DIRECT VALIDATION OF HUMAN KNEE-JOINT CONTACT MECHANICS DERIVED FROM SUBJECT-SPECIFIC FINITE-ELEMENT MODELS OF THE TIBIOFEMORAL AND PATELLOFEMORAL JOINTS

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ABSTRACT

The primary aim of this study was to validate predictions of human knee-joint contact mechanics derived from finite-element models of the tibiofemoral and patellofemoral joints (specifically, contact pressure, contact area and contact force) against corresponding measurements obtained in vitro during simulated weight-bearing activity. A secondary aim was to perform sensitivity analyses of the model calculations to identify those parameters that most significantly affect model predictions of joint contact pressure, area and force. Joint pressures in the medial and lateral compartments of the tibiofemoral and patellofemoral joints were measured in vitro during two simulated weight-bearing activities: stair descent and squatting. Model-predicted joint contact pressure distribution maps were consistent with those obtained from experiment. Normalized root-mean-square errors between the measured and calculated contact variables were on the order of 15%. Pearson correlations between the time histories of model-predicted and measured contact variables were generally above 0.8. Mean errors in the calculated centre-of-pressure locations were 3.1 mm for the tibiofemoral joint and 2.1 mm for the patellofemoral joint. Model predictions of joint contact mechanics were most sensitive to changes in the material properties and geometry of the meniscus and cartilage, particularly estimates of peak contact pressure. The validated FE modelling framework offers a useful tool for non-invasive determination of knee-joint contact mechanics during dynamic activity under physiological loading conditions.

Keywords: musculoskeletal model, model validation, sensitivity analysis, muscle, ligament, cartilage, meniscus

INTRODUCTION

The knee is one of the most complex joints in the body and plays a critical role in daily locomotion [1]. It is comprised of two separate joints – the tibiofemoral joint and patellofemoral joint – each of which is often affected by pain and disease (osteoarthritis) that undermine one's mobility and quality of life [2]. Abnormal joint loading is widely acknowledged as a major contributor to the progression of osteoarthritis [3,4]. Accurate knowledge of joint contact pressure and force on a subject-specific basis may provide useful insight into the early diagnosis and prevention of knee osteoarthritis. An improved understanding of knee joint contact mechanics therefore would be beneficial to clinicians and biomedical scientists and engineers interested in reducing the burden of joint disease.

Non-invasive measurement of joint contact mechanics, specifically, contact pressure, contact area and contact force, is currently impracticable. Instrumented knee implants have been used to directly measure knee-joint loading in living people [5,6], but this approach is invasive and restricted to a limited number of subjects; furthermore, data obtained from instrumented knee implants are currently available only for the tibiofemoral joint. Medical imaging techniques such as MRI [7–10] and biplanar radiography [11–15] have been used to evaluate knee-joint contact mechanics in vivo. Although such approaches are non-invasive, the results are limited to the assessment of contact areas and axial strains. Detailed measurements of knee-joint contact mechanics may be obtained by performing in vitro mechanical tests on cadaveric specimens, and these data then may be used to validate subject-specific biomechanical models of the tibiofemoral and patellofemoral joints.

Finite element (FE) models are often used to describe variations in contact pressure, area and force in vivo during daily physical activity. Various studies have validated FE models by comparing the

model calculations against measured joint displacements and ligament strains; however, this approach does not explicitly evaluate the accuracy of the model predictions of contact pressure, area and force [16]. A few studies have measured joint contact pressure in cadaver specimens [17–20] but the results are often limited to static loading regimens with the joint often held at a fixed flexion angle. In addition, existing FE studies have focused on either the tibiofemoral joint or the patellofemoral joint alone. No study to our knowledge has reported measurements of contact pressures at both the tibiofemoral and patellofemoral joints for flexion-extension movements of the knee during weight-bearing activity.

The primary aim of this study was to validate calculations of joint contact mechanics obtained from subject-specific FE models of the tibiofemoral and patellofemoral joints against in vitro measurements of contact pressure, area and force obtained during simulated weight-bearing activity. A secondary aim was to evaluate the sensitivity of the FE model predictions to changes in the parameters defining the geometry and mechanical properties of the soft tissues, specifically, cartilage, menisci and the knee ligaments.

METHODS

A combined experimental and computational protocol was used to validate subject-specific FE models of the tibiofemoral and patellofemoral joints created from three cadaveric knee specimens (Table 1). Subject specificity in the present study relates mainly to individualisation of joint geometry, including the geometries of the bones (i.e., distal femur, proximal tibia and patella), cartilage, meniscus and the attachment sites of the ligaments and the patellar tendon. While differences in material properties across subjects were not accounted for, it ensures the modelling framework can be readily translated to in vivo studies as currently there is no reliable non-invasive technique for accurately evaluating the

material properties (e.g. stiffness) of the soft-tissue structures such as the ligaments and menisci in vivo. Mechanical loading experiments and model simulations were performed on specimens 1 and 3 for the tibiofemoral joint, and on specimens 2 and 3 for the patellofemoral joint. No data were recorded from the tibiofemoral joint of specimen 2 as this joint was damaged during testing. Data also were not recorded from the patellofemoral joint of specimen 1 because the scope of the study was expanded to include the patellofemoral joint only after experiments had been completed on specimen 1. Model validation was accomplished in two steps: each tibiofemoral joint model was first validated in isolation without the patella; an FE model of the whole joint, including the patella, then was developed and used to validate the patellofemoral joint model. Model validation was performed by comparing the predicted peak contact pressure, contact area and contact force against corresponding experimental data.

Cadaver experiments

Zirconium dioxide beads, each 2 mm in diameter, were embedded in the distal femur, proximal tibia and patella (four or more beads in each bone) of each specimen to enable biplane X-ray imaging of knee joint motion during dynamic activity (see below). Volumetric CT scans of the tibiofemoral and patellofemoral joints were acquired to determine the relative positions of the bone-embedded beads. The scans were performed using a Philips Brilliance 64 CT scanner: acquisition matrix = 1024×1024 , slice thickness = 0.67 mm, in-plane (transverse plane) resolution = 0.24×0.24 mm, space between slices = 0.33 mm. MR images of each specimen then were obtained using a clinical 3T scanner (Philips Achieva, Amsterdam, Netherlands). Proton density 3D spin echo scans were acquired using the following parameters: repetition time = 1300 ms, echo time = 36.6 ms, matrix size = 432×432 , slice thickness = 0.7 mm, and voxel size = $0.35 \times 0.35 \times 0.35$ mm. Due to time constraints and limitations in available resources, we performed one MRI scan of each joint with a commonly adopted imaging

sequence. Overall, a proton density sequence was found to provide the best image quality for the entire joint, including bone, meniscus and cartilage, all of which were later segmented from the MR images to reconstruct the 3D joint model.

