau, both the medial and lateral menisci with their horn attachments, the transverse ligament, the anterior cruciate ligament, and the medial collateral ligament. The material properties were chosen within the range of material properties available in the literature and from our previous studies. Cancellous and compact bone were modeled using an elastic-plastic material law with variations of the elastic coefficients depending on the anatomical location1. Cartilage was modeled using an elastic material law. The constitutive relation of the menisci treated the tissue as transversely isotropic and linearly elastic. The surface to surface tangential contact algorithm (ABAQUS/standard) was applied for the surface interaction in the knee joint with friction formulation and a friction coefficient of 0.001. The model was imported in an implicit formulation at the ABAQUS FEA software. We input the 3D kinematics on part of the boundaries and the loads on the femur. The kinematics was applied using the displacement of the tantalum markers. The pressure maps on the cartilage of the tibia plateau were drawn and compared to the K-scan readings. The medial and lateral contact forces were computed by summing the contact forces on the nodes. Under the application of an 1000 N compressive load at 15 degrees of flexion, six contact variables in each compartment (i.e., medial and lateral) were computed including maximum pressure, mean pressure, contact area, total contact force, and coordinates of the center of pressure.

RESULTS:
The kinematics history of the hard and soft tissue is the different loading sequences was reproduced. The displacement of the menisci markers was not affected significantly by including the pressure sensor in the joint. The FE model predicted the articular pressure profile as measured by the K-scan sensor (Fig2 - a). Maximum predicted contact pressure reached 5.5 MPa at full loading. This predicted value matched the peak pressure reading on the Kscan (5.7MPa). Convergence of the finite element solution was studied using three mesh sizes ranging from an average element size of 4 mm by 4 mm to 0.5 mm by 0.5 mm. The solution was considered converged for an average element size of 1.5 mm by 1.5 mm. Using this mesh size, finite element solutions for the cartilage and meniscus indicated that none of the contact variables changed by more than 2% when the femur and tibia were treated as rigid. However, differences in contact variables as large as 14% occurred when the k-scan sensor was excluded from the FE model. The largest difference was in the maximum pressure (18%).

DISCUSSION:
Among the principal conclusions of the study are that accurate finite element solutions of patient specific tibio-femoral contact and cartilage stress behavior can be obtained by combining high-quality static 3D imaging with high-accuracy kinematics (from dynamic RSA). Co-registration of multiple imaging modalities increased topological and geometric model fidelity. Mesh size is a direct measure of subject specific geometry and an optimum element size of <1mm (with specific areas of finer meshing) is appropriate for studying contact behavior. Treating the bones as rigid did not significantly alter model predictions. We also concluded that, even in a very controlled uniaxial compression loading condition, exclusion of the K-scan sensor from the model can result in relatively large errors in contact and cartilage stress. We are now acquiring all of the elements required for this approach (e.g. CT, MRI, dynamic RSA) from patients undergoing ACL reconstruction. Thus, this validated method will enable prediction of in vivo articular stress in these subjects during strenuous activities.

REFERENCES