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Deformation of articular cartilage during static loading of a knee joint – Experimental and finite element analysis

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ABSTRACT

Novel conical beam CT-scanners offer high resolution imaging of knee structures with *i.a.* contrast media, even under weight bearing. With this new technology, we aimed to determine cartilage strains and meniscal movement in a human knee at 0, 1, 5, and 30 min of standing and compare them to the subject-specific 3D finite element (FE) model. The FE model of the volunteer's knee, based on the geometry obtained from magnetic resonance images, was created to simulate the creep. The effects of collagen fibril network stiffness, nonfibrillar matrix modulus, permeability and fluid flow boundary conditions on the creep response in cartilage were investigated. In the experiment, 80% of the maximum strain in cartilage developed immediately, after which the cartilage continued to deform slowly until the 30 min time point. Cartilage strains and meniscus movement obtained from the FE model matched adequately with the experimentally measured values. Reducing the fibril network stiffness increased the mean strains substantially, while the creep rate was primarily influenced by an increase in the nonfibrillar matrix modulus. Changing the initial permeability and preventing fluid flow through noncontacting surfaces had a negligible effect on cartilage strains. The present results improve understanding of the mechanisms controlling articular cartilage strains and meniscal movements in a knee joint under physiological static loading. Ultimately a validated model could be used as a noninvasive diagnostic tool to locate cartilage areas at risk for degeneration.

1. Introduction

Occupying nearly 80% of the wet weight (Buckwalter et al., 2005), interstitial fluid is the dominant constituent of articular cartilage. Primarily with the help of collagen network (60–80% of the dry weight (Buckwalter et al., 2005; Mow et al., 1990)), interstitial fluid pressure helps the tissue to carry high instantaneous loads (Mow et al., 1990). During creep loading the fluid slowly redistributes and proteoglycans (PGs) (20–40% of the dry weight (Buckwalter et al., 2005; Mow et al., 1990)) take the main role in resisting compressive loads at equilibrium, *i.e.* when the fluid flow has ceased (Bader et al., 1992; Mansour, 2004). This creep behavior has been extensively studied in *in vitro* studies (Armstrong and Mow, 1982; Ateshian et al., 1997; Boschetti, 2004; Kempson et al., 1970; Li et al., 2008; Mow et al., 1980) and *in situ* studies (Herberhold et al., 1999; Kääb et al., 1998). Yet, to our knowledge, the *in vivo* creep behavior of cartilage within a knee joint has only been examined during short-term loading (Eckstein et al., 2005; Hosseini et al., 2010, 2012).

The finite element (FE) modeling allows the simulation of internal strains and stresses within the knee joint during daily activities such as standing or walking (Bae et al., 2012; Bendjaballah et al., 1995; Donahue et al., 2002; Peña et al., 2006; Shirazi et al., 2008). It can also be used to study mechanisms that control tissue responses in mechanically loaded knee joints. In terms of the constituents of articular cartilage, past 3D models of knee joints have included collagen fibrils (Adouni et al., 2012; Gu and Li, 2011; Halonen et al., 2013; Mononen et al., 2012; Shirazi et al., 2008; Shirazi and Shirazi-Adl, 2009), strain-dependent permeability (Halonen et al., 2013; Mononen et al., 2012), nonfibrillar matrix mimicking the effect of PGs (Halonen et al., 2013; Mononen et al., 2012; Shirazi et al., 2008; Shirazi and Shirazi-Adl, 2009) and fluid flow through free surfaces (Gu and Li, 2011). To our knowledge, only Kazemi et al. (2011) simulated the creep behavior of articular cartilage in a 3D knee joint geometry, with their main focus on the analysis of fluid pressure under a load of 300 N. However, they did not investigate how the fluid flow through noncontacting surfaces or different mechanical properties (specifically the nonfibrillar matrix, collagen fibrils and permeability) of cartilage affect the *in vivo* creep response under physiological joint loads.