Skin, muscles and fibrous membranes were removed from each specimen by a licensed orthopaedic surgeon. All other major structures including the anterior and posterior cruciate ligaments (ACL and PCL), medial and lateral collateral ligaments (MCL and LCL), cartilage and menisci were preserved. The patella, patellofemoral ligament (PFL), proximal portion of the patellar retinacula, patellar tendon and quadriceps tendon were also retained for subsequent testing of the patellofemoral joint. Knee joint contact pressure was measured using a Tekscan 4000 pressure sensor (Tekscan Inc., Massachusetts, USA). Contact pressure at the tibiofemoral joint was measured by inserting and fixing the sensor between the tibial cartilage and the menisci, whereas the sensor was inserted and fixed onto the surface of the patellar cartilage to measure contact pressure at the patellofemoral joint.

A dynamic cadaveric knee simulation apparatus (DCKSA) (Figure 1) was constructed to perform the mechanical loading experiments. The DCKSA was a two-degree-of-freedom (DOF) joint simulator: one servomotor-driven linear actuator was used to generate knee flexion while the other generated axial (compressive) loading of the joint by applying a force through a high tensile-strength string that simulated the action of the quadriceps muscle group. The simulated quadriceps force was applied approximately along the longitudinal axis of the tibia, but the precise magnitude and direction of this force was not measured. The top portion of the cadaver specimen was connected to the linear actuator through a bearing and universal joint while the lower portion of the specimen was connected to a uniaxial load cell, which itself was mounted on a hinge joint. Whereas the universal joint, bearing and hinge joint accounted for 2 DOFs of the cadaver knee, the remaining 4 DOFs were left free.

Joint pressures in the medial and lateral compartments of the tibiofemoral joint were measured during simulated stair descent (Figure 1A). Stair descent was chosen because this task is associated with relatively high contact forces transmitted at the tibiofemoral joint [21]. Due to the power limits of the linear actuators and a desire to protect the specimen from damage, the maximum joint reaction force applied was smaller than the peak force measured in vivo (i.e., approximately 1000 N applied in the present study compared to 1700 N measured in vivo using an instrumented knee implant [5,6]). Joint pressures in the patellofemoral joint were measured during a simulated squatting task under the application of a constant quadriceps force (Figure 1B). In these experiments the knee flexed from 15° to 60° in approximately 2 secs, and joint contact loading was achieved by applying a single dead weight of 315 N to the high-tensile string that simulated the action of the quadriceps. The simulated quadriceps force was applied approximately in the sagittal plane and its line-of-action was determined by attaching two additional beads to the high-tensile string and using bi-plane fluoroscopy to measure the positions of these beads.

During mechanical testing motion of the beads was tracked to an accuracy of 0.2 mm using a bi-plane X-ray imaging system [22]. Joint contact pressure was measured using a Tekscan system (Figure 1D) while a uniaxial load cell was used to record the tibial axial force (Figure 1A). Details regarding the specifications, calibration and instrumentation of the Tekscan pressure sensor are provided in the Supplemental Material. At the completion of the experiments, eight additional beads were placed at the corners of the sensor (Figure 1E), and the *in-situ* position of the pressure sensor on the tibial plateau and its covered regions were determined from biplane X-ray imaging.

Computational model and simulations

The geometry of the bones, cartilage and menisci were segmented from the MR images using Amira (FEI Corporate, Oregon, USA) and subsequently refined using Geomagic Studio (3D Systems, South Carolina, USA). The surface smoothing process utilized standard functions in Geomagic, including removing spikes, patching small holes, relaxing surfaces and reducing noise. The average difference between the original and smooth geometry was less than 0.05 mm. Bone surfaces were meshed using quadrilateral and triangular 2D elements with an average size of 2 mm while hexahedral 3D elements with an average size of 0.9 mm were used to represent cartilage and the menisci. Each mesh component was imported into Abaqus v6.13 (Dassault Systemes, Vélizy-Villacoublay, France) and assigned a material property (Table 2). The bones were assumed to be rigid. Cartilage was defined as an incompressible hyperelastic material model while the menisci were represented as a transversely isotropic, linearly elastic material. The cruciate and collateral ligaments were each modelled as a multi-stranded 1D non-linear spring. The force-strain behaviour of each model ligament was described as follows [23]:

$$f = \begin{cases} \frac{1}{4}k \cdot \varepsilon^2 / \varepsilon_1, & 0 \leq \varepsilon \leq 2\varepsilon_1 \\ k(\varepsilon - \varepsilon_1), & \varepsilon \leq 2\varepsilon_1 \\ 0, & \varepsilon < 0 \end{cases} \quad (1)$$

where f is the tensile force developed by the ligament, k is its stiffness, ε is ligament strain, and ε_1 is non-linearity strain parameter, which was assumed to be 0.03 [24]. The ligament attachment sites were estimated from the MR images. The ligaments were modelled as multi-stranded structures in an attempt to distribute the reaction forces to a larger region and to mimic their physiological behaviour. For ligaments comprised of different bundles with different stiffnesses, an average force-strain curve was calculated and assigned to each strand. The patellar retinaculum and patellar tendon were each modelled as multi-stranded 1D linear springs (Table 2). The remaining joint capsular structures were not included in the model as they were removed in the cadaver specimens

to accommodate the pressure sensor. The knee joint FE model was assembled in Abaqus (Figure 2). Cartilage was rigidly connected to the outer surfaces of the corresponding bones. The medial and lateral menisci were connected to the tibial plateau through horn attachments and to each other via the transverse ligament (Table 2). The MCL was attached to the medial external surface of the medial meniscus. Penalty-based frictionless, hard, surface-to-surface contacts were defined between all possible pairs of cartilage and meniscal elements.

Mesh sizes of 0.5, 0.9 and 2.0 mm were used to test model convergence, and the calculated values of peak pressure, contact force and contact area were compared for these three cases. A mesh size of 0.9 mm was found to provide an acceptable compromise between contact prediction accuracy and computational efficiency: the average error was less than 5% of that obtained using a 0.5 mm mesh. Two layers of elements were used to represent the thicknesses of the femoral and tibial cartilage while three layers were used to model the thickness of the patellar cartilage.

