Journal of Biomechanics 49 (2016) 1111–1120

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

Effect of bone inhomogeneity on tibiofemoral contact mechanics during physiological loading

M.S. Venäläinen [a,b,*], M.E. Mononen [a], S.P. Väänänen [a], J.S. Jurvelin [a,c], J. Töyräs [a,c], T. Virén [b], R.K. Korhonen [a,c]

[a] Department of Applied Physics, University of Eastern Finland, PO Box 1627, FI-70211 Kuopio, Finland
[b] Cancer Center, Kuopio University Hospital, PO Box 100, FI-70029 KUH Kuopio, Finland
[c] Diagnostic Imaging Center, Kuopio University Hospital, PO Box 100, FI-70029 KUH Kuopio, Finland

ARTICLE INFO

Article history:
Accepted 16 February 2016

Keywords:
Knee joint
Bone
Quantitative computed tomography
Bone mineral density
Articular cartilage
Finite element analysis

ABSTRACT

It is not known how inhomogeneous mechanical properties of bone affect contact mechanics and cartilage response during physiological loading of the knee joint. In this study, a finite element model of a cadaver knee joint was constructed based on quantitative computed tomography (QCT). The mechanical properties of bone were altered and their effect on tibiofemoral contact mechanics and cartilage stresses, strains and pore pressures were evaluated during the first 20% of stance. For this purpose, models with rigid, homogeneous and inhomogeneous bones were created. When bone was modeled to be rigid, the resulting contact pressures were substantially higher in the medial side of the joint, as compared to the non-rigid bones. Similar changes were revealed also in stresses, strains and pore pressures throughout the cartilage depth at the cartilage–cartilage contact area. Furthermore, the mechanical response of medial tibial cartilage was found to be highly dependent on the bone properties. When Young's modulus in the model with homogeneous bone was 5 GPa, cartilage mechanical response approached to that of the model with inhomogeneous bone. Finally, when the apparent bone mineral densities were decreased globally in the inhomogeneous bone, stresses, strains and pore pressures were decreased at all layers of medial tibial cartilage. Similar changes were observed also in cartilage–cartilage contact area of the lateral compartment but with a lesser extent. These results indicate that during physiological loading Young's modulus of bone has a substantial influence on cartilage stresses and strains, especially in the medial compartment.

1. Introduction

Finite element (FE) analysis can provide information about knee joint mechanics difficult or impossible to measure *in vivo*. To date, it has been applied to a variety of cases ranging from optimal knee replacement implant design to observing the effect of meniscectomy or meniscal tear on cartilage stresses, strains and contact pressures (Fitzpatrick et al., 2012; Mononen et al., 2013; Pena et al., 2005). Some FE models utilize sophisticated material models such as fibril-reinforced poroviscoelastic model in articular cartilage (Mononen et al., 2015), which can accurately capture tissue response observed in mechanical experiments (Julkunen et al., 2007, 2008; Wilson et al., 2004). However, the way bones are considered in whole knee joint models varies largely, and compromises between practicality and accuracy are often needed since otherwise the model may easily become computationally too extensive.

Structural properties of bone vary between anatomical regions, affecting the mechanical properties of bone both at structural and organ level (Turner and Burr, 1993). However, in FE models of the knee joint, bones are often modeled as rigid since the stiffness of bone is much higher than the stiffness of the other tissues in the joint (Adouni et al., 2012; Akbarshahi et al., 2014; Donahue et al., 2002; Li et al., 1999; Mesfar and Shirazi-Adl, 2005; Pena et al., 2005; Yao et al., 2008). This simplification is typically justified based on the results from axial compression models where differences between contact parameters of models using rigid or deformable bones were found to be less than 2% (Donahue et al., 2002). However, this comparison was performed under an axial load equal to one body weight (800 N) and with only flexion and extension rotation constrained. In order to ensure that the rigid bone assumption has no effect on the results when simulating physiological loading such as gait, a study with more complex

* Corresponding author at: Department of Applied Physics, University of Eastern Finland, PO Box 1627, FI-70211 Kuopio, Finland. Tel.: +358 40 7458193; fax: +358 17 162585.
E-mail address: mikko.s.venalainen@uef.fi (M.S. Venäläinen).

http://dx.doi.org/10.1016/j.jbiomech.2016.02.033
0021-9290/

loading conditions is needed. To our knowledge, this kind of comparison has not been conducted yet.

