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Comparison of stress on knee cartilage during kneeling and standing using finite element models

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ABSTRACT

Kneeling is a common activity required for both occupational and cultural reasons and has been shown to be associated with an increased risk of knee disorders. While excessive contact pressure is considered to be a possible aggressor, it is not clear whether and to what extent stress on the cartilage during kneeling is different from that while standing. In this study, finite element models of the knee joint for both kneeling and standing positions were constructed. The results indicated differences in high-stress regions between kneeling and standing. And both the peak von-Mises stress and contact pressure on the cartilage were larger in kneeling. During kneeling, the contact pressure reached 4.25 MPa under a 300 N compressive load. It then increased to 4.66 MPa at 600 N and 5.15 MPa at 1000 N. Changing the Poisson's ratio of the cartilage, which represents changes in compressibility caused by different loading rates, was found to have an influence on the magnitude of stress.

1. Introduction

Kneeling is a daily requirement for many occupations such as mining, baggage handling, building construction and agricultural works [1–4]. It also plays a critical role in Middle Eastern and Asian society for religious or cultural reasons [5–7]. Previous studies have shown that kneeling and crouching can increase the risk of knee osteoarthritis [8–10]. Recent research further revealed the dose–response relationship between kneeling and knee disorders [11,12]. Since the exact mechanisms for these diseases are not clear, excessive cartilage stress is considered to be a possible explanation. Through in vitro experiments it has been demonstrated that even a pressure as low as 4.5 MPa can induce cartilage apoptosis [13], and clinical observations reported high pressure being associated with radiographic and biochemical changes of cartilage tissue [14]. Therefore, determining the stress on the cartilage is fundamental to understanding and preventing knee injuries associated with kneeling.

Finite element (FE) analysis serves as a useful tool for observing the distribution of stress/strain across the knee joint and investigating how different material properties influence its mechanical status [15,16]. The stress and strain of ligaments can also be predicted if solid models of these parts were introduced [17,18]. Changes in cartilage stress associated with different flexion angles have also been reported [19,20]. Although previous publications have calculated the stress from full extension to deep flexion, very few have addressed this problem with respect to kneeling. A recent experimental study by Hofer et al. measured the tibio-femoral contact area and contact pressure under different kneeling angles [21]. However, this did not include details of patella–femur contact, which is a major load transmission path for kneeling, especially with flexion angles approaching 90° [22]. It is also important to note that the reported stress on cartilage can vary dramatically amongst published studies. While this is understandable when considering individual differences and different measuring methods, it makes comparison difficult and therefore prevents us from answering a basic but important question: whether and to what extent stress on the cartilage during kneeling is different from that while standing.

The purpose of this study was to compare the stress distributions on knee joint cartilage between kneeling and standing positions. In this study, two finite element models for both postures are presented and the mechanical status of the cartilage is investigated. The models were established from magnetic resonance (MR) images of the same subject and assigned with identical material properties.

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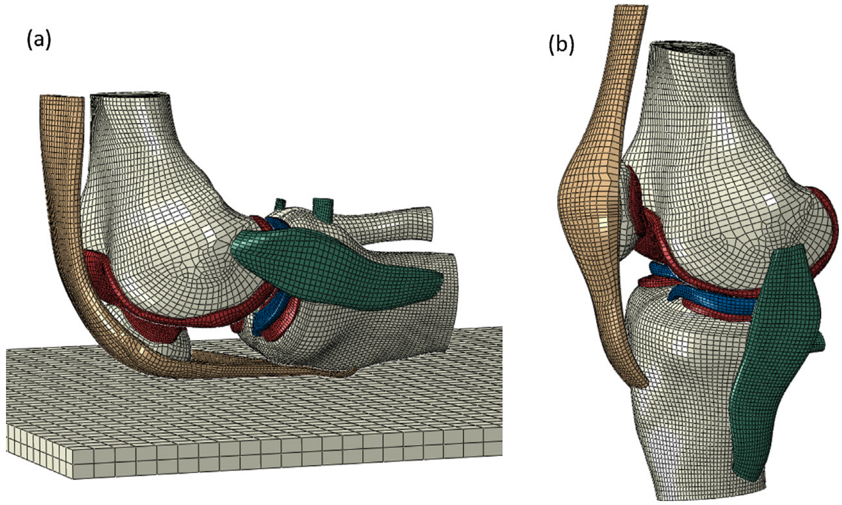

Fig. 1. Finite element model of the knee joint. (a) Kneeling position and (b) standing position.