An explicit method was used to overcome difficulties with model convergence encountered when implementing the implicit method. Mass scaling was used in the explicit method to improve computational performance of the model simulations. For the tibiofemoral model, a stable time increment of 1.0×10^{-6} s was used for local mass scaling, which resulted in only 306 out of 18200 total elements (~1.7%) having artificially inflated masses. Global mass scaling with stable time increment targets of 1.0×10^{-5} s and 1.0×10^{-4} s were also evaluated. Based on the results of the convergence tests, global mass scaling with a stable time increment of 1.0×10^{-5} s was used. The maximum error calculated across all time steps was 3.3% for all contact variables in both the medial and lateral compartments of the tibiofemoral joint. The local mass scaling simulation took ~21 hours to converge on a standard desktop PC (Intel i7-4770 quad-core CPU, 16 GB RAM), while global mass scaling with the stable time increment target of 1.0×10^{-5} s reduced simulation time to ~4 hours. A similar procedure was followed

for the patellofemoral joint model to confirm the validity of global mass scaling with a stable time increment target of 1.0×10^{-5} s.

Bone geometries were also obtained from the CT scan along with the locations of the beads embedded in the bones. The CT bone model was aligned to the MRI bone model using the “Best Fit Alignment” algorithm available in Geomagic Studio, and the bead positions were then registered to the FE model. Bead motions obtained from bi-plane fluoroscopy then were used to calculate joint kinematics in the FE models. Alignment of the CT and MRI bone geometry models resulted in average deviations of less than 0.5 mm.

All simulations were run using a kinematics and force hybrid control scheme. Tibiofemoral contact was simulated by prescribing all three rotations of the tibia relative to the femur as well as anterior tibial translation and mediolateral translation with only superior-inferior translation left free; an axial (compressive) force was applied to the tibia to control superior-inferior translation at the tibiofemoral joint. Patellofemoral contact was simulated by prescribing all three rotations of the patella relative to the femur as well as patellar shift with superior-inferior and anterior-posterior translations of the patella relative to the femur left free. Force control was implemented in the superior-inferior and anterior-posterior directions.

Sensitivity analyses

Sensitivity studies were undertaken to investigate how changes in the material properties and soft tissue thicknesses affect model predictions of joint contact mechanics. The sensitivity analyses were performed on one tibiofemoral joint model and one patellofemoral joint model. The following parameters were altered in the tibiofemoral joint model: the stiffnesses and thicknesses of cartilage and the menisci; stiffnesses of the meniscal horn attachments and transverse ligament; stiffnesses of

the ACL, PCL, MCL and LCL; and the coefficient of friction at the articulating surfaces of the joint. In the patellofemoral joint model, we examined the sensitivity of model response to changes in the stiffness and thickness of cartilage; stiffnesses of the PFL and retinaculum; stiffness of the patellar tendon and its tibial insertion sites; and the coefficient of friction at the joint contact surfaces. The stiffnesses assumed for cartilage, meniscus, ligaments and tendon were each perturbed by $\pm 20\%$. Because the stiffnesses of the retinaculum and PFL are not available in the literature, and also because the lower portions of the retinaculum were trimmed in the cadaver specimens to accommodate the pressure sensor, the overall stiffnesses of the retinaculum and PFL were perturbed by $\pm 20\%$ and $\pm 40\%$, respectively. Cartilage and meniscal thicknesses were each altered by $\pm 15\%$ to account for differences in segmentation thresholds. Further reductions in the thicknesses of either cartilage or the menisci produced very thin elements and prevented model convergence. The thickness alterations introduced for cartilage and the meniscus were based on local thickness and were achieved through additional hypothetical simulations. Specifically, the Poisson's ratio of the corresponding component was first set to be zero. A hypothetical surface pressure then was applied to compress (positive pressure) or inflate (negative pressure) the component by 15% of its original volume. Under the assumption of zero Poisson's ratio, the component underwent no lateral expansion when the surface pressure was applied. Thus, the change in tissue volume was equal to the change in its thickness.

Data analysis

A root-mean-square-error (RMSE) criterion was used to compare model-predicted peak contact pressure, contact area and contact force (collectively referred to here as 'contact variables') against corresponding experimental data. RMSE was normalized by the peak value measured for each contact variable to obtain a normalized root-mean-square-error (NRMSE). Pearson correlation coefficients were also computed for the contact variables to demonstrate the similarities in trends between the

FE model predictions and experimental results. A contact centre-of-pressure (COP) was calculated from the FE simulations and compared against experiment. Contact COP was defined as:

$$COP = \frac{\sum_{i=1}^n X_i P_i A_i}{\sum_{i=1}^n P_i A_i}$$

where X_i is the spatial coordinate of the contact surface centroid for the i^{th} element; P_i is the contact pressure acting on the i^{th} element, which was taken as the average of the nodal contact pressure; and A_i is the area of the corresponding condyle or facet.

The pressure sensor was not large enough to cover the entire tibial plateau, especially the medial compartment. A custom Python program was written to calculate the contact force and contact area for the sensor-covered region on the basis of the Abaqus simulation results. Comparisons between the model predictions and the experimental measurements were made for the sensor-covered region.

RESULTS

Tibiofemoral joint

The tibiofemoral contact pressures measured for the two specimens displayed a similar pattern (Figure 3). Overall, peak pressures acting between the femoral and tibial cartilage occurred in the medial compartment, and those acting between the menisci and tibial cartilage occurred in the lateral compartment. As the knee flexed, the contact areas moved posteriorly (Figure 4) and peak contact pressure shifted gradually from the medial to the lateral compartment (Figure 3).

The patterns of tibiofemoral joint contact area and peak pressure calculated in the model were consistent with experiment throughout the trial cycle (Figure 3). The COPs in the medial and lateral compartments translated posteriorly as the trial cycle proceeded, except for the lateral

compartment of specimen 3, where the COP remained relatively stationary (Figure 4A). Throughout the trial cycle, the mean difference between the locations of the model-predicted and measured COPs was 2.3 ± 1.3 mm for the medial compartment and 2.4 ± 0.3 mm for the lateral compartment in specimen 1; and 4.7 ± 1.3 mm and 2.9 ± 1.4 mm, respectively, for the medial and lateral compartments in specimen 3. Differences between model-predicted and measured superior-inferior translations were within 1.0 mm throughout the trial cycle for both specimens (Figure 4B). Maximum errors were 0.59 mm and 0.43 mm for specimens 1 and 3, respectively, while the corresponding mean errors were 0.35 mm and 0.05 mm. Peak contact pressure acting on the tibial cartilage in the cadaver specimens was approximately 3.3 MPa while peak contact area was approximately 1000 mm². The FE models generally under-predicted the measured contact areas. Overall, the NRMSE was below 20% for most of the contact variables, except for the contact area in the lateral compartment of specimen 1 and the contact force acting in the medial compartment of specimen 3, where the NRMSEs were approximately 30% (Figure 5, Table 3). Pearson correlations between the FE model predictions and experimental results were greater than 0.8 ($p < 0.005$), except for the medial contact area where the correlation coefficient was above 0.6 for both specimens. For specimen 1, mean reaction forces were 15.6 N and 58.2 N for the medial-lateral and anterior-posterior translational DOFs while mean reaction torques were 2.16 Nm and 1.95 Nm for the varus-valgus and internal-external rotational DOFs. For specimen 2, mean reaction forces were 97.5 N and 61.9 N for the medial-lateral and anterior-posterior translational DOFs and mean reaction torques were 0.98 Nm and 0.44 Nm for the varus-valgus and internal-external rotational DOFs.