In some previous knee joint models with deformable bones, the mechanical function of bones was accounted for by considering them as homogeneous structures with linear elastic material behavior (Guess et al., 2010; Mootanah et al., 2014; Shirazi and Shirazi-Adl, 2009). However, the selected Young's modulus for representing the whole bone varies greatly and it typically ranges from 1 to 20 GPa (Guess et al., 2010; Mattei et al., 2014; Mootanah et al., 2014; Nagasaka et al., 2003). In our experience (Venalainen et al., 2014), the most realistic value for Young's modulus should be within these two extremes that represent the structural stiffness of the trabecular and the cortical bone. In any case, modeling the mechanical function of bone as a single homogeneous entity is an approximation.

With the aid of computed tomography (CT), inhomogeneous density-specific mechanical properties of bone can be obtained and implemented into FE knee joint models (McErlain et al., 2011; Papaioannou et al., 2010). With quantitative computed tomography (QCT), an estimate for element-wise bone mineral density (BMD) and, thus, also elastic modulus can be computed (McErlain et al., 2011; Tuncer et al., 2014; Vaananen et al., 2011) using density–elasticity relationships widely reported in the literature (Morgan et al., 2003; Snyder and Schneider, 1991). Therefore, this approach enables obtaining inhomogeneous mechanical properties for bone with clinical-level imaging resolution.

Our previous 2D simulations of the knee joint function under physiological level loading suggested that the potential pathophysiological state, and especially the trabecular structure of articulating bones affect the mechanical response of articular cartilage to loading (Venalainen et al., 2014). Briefly, we observed that increased porosity and decreased apparent stiffness of trabecular bone resulted in decreased levels of stresses, strains and pore pressures, especially in the lateral tibial cartilage. Considering that high levels of stresses and strains could be related to the onset and progression of osteoarthritis (Guilak, 2011; Miyazaki et al., 2002), this finding is consistent with the hypothesis that osteoporosis and osteoarthritis may be inversely related (Foss and Byers, 1972; Hart et al., 1994). However, due to limitations of two-dimensional (2D) modeling, it is difficult to generalize these results. Therefore, a study with varying properties of bone in a three-dimensional (3D) FE model of a knee joint is needed.

In the present study, the main aim was to vary the mechanical properties of the bone using FE modeling and evaluate their effect on the contact mechanics and mechanical response of articular cartilage during stance phase of walking. For this purpose, a knee joint model including bones, cartilage and menisci was constructed based on QCT scans. In order to study the effect of different bone properties on simulation outcomes, FE models with rigid, homogeneous elastic and inhomogeneous elastic bone (element-wise properties computed using typical density–elasticity relationships for bone) were created. In all cases, tibiofemoral contact mechanics and mechanical response of articular cartilage were analyzed under loading conditions typical to the first peak load during walking. Finally, in the model with inhomogeneous bone properties, the apparent bone mineral densities were decreased globally in order to evaluate if changes in the pathophysiological state of bone (*e.g.*, due to osteoporosis) affect resulting stresses, strains and pore pressures in the articular cartilage.

2. Methods

2.1. Model construction

A right knee joint of a male cadaver was imaged using a lower extremity CT protocol (Siemens SOMATOM Definition Edge, Siemens, Forcheim, Germany) along with a QCT phantom (Model 3 CT Calibration Phantom, Mindways Software Inc., Austin, TX, USA). From CT data (Fig. 1a), all remaining tissues, *i.e.*, bones (femur and tibia), articular cartilage and menisci were segmented using image processing software (Mimics v.12.3, Materialise, Leuven, Belgium). Because synovial fluid was not present during imaging, cartilage and menisci could be easily segmented manually. Bones, on the other hand, were segmented with a threshold value of 400 HU and after thresholding, voids inside bone were filled with cavity fill to obtain a uniform geometry. The segmented tissues were converted into solid geometries using a custom Matlab script (MATLAB R2012a, The MathWorks Inc., Natick, MA, USA) and imported into a FE modeling package (Abaqus v6.14, Dassault Systèmes, Providence, RI, USA) in which all FE meshes were created (Fig. 1b). See Supplementary Appendix A for more details about the studied knee joint and utilized FE meshes.