2. Method

The MR images were obtained from a 26 year old healthy male volunteer. Scanning was carried out on the right knee when fully extended and when flexed to 90°. During scanning, the volunteer was asked to remain relaxed in order to eliminate the influence of pre-stressing.

The images of the flexed knee was used to build a geometrical model using commercial software MIMICS, including the femur, fibula, tibia, patella, cartilage, medial and lateral menisci, patellar tendon, and ligaments. The points of attachment of biceps and semimembranosus were also identified. All of these parts were smoothed and further modified in Rapidform XOR. And the thickness maps of the cartilages were obtained at this step. To facilitate the comparison, when building the standing model, rigid translation for the bony parts and their attached cartilage was used instead of building new ones. MR images of the extended knee served as the reference for this procedure. By this approach, possible incoherence resulting from artificial errors was avoided during geometrical modeling.

Following these procedures, all parts were meshed with hexahedral elements in Abaqus. According to previous methods [16], the element size within the cartilage surface was set to be less than 2 mm. Also, the cartilage was set to be two elements thick (Fig. 1). After reducing the element size by half and increasing the thickness to three elements, the change of von-Mises stress and contact pressure was within 5% for both kneeling and standing models. Therefore the elements in our models were dense enough for analysis.

The material properties were assigned according to relevant literature. The bony parts were set as cortical bone with an elastic modulus of 20,000 MPa and Poisson's ratio of 0.3 [23,24]. The inner structures, like trabecular bone, were ignored since they would not interact directly with cartilage. The cartilage was assumed to be isotropic elastic with an elastic modulus of 10 MPa and Poisson's ratio ranging from 0.05 to 0.45. When near 0.5, it simulated incompressible behavior during instantaneous loading; and for a Poisson's ratio near 0, compressible behavior under prolonged loading was simulated [15,25,26]. The menisci were modeled as a transversely isotropic material, with a radial and axial modulus of 20 MPa and a circumferential modulus of 140 MPa. The in-plane Poisson's ratio was 0.2 while the out of plane Poisson's ratio was 0.3 [16,27]. All ligaments were assumed to behave as hyperelastic materials. The stress–strain relationship of each ligament was obtained from previous FE and experimental studies [20,28] and was input to Abaqus as test data. The compressive property of the patellar tendon was assumed to be similar to tension since it bore compressive loads from contact with the ground. The other ligaments were defined to have only a very small compressive modulus, less than 5% of their tension behavior, as a compromise between simulating their incapability of bearing compressive load and the difficulty of convergence from assigning zero compressive modulus to a solid part. An Ogden hyperelastic model with second-order energy potential was used in this study. And a material evaluation was carried out to make sure this description can fit the stress–strain relationship. According to literature [17], an initial strain was also applied to the ligaments (Table 1).

The inner surfaces of all cartilages and the connecting interfaces of the ligaments were tied to corresponding bones. And the two horns of the menisci were fixed to the tibial plateau. The interaction between cartilages and menisci was set as frictionless contact. And the possible contact relationship between ligaments and bony structures, including contact between the patellar tendon and femur, was also defined. The contact model in this study was finite-sliding, surface-to-surface contact with a "hard" pressure–overclosure relationship in effect.

For the kneeling model, the femur was fixed in space, and the tibia set totally free. The ground plane was permitted to only move perpendicularly, with the other five degrees of freedom restricted. The end surface of the patellar tendon was constrained as it can only be displaced parallel to the direction of the femur. Muscle forces (quadriceps 215 N, biceps 31 N, and semimembranosus 54 N) were used to simulate the physiological loading of the knee joint [21,29], and an additional compressive load of up to 1000 N being applied to the joint. For the standing model, the loads and boundary conditions were kept the same, except for with the tibia. A rigid plane was tied to the end surface, through which the flexion angle was restricted with the other degrees of freedom unconstrained. And

Table 1
Initial strain of the ligaments (%).

aACL	pACL	aPCL	pPCL	aLCL	pLCL	aMCL	pMCL
0.06	0.1	0.0	0.0	0.0	0.0	0.04	0.04

a, anterior; p, posterior.