Model-predicted contact variables for the tibiofemoral joint were most sensitive to changes in the material properties of cartilage and the meniscus (see Table S1 in Supplemental Material). The largest differences occurred in the peak contact pressure: a 20% increase (decrease) in tibiofemoral

1 cartilage stiffness resulted in an 8% increase (decrease) in the calculated peak pressure. Similar
2 results were obtained for a change in the stiffness of the model meniscus. In contrast to the
3 behaviour of cartilage, a softer (stiffer) meniscus did not always produce lower (higher) peak
4 pressures throughout the trial cycle. A softer (stiffer) cartilage or meniscus resulted in larger
5 (smaller) tibiofemoral contact area predictions. Tibiofemoral contact force was relatively insensitive
6 to a change in tibiofemoral cartilage stiffness (<2%) and moderately sensitive to a change in the
7 stiffness of the meniscus (~6%). The results obtained for the models with linear elastic cartilage and
8 contact friction were similar to those predicted by the nominal model, with a maximum difference in
9 tibiofemoral peak pressure of no more than 7%. Changes in the material properties of the soft
10 tissues did not result in significant changes in either the calculated values of the COPs (< 0.6 mm) or
11 the Pearson correlations between model-predicted and measured contact variables for the
12 tibiofemoral joint.

13
14 Alterations in the thicknesses of cartilage and meniscus produced substantial changes in model-
15 predicted contact variables (Supplemental Material, Table S2), especially peak tibiofemoral pressure.
16 In some cases, changes in cartilage and meniscal thicknesses also resulted in substantial decreases in
17 the correlations between model-predicted and measured contact variables. The location of the COP
18 at the tibiofemoral joint was relatively insensitive to changes in the thicknesses of cartilage and
19 meniscus, with variations in COP being no greater than 1.4 mm.

20 21 *Patellofemoral joint*

22 Model-predicted patellofemoral contact pressure distribution patterns were consistent with
23 experiment, with the FE models reproducing the striped pressure pattern observed in both

specimens (Figure 3). As the knee flexed, the calculated contact regions moved proximally on the medial and lateral patellar facets, in agreement with experiment.

Mean differences between model and experiment in the locations of the patellofemoral COP were 2.7 ± 1.1 mm and 1.5 ± 0.8 mm for specimens 2 and 3, respectively (Figure 4A). Maximum errors between model-predicted and measured translations (anterior-posterior and superior-inferior combined) in the sagittal plane were 1.32 mm and 0.50 mm for specimens 2 and 3, respectively, while the corresponding mean errors were 0.81 mm and 0.44 mm (Figure 4B). The NRMSEs for peak patellofemoral pressure and contact area were on the order of 15% while that relating to contact force was around 6% (Table 3). Pearson correlations were larger than 0.8 ($p < 0.0005$) for all patellofemoral joint contact variables, except for peak pressure estimated for specimen 3 ($r = 0.711$) (Figure 5, Table 3). Maximum patellofemoral contact pressure was around 2.94 MPa for specimen 2 and 2.41 MPa for specimen 3. Patellofemoral contact area and contact force increased with knee flexion angle in both specimens. The model under-predicted patellofemoral contact area in both specimens: peak contact areas calculated in the model at 60° of knee flexion were 256 mm^2 and 276 mm^2 for specimens 2 and 3, respectively, compared to 339 mm^2 and 348 mm^2 obtained from experiment.

Model-predicted patellofemoral contact variables were most sensitive to changes in cartilage stiffness (Supplemental Material, Table S3). A 20% increase (decrease) in patellofemoral cartilage stiffness resulted in an 8% increase (decrease) in peak patellofemoral pressure and a 6% decrease (increase) in peak contact area. Patellofemoral contact force predictions varied by 6% when the stiffnesses of the PFL and retinaculum were each altered by 40%. The linear elastic cartilage model predicted a relatively large increase (9%) in peak patellofemoral pressure and a moderate decrease (4%) in peak contact area, similar to the results derived from the hyper-elastic model when

patellofemoral cartilage stiffness was increased by 20%. Contact friction had a relatively small effect on the calculated values of the contact variables. Varying the material properties of cartilage did not have a significant influence on the location of the patellofemoral joint COP, with changes of less than 0.2 mm observed throughout the trial cycle. Pearson correlations between model-predicted and measured patellofemoral contact variables were relatively insensitive to changes in the material properties of the soft tissues.

Changes in patellofemoral cartilage thickness had the most significant effect on the calculated values of peak patellofemoral pressure (Supplemental Material, Table S4). When the thickness of the femoral or patellar cartilage was altered by 15%, peak patellofemoral contact pressure varied by 16%. Substantial decreases in the Pearson correlations between model and experiment were also observed. Model-predicted patellofemoral contact variables were relatively insensitive to changes in the locations of the insertion sites of the patellar tendon on the tibial tuberosity.

DISCUSSION

We compared tibiofemoral and patellofemoral joint contact pressures predicted by explicit FE models created from subject-specific geometries of each joint against direct measurements obtained from in vitro experiments of simulated weight-bearing activity. Average NRMSEs for the contact variables (peak pressure, contact area and contact force) were 17.4% for the tibiofemoral joint and 12.4% for the patellofemoral joint when comparing model-predicted and experimental results. Pearson correlation coefficients quantifying differences in the time histories of the calculated and measured contact variables were generally above 0.8. Errors in the calculated locations of the COPs were 3.1 mm and 2.1 mm for the tibiofemoral and patellofemoral joints, respectively.

