Journal of Biomechanics 48 (2015) 1356–1363

Contents lists available at ScienceDirect

Journal of Biomechanics

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

Influence of meniscus shape in the cross sectional plane on the knee contact mechanics

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ARTICLE INFO

Article history:
Accepted 7 March 2015

Keywords:
Meniscus geometry
Meniscal extrusion
Open Knee model
Slope angle

ABSTRACT

We present a three dimensional finite element analysis of stress distribution and menisci deformation in the human knee joint. The study is based on the Open Knee model with the geometry of the lateral meniscus which shows some degenerative disorders. The nonlinear analysis of the knee joint under compressive axial load is performed. We present results for intact knee, knee with complete radial posterior meniscus root tear and knee with total meniscectomy of medial or lateral meniscus. We investigate how the meniscus shape in the cross sectional plane influences knee-joint mechanics by comparing the results for flat (degenerated) lateral and normal medial meniscus. Specifically, the deformation of the menisci in the coronal plane and the corresponding stress values in cartilages are studied. By analysing contact resultant force acting on the menisci in axial plane we have shown that restricted extrusion of the torn lateral meniscus can be attributed to small slope of its cross section in the coronal plane. Additionally, the change of the contact area and the resultant force acting on the menisci as the function of compressive load are investigated.

1. Introduction

Menisci play important roles in the knee, by transmitting the force from the femur to the tibia. Their function is to enhance the distribution of forces in the knee. The load from the body weight is transferred by femoral condyles to the menisci, yielding their motion in the radial direction. However, this motion is constrained by the attachments of the anterior and posterior horns to the tibia. The collagen fibres on the circumference convert axial loads into hoop stress (McDermott et al., 2008), which counteracts further displacements (Jones et al., 1996). Disruption of the circumferential fibres causes the inability of the meniscus to generate hoop stress. This results in meniscal extrusion, which entails a shift of the central part of the meniscus by more than 3 mm, as seen on coronal MRI (Costa et al., 2004; Ding et al., 2007).

In knees with osteoarthritis the menisci are most often changed, frequently ruptured, emaciated and displaced, which indicates a strong association between the disorder and the meniscus geometry (Bhattacharyya et al., 2003; Englund et al., 2008; Hunter et al., 2006). A connection has been demonstrated between meniscal body extrusion and the development of knee osteoarthritis and pain, caused by the accompanied cartilage degeneration (Choi et al., 2014; Wenger et al., 2012), osteochondral defects (Shirazi and Shirazi-Adl., 2009), bone marrow lesions (Englund et al., 2010) and synovitis (Grainger et al., 2007).

The possible causes of extrusion present in knees with abnormal menisci have not been explained so far. It was suggested that, in addition to the damaged menisci and ruptured circumferential fibres, other factors may be associated with meniscal extrusion in the osteoarthritic knee, e.g. varus alignment of the leg, or the narrowing of the joint space (Lee et al., 2011). Although the influence of meniscal damage on its extrusion is very well documented, very little is known about factors not involved in injury, which can cause some changes in meniscal position (Erquicia et al., 2012; Lee et al., 2011; Petersen et al., 2014; Wenger et al., 2013). It has not been clarified whether extrusion is the cause or effect of osteoarthritic changes (Chan et al., 2008).

Clinical studies are not sufficient to determine predictors of meniscal extrusion, because they are typically cross-sectional and retrospective, leaving the causality analysis of the meniscal displacement in the sphere of speculation (Wenger et al., 2013).

The relationship between meniscus tears and the development of knee osteoarthritis requires further study (Mascarenhas et al.,

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http://dx.doi.org/10.1016/j.jbiomech.2015.03.002
0021-9290/

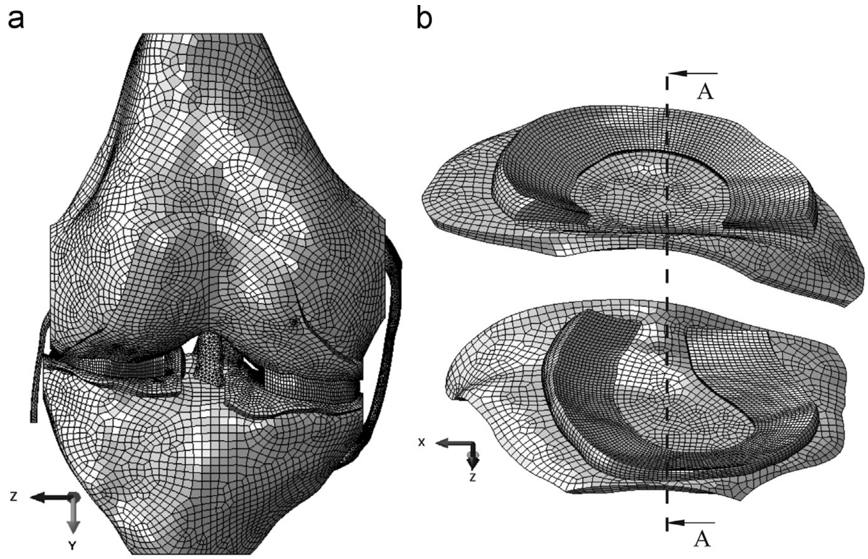

Fig. 1. Knee joint, FEM discretization: (a) anterior view, (b) menisci and tibial cartilages.

2012). Several biomechanical studies of meniscus tears and extrusion have been reported in the literature (Allaire et al., 2008; Bao et al., 2013; Hein et al., 2010; Marzo and Gurske-DePerio, 2009). However, none of these studies was carried out on a model with abnormal geometry of lateral meniscus. Better understanding of the relationship among meniscus tear morphology, extrusion and cartilage degeneration is essential to devise an appropriate treatment plan. To investigate this issue we use finite element method. We study the resultant contact force acting on the menisci in axial plane and its connection to the shape of the menisci in their coronal cross plane. We postulate that one of the factors contributing to the extrusion motion is the angle between cartilages surfaces and menisci surfaces.