2.2. Material properties and model variations

Articular cartilage and meniscus were modeled as a fibril-reinforced poroviscoelastic material (FRPVE) (Julkunen et al., 2007; Mononen et al., 2012; Wilson et al., 2004) and as a linear elastic, transversely isotropic material (Donahue et al., 2002; Halonen et al., 2013), respectively. More information about the material properties for cartilage and meniscus have been provided in Supplementary Appendix A.

The mechanical response of articular cartilage in the knee was studied with the following three material properties for bones (Fig. 1d–f):

1. Rigid bone,
2. Homogeneous elastic bone, and
3. Inhomogeneous elastic bone.

In the model with rigid bone (Fig. 1d), the mechanical function of the bone was modeled completely using rigid body constraints whereas in all other cases, bones were assumed to be isotropic with Poisson's ratio of 0.3 (Wirtz et al., 2000). In the model with homogeneous bone (Fig. 1e), various values of Young's modulus were tested within the range of 2–15 GPa in order to reveal which modulus gave results closest to the inhomogeneous bone model. Instead, in inhomogeneous bone model, Young's modulus was assigned separately for each element based on the CT-derived density–elasticity relationships available in the literature (Morgan et al., 2003; Snyder and Schneider, 1991). Apparent BMD for each voxel (Fig. 1c) was obtained by assuming a relationship between the Hounsfield unit (HU) values and known densities of the reference materials (see Vaananen et al. 2011 for further details). After calibration, apparent densities were converted to Young's moduli depending on the bone and anatomical region. For femur, density–elasticity relationship from Morgan et al. (2003) was utilized. For tibia, the densities were split into three categories (trabecular, cortical and intermediate) and converted to Young's moduli using a category-specific rule (Tuncer et al., 2014). For trabecular ($\rho_{app} < 0.37$ g/cm^3) and cortical regions ($\rho_{app} > 1.5$ g/cm^3), bone-specific density–elasticity relationships from Morgan et al., (2003) and Snyder and Schneider (1991) were used, respectively. For intermediate densities, linear interpolation similar to Tuncer et al., (2014) was applied. Young's modulus for each element (Fig. 1f) was assigned with a custom Matlab script utilizing a mapping strategy based on numerical integration of Young's modulus continuum field over the element's volume (Taddei et al., 2004, 2007).

In addition to the inhomogeneous bone model described above, three additional inhomogeneous models with modified bone densities were created. In these models the apparent BMDs were decreased globally by 15%, 30% and 45%. In all cases, the density–elasticity conversions and steps for elasticity mapping were done similarly as with the original densities.

2.3. Simulations and boundary conditions

The mechanical response of articular cartilage was simulated under loading conditions typical to the first 20% of stance (first peak load during walking). Kinematic data of a single gait cycle during treadmill walking (Fig. 2a–b), obtained with a walking speed of 0.67 m/s, was acquired from a previous study (Kozanek et al., 2009). In addition to kinematic data, axial force data of a single gait cycle was also obtained from literature (Komistek et al., 1998). Both data were implemented as boundary conditions and loads into a reference point, similarly as before (Mononen et al., 2015), located at the center of transepicondylar axis of femur (Fig. 2(a) (Most et al., 2004). Previously, the bone–cartilage interface of femur was coupled to the kinematics of the reference point to simulate this motion (Mononen et al., 2015). In this study, however, the proximal end of femur was used to account also for the overall deformation of femur during simulations. The global coordinate system was aligned with anterior–posterior and medial–lateral axes of tibia so that the input kinematics could be directly applied as relative motion of the femur with respect to the tibia. By using the coordinate reference frame in Fig. 2a), flexion–extension, internal–external and varus–valgus motions were defined as rotation of femur in the *zy*-, *xz*- and *xy*-planes, respectively. Similarly, anterior–posterior, medial–lateral and distal–proximal translations were defined as movement of femur along the *z*-, *x*- and *y*-axes, respectively. Only varus–valgus angle was