The level of agreement between model and experiment is comparable to that reported previously by others. Donahue et al. [17] investigated tibiofemoral joint contact mechanics under quasi-static loading and reported a peak contact pressure of 3.78 MPa and a total contact area of approximately 750 mm² when a 1200 N compressive force was applied. These results compare favourably with those obtained in the present study, where a maximum tibiofemoral compressive force of 1000 N produced a peak contact pressure of 3.3 MPa and a corresponding contact area of 950 mm². Haut Donahue's model yielded a much smaller NRMSE of 5.5%, which may have resulted from the simpler form of quasi-static loading and the subject-specific optimization of model parameters used in their work. Mootanah et al. [19] compared estimates of joint contact pressures derived from a subject-specific FE model of the tibiofemoral joint against pressure measurements obtained from a series of static loading experiments in which a combined compressive load of 374 N and a varus-valgus bending moment of 15 Nm was applied to cadaver knees. They reported RMSEs of 0.16 MPa and 1.49 MPa for the medial and lateral peak contact pressure, and RMSEs of 139 N and 65 N for the medial and lateral compartmental contact forces. These results are similar to those given in Table 3 even though the magnitude of the compressive force used by Mootanah was much lower than that applied in the present study. Besier et al. [8] validated a subject-specific FE model of the patellofemoral joint using in vivo MRI data to estimate patellofemoral contact area. They reported errors in the calculated joint contact areas of less than 5%, though maximum errors obtained for some models were as high as 32%. By comparison, NRMSE for the patellofemoral contact area obtained in the present study was 18%. The difference in error estimates between these two studies is likely to be explained by the different loading conditions employed; Besier et al. applied static loads with the knee flexion angle held fixed whereas we simulated dynamic weight-bearing tasks with the knee moving through relatively large flexion angles.

The specimen-specific FE models consistently under-predicted the contact areas measured at the tibiofemoral and patellofemoral joints (Figure 5). Several factors may have contributed to this result. First, the hyperelastic material constants assumed for cartilage were obtained from measurements performed on small cylindrical plugs of cartilage rather than the whole joint, and implementation of these constants were previously found to under-predict contact area on a much smaller scale than that investigated here [25]. Second, it is possible that insertion of the pressure sensor, which was thin (0.102 mm) and flexible, increased the joint contact area during the experiments, an effect that was not taken into account in the FE simulations. Third, the shapes of the specimen-specific cartilage and meniscal surfaces may not have been reproduced precisely in the model due to the smoothing algorithms used to reconstruct the solid models.

We found that the stiffness and thickness of both cartilage and the menisci had the largest influence on predictions of joint contact mechanics. Consistent with results reported previously [26] the calculated values of peak contact pressure in both the tibiofemoral and patellofemoral joints were more sensitive to variations in cartilage and meniscal stiffness and thickness than were contact area, contact force and the location of the COP. Changes in cartilage and meniscal thickness also decreased the Pearson correlations between the calculated and measured contact variables, whereas changes in the material properties typically had a minimal effect in this respect. This finding indicates that inaccuracies in model geometry can quantitatively and qualitatively affect predictions of joint contact mechanics; for example, a change in cartilage thickness may alter not only the magnitude of peak joint pressure but also the time at which peak pressure occurs during a particular activity. Li et al. [27] concluded that under axial tibia compression of up to 1400 N, a 10% change in cartilage thickness resulted in a 10% change in contact pressure, which is similar to the results derived from the present study. Our model simulations showed that a 15% change in femoral or tibial cartilage thickness resulted in a 15% change in peak contact pressure.

Dhafer et al. [28] concluded that model-predicted values of knee joint kinematics and contact pressure were sensitive to changes in the material properties of the ACL. Smith et al. [29] found that tibiofemoral joint kinematics were particularly sensitive to the values assumed for the stiffnesses and reference strains of the ACL, MCL and LCL while contact pressure magnitudes were most sensitive to changes in ACL stiffness and reference strain. In contrast, we observed that changes in cruciate and collateral ligament stiffnesses had a relatively small influence on the predictions of the contact variables (contact area, force and pressure). Both of these studies used the Monte Carlo method and explored a much wider spectrum of ligament material properties than that investigated here. However, we used accurate measurements of knee joint kinematics obtained from biplane X-ray imaging to prescribe the boundary conditions in the model simulations, thus minimizing the effects of changes in the ligament properties on tibiofemoral joint kinematics. Indeed, Dhafer et al. (2010) found that the ligaments which caused small changes in tibiofemoral kinematics also caused small variations in the calculated values of joint contact pressure.

Because of the sensitivity of FE model predictions to changes in the geometry and stiffness of cartilage and meniscus, future studies ought to focus on implementing more advanced imaging methods to characterise these properties more accurately. Higher resolution MRI may allow more detailed joint geometries to be included in the model while quantitative MRI techniques, such as T1rho [30,31] and diffusion tensor imaging [32] may enable in vivo soft tissue material properties to be measured non-invasively.

Several limitations ought to be considered when interpreting the results of this study. First, experiments were performed on three cadaver specimens, with contact pressures measured in the

medial and lateral compartments of two tibiofemoral joints and two patellofemoral joints. This small number of specimens was necessitated by the time-intensive nature of the cadaver experiments and the subject-specific FE model development and analysis. Importantly, similar results were obtained at both joints across the different specimens tested. In addition, extensive sensitivity analyses were performed on both the tissue material properties and soft tissue geometries to obtain a more thorough appreciation of how variations in model parameters may potentially alter predictions of joint contact mechanics.

Second, the loading applied to the specimens as well as the rates at which the cadaver knees were actuated are lower than that typically observed in vivo for the two activities simulated: stair descent and squatting. The angular velocity of the knee was limited to a maximum of 75 deg/sec because this was the maximum speed at which the DCKSA was capable of operating. Lower magnitudes of compressive joint force were also used to limit the risk of damage to the structures in and around the cadaver knee. In the stair descent simulation, peak compressive force at the tibiofemoral joint was approximately 1.7 BW compared to in vivo measurements of around 3 BW. In the patellofemoral joint experiments, a constant load of about 0.6 BW was applied to the quadriceps tendon whereas the peak quadriceps force during squatting has been estimated to be 2-3 BW [33].

Third, the tibiofemoral and patellofemoral joint models were validated separately rather than simultaneously. It was not possible to measure contact pressures and contact areas in both joints concurrently because two sets of Tekscan systems were not available, and also because it was difficult to position the pressure sensor in the tibiofemoral joint with the patella still attached. Future work should focus on developing improved pressure sensor technology to enable joint contact measurements to be performed in both joints simultaneously under dynamic loading.