2. Materials and methods

2.1. Knee joint geometry

The geometry of the knee joint is obtained from Open Knee project, version 1.0.1.202 (Erdemir and Sibole, 2010; Sibole et al., 2010), where the geometry was manually extracted from Magnetic Resonance (MR) images of the right cadaver knee (70-year-old female donor, height 1.68 m, weight 77.1 kg). The knee model (Fig. 1a) consists of the femur, tibia, menisci, articular cartilages, anterior cruciate ligament (ACL), posterior cruciate ligament (PCL), medial collateral ligament (MCL) and lateral collateral ligament (LCL). The geometry of all the above components of the Open Knee model was used as-is in this study.

The Open Knee model offers two slightly different menisci geometries available either in IGES format or mesh file i.e. INPUT format. The main difference between them is the thickness. In this paper, we used the geometry of the menisci with thickness as from the input file that is the actual anatomic thickness and was applied also in Open Knee calculations.

To show the influence of menisci geometry on the knee contact mechanics we present results from the intact knee, knee with complete radial posterior medial meniscus root tear (PMMRT), knee with total meniscectomy of medial meniscus, knee with complete radial posterior lateral meniscus root tear (PLMRT) and knee with total meniscectomy of lateral meniscus.

In our model we compare the results for normal medial side and lateral side with abnormally flat lateral meniscus in the middle region. To explain the abnormality, which is present in the Open Knee geometry, we consider the meniscus geometry in their coronal plane. Fig. 2a and b illustrates the cross section geometry of lateral and medial meniscus respectively in section A–A, see Fig. 1b. That geometry can be approximately described by slope (Haut et al., 2000; Huang et al., 2002; Vrancken et al., 2014) or meniscus–cartilage angle (Sturnick et al., 2014). In our study we measure the slope angle (different from the slope defined by Haut et al. (2000)) between the contact surface of the meniscus with the femoral cartilage and the contact surface of the meniscus with the tibial plateau. The obtained values, measured exactly from Open Knee without any modifications, are approximately equal: for lateral meniscus $\alpha_L \approx 7^\circ$ (Fig. 2a) and for medial meniscus $\alpha_M \approx 25^\circ$ (Fig. 2b). The slope angle for the lateral meniscus is outside the range usually reported in the literature, for instance $40.5 \pm 6.3^\circ$ in sagittal plane (Sturnick et al., 2014). On the basis of MRI attached to the Open Knee model, we speculate that abnormally flat geometry of the lateral meniscus is a consequence of degenerative disorders.

2.2. Material properties

Given the large difference in stiffness between bone and cartilage, for the purposes of this analysis bones were treated as rigid structures (Kazemi et al., 2013). The material properties of soft tissues were chosen from values identified in the literature. The articular cartilages are assumed to be made of single-phase linear elastic, isotropic material with Young's modulus $E = 10$ MPa, obtained for time-independent compressive load (Shepherd and Seedhom, 1999), and a Poisson ratio of $\nu = 0.45$ (Li et al., 2001). For menisci a linear elastic, transversely isotropic material is used with: elastic moduli $E_\theta = 120$ MPa in the circumferential direction and $E_z = E_r = 20$ MPa in the axial and radial directions, Poisson's ratio $\nu_{rz} = 0.2$ in the circumferential direction and $\nu_{r\theta} = \nu_{z\theta} = 0.3$ in both axial and radial directions, the shear modulus $G_{rz} = 8.33$ MPa, $G_{r\theta} = G_{z\theta} = 57.7$ MPa (Haut Donahue et al., 2003; Yang et al., 2010).

Meniscal horn attachments were modelled using linear springs connecting each internal node on the meniscal horn surfaces with an appropriate node on the tibial plateau. The lengths L_i, cross-sectional areas A_i and Young's modulus E_i of each meniscal attachment are presented in Table 1, following experimental data from Hauch et al. (2010). The stiffness for tension and compression, S_i, of each meniscal attachment and corresponding spring stiffness k_i are calculated respectively as

$$S_i = \frac{E_i A_i}{L_i}, \tag{1}$$

$$k_i = \frac{S_i}{N_i}, \tag{2}$$

where N_i is the number of internal nodes (springs) on the meniscal horn face for a chosen meniscal attachment. The calculated values of S_i and k_i are presented in Table 1. In previous papers, for instance, (Yang et al., 2010; Zielinska and Haut Donahue, 2006), the stiffness of horn attachments was often assumed as 2000 N/mm, significantly higher than calculated from (1). However, the values from Table 1 are in good agreement with experimental work by Seitz et al. (2012).

The ligaments were assumed as transversely isotropic hyperelastic material (Weiss et al., 1996) with the strain-energy function

$$\Phi = C_{10}(\bar{I}_1 - 3) + \frac{1}{D_1}(J-1)^2 + F(\lambda), \quad J = \det\mathbf{F}. \tag{3}$$

that was implemented as UMAT into Abaqus, (Abaqus, 2012). Here $\bar{I}_1 = \mathrm{tr}\bar{\mathbf{B}} = \mathrm{tr}\bar{\mathbf{F}}\bar{\mathbf{F}}^{\mathrm{T}}$ denotes the first invariant of the (modified) left Cauchy-Green tensor with $\bar{\mathbf{F}} = J^{-(1/3)}\mathbf{F}$ as the modified deformation gradient, cf. for instance Holzapfel (2001). The term $F(\lambda)$ denotes the fibre family strain energy (Peña et al., 2006b;