Finally, a number of simplifications were introduced in the subject-specific FE models. Each bone was modelled as a rigid element, which may be justified by its relatively high elastic modulus (i.e., 10 GPa) [34]. Previous analyses have shown that modelling bone as a rigid structure has no substantial effect on the predictions of joint contact pressures and areas [35]. The ligaments were represented as one-dimensional springs, ignoring their three-dimensional geometries. However, multiple strands of nonlinear springs (three each for the LCL, ACL and PCL and five for the MCL) were used to simulate the volume of each ligamentous structure. Although specimen-specific material properties were not directly implemented in the joint models, extensive sensitivity studies were performed to understand how changes in the mechanical properties and geometries of the soft tissues assumed in the model affect estimates of joint pressure, contact area and compressive load. Patellofemoral cartilage was modelled using material constants obtained from tibiofemoral cartilage samples, which may not be entirely appropriate. Future work comparing cartilage from these two joints may provide insight into the development of more accurate models of patellofemoral cartilage. Similar to previous studies, the geometry and mechanical behaviour of the pressure sensor was not included in the model simulations. Because the Tekscan sensor was thin (0.102 mm) and flexible, it was not possible to accurately determine the initial deformation and boundary conditions of the pressure sensor.

In summary, the present study validated knee joint FE models with subject-specific geometries of the tibiofemoral and patellofemoral joints using novel experimental data recorded from simulated dynamic weight-bearing tasks. Average differences in peak contact pressure, contact area and contact force between model and experiment were on the order of 15% for both the tibiofemoral and patellofemoral joint. Errors in the calculated locations of the COPs were around 2.5 mm. The modelling framework proposed in this study may be used to analyse general trends related to knee contact mechanics in a cohort of individuals. In addition, the model can also be applied in predictive

simulation scenarios, such as joint contact mechanics after ligament resections or meniscectomy [36]. However, caution is advised when interpreting the results on a subject-specific basis. Sensitivity analyses showed that the model predictions are most profoundly affected by changes in the material properties of cartilage and meniscus and the parameters describing articular surface geometry. The accuracy of FE knee model predictions may be improved by obtaining more precise subject-specific measurements of cartilage and meniscal thicknesses and stiffnesses as well as articular surface geometry.

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FIGURE CAPTIONS

Figure 1: A) Schematic diagram showing the key components of the dynamic cadaveric knee simulation apparatus (DCKSA) and the test rig used to simulate tibiofemoral joint mechanics during weight-bearing activity. B) Schematic diagram showing the test rig used to simulate patellofemoral joint mechanics during weight-bearing activity. C) Photograph showing the experimental setup used to measure knee joint contact mechanics in vitro. D) Close-up of the cadaver knee specimen mounted in the DCKSA showing the Tekscan sensor inserted within the tibiofemoral joint prior to mechanical testing. E) Photograph showing the eight additional beads attached to the corners of the Tekscan pressure sensor. The position of the sensor relative to the tibial plateau was determined using bi-plane X-ray imaging. Two different sizes of beads – 2 mm and 3 mm – were used to assist in processing the bi-plane images.

Figure 2: Schematic diagram illustrating the structure of the FE model developed in this study. Symbols appearing in the diagram are as follows: ACL, anterior cruciate ligament; PCL, posterior cruciate ligament; MCL, medial collateral ligament; LCL, lateral collateral ligament; PFL, patellofemoral ligament and retinaculum.

Figure 3: Pressure distribution maps displayed on the tibial cartilage at 20%, 40%, 60%, 80% and 100% of the trial cycle (left) for specimens 1 and 3, and those for patellar cartilage at 10%, 50% and 100% of the trial cycle for specimens 2 and 3 (right). For each specimen, the left panel shows in situ Tekscan pressure measurements while the right shows the FE model predictions. The black shaded regions in the left-hand panels represent the pressure sensor. Specimen 1 was a left knee and specimens 2 and 3, right knees. For the tibiofemoral joint, the FE model simulation results and

experimental data are projected onto the plane representing the tibial condyles while for the patellofemoral joint the model and experimental results are projected onto the patellar facets. All results were normalized by the maximum pressure whose value is indicated above each plot.

Figure 4: A) Measured and model-predicted contact locations described by the center-of-pressure displayed on the tibial cartilage at 20% (*), 40% (○), 60% (□), 80% (◇) and 100% (+) of the trial cycle (left), and on the patellar facets at 10% (*), 50% (○) and 100% (□) of the trial cycle (right). Blue symbols represent experimental results while red symbols represent model predictions. B) Left: Differences between model-predicted and measured femoral translations in the superior-inferior direction (superior(+)/inferior(-)); Right: Differences between model-predicted and measured patellar translations in the anterior-posterior (anterior(+)/posterior(-)) and superior-inferior (superior(+)/inferior(-)) directions.

Figure 5: Time histories of the measured and model-predicted contact variables for the tibiofemoral joint (left) and patellofemoral joint (right) obtained during simulated weight-bearing activity. a) Peak contact pressure, b) contact area, and c) contact force.

Figure 1

Figure 2

Figure 3

Figure 4

Figure 5

Table 1: Physical characteristics of the specimen donors.

	Gender	Donor Age	Body Mass (kg)	Height (cm)
Specimen 1	Female	67	66	157
Specimen 2	Female	32	54	153
Specimen 3	Female	45	57.5	158
Mean $\pm$ Standard Deviation	n/a	48 $\pm$ 17.7	59.2 $\pm$ 6.2	156 $\pm$ 2.6

Table 2: Material properties assigned to the various soft tissues included in the FE models developed in this study.

Structure	Material Type	Material Property	Reference
Bones	rigid	N/A	[25]
Cartilage	Yeoh hyperelastic	Density: 1.5 g/cm ³ $C_{10}=1.675$, $C_{20}=3.225$	[18,26]
Meniscus	Linear elastic, transversely isotropic	Density: 1.5 g/cm ³ *Modulus (E, MPa): $E_1=120$, $E_{2,3}=20$ Poisson's ratio (ν): $\nu_{12}=\nu_{13}=0.3$, $\nu_{23}=0.2$	[12,14,27]
Meniscal horn attachments	1D springs	2000 N/mm	[12]
Meniscal transverse ligament	1D spring	900 N/mm	[12]
Patellar tendon	1D springs	4500 N/mm	[28]
Patellar retinaculum and PFL	1D springs (3 strands)	Medial portion: 15 N/mm Lateral portion: 9 N/mm	[15,29]
**ACL	non-linear 1D springs (3 strands)	Reference strain: 0.093 (aACL), 0.083 (pACL), stiffness (N): 1000 (aACL), 1500 (pACL)	[30]
PCL	non-linear 1D springs (3 strands)	Reference strain: -0.390 (aPCL), -0.120 (pPCL), stiffness (N): 2600 (aACL), 1900 (pPCL)	[30]
MCL	non-linear 1D springs (5 strands)	Reference strain: -0.017 (aMCL), 0.044 (cMCL), 0.049 (pMCL), stiffness (N): 2500 (aMCL), 3000 (cMCL), 2500 (pMCL)	[30]
LCL	non-linear 1D springs (3 strands)	Reference strain: 0.056, stiffness (N): 4000	[30]

PFL, patellofemoral ligament; ACL, anterior cruciate ligament; PCL, posterior cruciate ligament; MCL, medial collateral ligament; LCL, lateral collateral ligament

*For the meniscus, 1 represents the circumferential direction while 2, 3 represent the radial and axial directions, respectively.

**Different bundles of a ligament were combined to calculate the overall force-strain relationship.

Table 3: Root-mean-square errors (RMSE), Normalized root-mean-square errors (NRMSE) and Pearson correlation coefficients calculated from regression analysis for the measured and model-predicted contact variables corresponding to the tibiofemoral joint (top) and patellofemoral joint (bottom) during simulated weight-bearing activity.

Tibiofemoral joint							
		Medial compartment			Lateral compartment		
		Peak pressure (MPa)	Contact area (mm ²)	Contact force (N)	Peak pressure (MPa)	Contact area (mm ²)	Contact force (N)
Specimen 1	RMSE	0.45	69	56.9	0.43	163	70.1
	NRMSE	13.8%	14.4%	18.8%	12.8%	26.3%	11.8%
	r (Pearson correlation)	0.806	0.624	0.971	0.934	0.817	0.964
	p-value	0.00153	0.0300	1.49e-7	8.69e-6	0.00119	4.34e-7
Specimen 3	RMSE	0.25	97	101.3	0.73	72	33.7
	NRMSE	7.6%	18.1%	32.8%	21.7%	17.8%	12.4%
	r (Pearson correlation)	0.959	0.728	0.902	0.910	0.877	0.966
	p-value	8.18e-7	0.00856	6.08e-5	3.90e-5	1.80e-4	3.18e-7
Patellofemoral joint							
		Peak pressure (MPa)		Contact area (mm ²)		Contact force (N)	
Specimen 2	RMSE	0.40		64		13.2	
	NRMSE	13.7%		18.8%		5.6%	
	r (Pearson correlation)	0.875		0.985		0.997	
	p-value	4.26e-4		3.71e-8		3.82e-11	
Specimen 3	RMSE	0.33		60		15.5	
	NRMSE	13.7%		16.7%		5.8%	
	r (Pearson correlation)	0.711		0.982		0.983	
	p-value	0.0142		7.38e-8		5.76e-8	

Supplemental Material

DIRECT VALIDATION OF HUMAN KNEE-JOINT CONTACT MECHANICS DERIVED FROM SUBJECT-SPECIFIC FINITE-ELEMENT MODELS OF THE TIBIOFEMORAL AND PATELLOFEMORAL JOINTS

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The Tekscan 4000 pressure sensor was used to measure the joint contact pressure in the present study (sensor resolution: 62.0 sensels/cm², max pressure: 10.3 MPa). To minimize the performance degradation of the sensor, a new piece of the pressure sensor was calibrated and used in testing each specimen. It was calibrated with an improved custom method over the default method set by the manufacture (Brimacombe et al. 2009). Briefly, the custom method requires measurements of ~10 data points within the expected pressure range, and performs a multipoint cubic polynomial optimization using the least-squares criterion. Alternatively, the manufacture suggests that two known loads be measured (preferably 20% and 80% maximum expected pressure) prior to performing a power law interpolation (Equation 4.1b) over the measurements. The custom method was found to yield better measurement accuracies.

The sensor was calibrated with a Instron 8874 testing system and Instron Dynacell 2527-202 load cell in a “sandwich” set-up (see Figure S1 below). Rubber was chosen to evenly spread the load among the sensels and also because its Young’s modulus is comparable to cartilage (Lindley, 1966; Shepherd and Seedhom, 1999). The rubber pieces were carefully tailored such that they covered most of the sensor without exceeding the sensing area when loaded. The sensor was first conditioned for 2 minutes using a force approximately 20% higher than the expected peak force during subsequent cadaver knee tests, and the conditioning was repeated 3 times. Then the sensor was loaded under 11 different forces ranging from 150 N to 1500 N in a randomized order. For each data point, the Instron force slowly ramped up to the desired value in about 30 s and was then held constant for approximately 5 s before data were recorded, with a 2-minute interval between each run as recommended in the Tekscan user manual. Cubic polynomial optimization was then performed to obtain the calibration curve for the pressure sensor.

Figure S1 Configuration used to calibrate the Tekscan pressure sensor used in this study.

The surrounding padding areas of the sensor were trimmed and folded (Figure S2) so that it could be inserted into the tibiofemoral joint. Care was taken not to damage any sensels or the signal transmission wires. The sensor was inserted from the anterior side of the tibia plateau and located between tibia cartilage and menisci. A lab spatula was used to ensure the sensor to lay flat inside the joint without obvious wrinkles. The sensor was taped around the tibia firmly with duct tape and was confirmed to stay fixed on the tibia plateau (Figure 1D).

Figure S2: The pressure sensor was carefully trimmed and folded so it could be inserted into the joint.

Figure S3: Close-up of the Tekscan sensor inserted and fixed within the joint prior to the mechanical testing.

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Table S1 Effects of changes in the material properties of the soft tissues on contact peak pressure, area, force and centre of pressure calculated in the TFJ model. Comparisons were made against the nominal model.

		Medial							Lateral						
		Peak pressure		Contact area		Contact force		COP	Peak pressure		Contact area		Contact force		COP
		RMS (%) ^b	Sign ^b	RMS (%)	Sign	RMS (%)	Sign	Mean distance (mm) ^c	RMS (%)	Sign	RMS (%)	Sign	RMS (%)	Sign	Mean distance (mm)
Cartilage	Soft ^d	8.3	-	5.1	+	1.6	±	0.1	5.9	-	3.4	+	2.6	±	0.06
	Stiff	7.2	+	4.1	-	1.1	±	0.1	5.2	+	2.6	-	2.1	±	0.05
	Linear elastic ^e	3.5	±	0.7	-	2.4	±	0.1	7.6	±	2.3	-	2.2	±	0.09
Menisci	Soft	8.5	±	4.5	+	6.8	+	0.6	7.2	±	1.6	+	7.2	-	0.15
	Stiff	7.4	±	4.0	-	5.2	-	0.5	7.7	+	1.5	±	5.6	+	0.13
Meniscal constraints ^a	Soft	0.2	±	0.4	+	0.3	±	0.0	0.2	±	0.4	+	0.3	±	0.02
	Stiff	0.1	±	0.3	±	0.4	-	0.0	0.1	±	0.2	±	0.1	±	0.01
Ligaments	Soft	1.6	+	1.0	±	2.1	±	0.1	0.9	+	0.6	+	1.1	+	0.02
	Stiff	1.8	-	1.0	±	2.3	±	0.1	0.8	-	0.3	±	1.3	-	0.02
Friction	0.02	3.8	±	2.0	±	3.5	±	0.3	6.8	±	1.1	±	0.8	+	0.46

a Menisci constraints refer to meniscus horn attachments and transverse ligament.

b For contact pressure, force and area, the RMS difference (normalized by the nominal model results) was reported. Also the sign of the change was also reported: +/- means the new model predicted value larger/smaller than the nominal model, ± indicates larger in some frames while smaller in the others throughout the trial cycle.

c The average centre of pressure distance between the current model and the nominal model over the trial cycle,

d “Soft” and “Stiff” represent 20% below and above default material stiffness, respectively.

e linear elastic model of cartilage with Young’s modulus of 15 MPa and Poisson’s ratio of 0.45.

Table S2 Effects of changes in cartilage and meniscal thickness on contact peak pressure, area, force and centre of pressure calculated in the TFJ model. Comparisons were made against the nominal model.

		Medial							Lateral						
		Peak pressure		Contact area		Contact force		COP	Peak pressure		Contact area		Contact force		COP
		RMS (%) ^c	Sign ^c	RMS (%)	Sign	RMS (%)	Sign	Mean distance (mm) ^d	RMS (%)	Sign	RMS (%)	Sign	RMS (%)	Sign	Mean distance (mm)
fCartilage ^a	Thick ^e	12.3	±	6.1	±	8.1	±	0.8	18.9	±	7.1	+	7.9	±	0.54
	Thin	12.5	±	6.7	-	6.3	±	0.7	14.4	±	7.2	-	5.4	±	0.53
tCartilage ^b	Thick	22.2	±	8.3	±	15.7	±	1.5	28.3	±	11.0	+	10.3	±	0.86
	Thin	16.9	±	14.7	±	9.0	±	1.3	14.9	+	11.5	-	5.1	±	0.78
Menisci	Thick	7.5	±	3.6	±	6.9	-	0.6	6.9	±	5.7	+	5.5	+	0.11
	Thin	9.7	±	5.6	±	11.1	+	0.8	5.2	±	7.2	-	12.4	-	0.15

a Femur cartilage.

b Tibia cartilage of both medial and lateral compartments.

c For contact pressure, force and area, the RMS difference (normalized by the nominal model results) was reported. Also the sign of the change was also reported: +/- means the new model predicted value larger/smaller than the nominal model, ± indicates larger in some frames while smaller in the others throughout the trial cycle..

d The average centre of pressure distance between the current model and the nominal model over the trial cycle.

e “Thick” and “Thin” represent ±15% of thickness change from the default segmented geometries, respectively.

Table S3 Effects of changes in the material properties of the soft tissues on contact peak pressure, area, force and centre of pressure calculated in the PFJ model. Comparisons were made against the nominal model.

		Peak pressure		Contact area		Contact force		COP
		RMS (%) ^a	Sign ^a	RMS (%)	Sign	RMS (%)	Sign	Mean distance (mm) ^c
Cartilage	Soft ^c	7.9	-	7.6	+	1.1	-	0.2
	Stiff	9.1	+	4.1	-	0.8	+	0.1
	Linear elastic ^d	8.6	+	4.0	-	0.3	±	0.1
PFL and Retinaculum	Soft (20%)	1.9	+	1.5	+	3.2	+	0.0
	Stiff (20%)	1.8	-	1.2	-	3.2	-	0.0
	Soft (40%)	3.8	+	2.5	+	6.4	+	0.0
	Stiff (40%)	3.7	-	1.7	-	6.3	-	0.0
Patellar tendon	Soft	0.5	+	1.2	±	0.2	±	0.0
	Stiff	0.3	±	1.3	+	0.1	±	0.0
Friction	0.02	0.7	±	0.8	+	1.0	±	0.0

a For contact pressure, force and area, the RMS difference (normalized by the nominal model results) was reported. Also the sign of the change was also reported: +/- means the new model predicted value larger/smaller than the nominal model, ± indicates larger in some frames while smaller in the others throughout the trial cycle.

b The average centre of pressure distance between the current model and the nominal model over the trial cycle.

c “Soft” and “Stiff” represent 20% below and above default material stiffness, respectively.

d Linear elastic model of cartilage with Young’s modulus of 15 MPa and Poisson’s ratio of 0.45.

Table S4 Effects of changes in cartilage thickness and the insertion sites of the patellar tendon on the tibial tuberosity on contact peak pressure, area, force and centre of pressure calculated in the PFJ model. Comparisons were made against the nominal model.

		Peak pressure		Contact area		Contact force		COP
		RMS (%) ^a	Sign ^a	RMS (%)	Sign	RMS (%)	Sign	Mean distance (mm) ^b
Femoral cartilage	Thick ^c	8.1	±	2.5	±	1.5	+	0.6
	Thin	15.6	±	2.7	±	1.3	-	0.9
Patellar cartilage	Thick	16.6	±	6.0	±	3.7	+	1.3
	Thin	25.3	±	9.3	±	3.9	-	1.9
Patellar Tendon Tibial Insertion Points	Superior	0.6	+	0.9	+	0.9	+	0.0
	Inferior	1.2	+	0.9	+	1.9	+	0.0
	Medial	1.9	-	0.8	±	2.7	-	0.1
	Lateral	4.4	+	2.9	+	6.3	+	0.1

a For contact pressure, force and area, the RMS difference (normalized by the nominal model results) was reported. Also the sign of the change was also reported: +/- means the new model predicted value larger/smaller than the nominal model, ± indicates larger in some frames while smaller in the others throughout the trial cycle.

b The average centre of pressure distance between the current model and the nominal model over the trial cycle.

c “Thick” and “Thin” represent 15% increase and decrease of thickness change, respectively.

d “Superior”, “Inferior”, “Medial”, “Lateral” represent ~2 mm shift of the patellar tendon insertion points on tibia tuberosity in corresponding directions.