In computed tomography arthrography (CTa), a contrast agent is injected into the synovial cavity of a joint, causing a clear contrast between cartilage and its surroundings, thus, giving

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geometrical information of the tissue when imaged (De Filippo et al., 2009). In human cadaver studies, CTa has been shown to assess cartilage thickness with even better accuracy than MRI (El-Khoury et al., 2004; Llopis et al., 2012).

Being less tightly attached to the joint capsule, the lateral meniscus is more mobile than the medial meniscus (Masouros et al., 2008). While the meniscal movement has been well studied during flexion (Thompson et al., 1991; Tienen et al., 2005; Vedi et al., 1999), the displacement of menisci is not well known during static loading of a human knee.

This study aims to evaluate *in vivo* strains of articular cartilage in a knee joint during 30 min of standing and compare the results to the subject-specific FE model. The model is further used to study the effect of the collagen fibrils, nonfibrillar matrix and internal fluid flow and flow through noncontacting surfaces on the creep response in cartilage. This study provides new information about the mechanical behavior of articular cartilage and meniscal movement in a knee joint during static loading, while taking a step towards validating the FE model of a knee joint.

2. Materials and methods

The workflow of both the experiment and the simulations is represented in Fig. 1.

2.1. Experimental setup

In order to determine the cartilage thickness (El-Khoury et al., 2004; Llopis et al., 2012), an ioxaglate-based isotonic contrast agent solution of Hexabrix© (Mallinckrodt Inc., USA) (53% of the total volume) and distilled water (47% of the total volume) was prepared to match the osmolarity of synovial fluid (300 mOsm/L) in the knee joint (Bertram and Krawetz, 2012). The left knee of a healthy 28-year-old volunteer ($m=82$ kg) was imaged using CTa (voxel size $=0.2\times0.2\times0.2$ mm^3) (Verity©, Planmed, Finland). First, the imaging was conducted during light cartilage–cartilage contact ($12.8\pm2.8\%$ of body weight, BW), then immediately after introducing $47.1\pm2.3\%$ of BW, then subsequently 1, 5 and 30 min after the initial contact. A creep time of 30 min was chosen because the preliminary test showed that it was the last time point when the cartilage surfaces were distinguishable. It took ~25 s to complete each imaging step and approximately 35 s to prepare for the next one. Imaging under light cartilage–cartilage contact was conducted in order to obtain information about initial cartilage thicknesses and locations of menisci. A custom made Linux program and a Wii Balance Board© (Nintendo, Japan) were used to monitor the weight distribution during the experiment. In order to control the load on the knee joint the subject wore harnesses fastened to the ceiling (Fig. 2a). A quick-release lever was used to increase the load from 12.8% to 47.1% of BW instantaneously.

In CTa images, femoral and tibial cartilages as well as menisci were manually segmented using Mimics v.15.01 (Materialise, Belgium). At femoro-tibial contact areas the cartilage–cartilage interface was determined to be in the halfway point of contrast agent film between articulating surfaces (Fig. 2b, blue line). Local cartilage thickness was determined in Matlab v. R2012a (MathWorks Inc., USA) by calculating the minimum distance between the vertices on the cartilage surface and the cartilage–bone interface of the 3D surface mesh (Stammberger et al., 1999). The image stacks were manually co-registered on Analyze v. 10.0 (Biomedical Imaging Source, MN) by rotating the stacks so that the edges of tibia bone in the reference stack (before the load application) matched to those in all subsequent measurement points. This was done in order to enable the comparison of cartilage thickness maps in the analysis of engineering strains in cartilage (cartilage deformation divided by the initial thickness). CTa resolution and the subject's relatively thin femoral cartilage caused some areas of the femoral cartilage to appear only 5 pixels thick in the images, resulting in one pixel to cause a maximum error of 20% of the tissue thickness (average error being 14.3%). The corresponding maximum and average errors in the tibial cartilage were 6.7% and 4.4%, respectively. Therefore only the thicker tibial cartilage was analyzed.

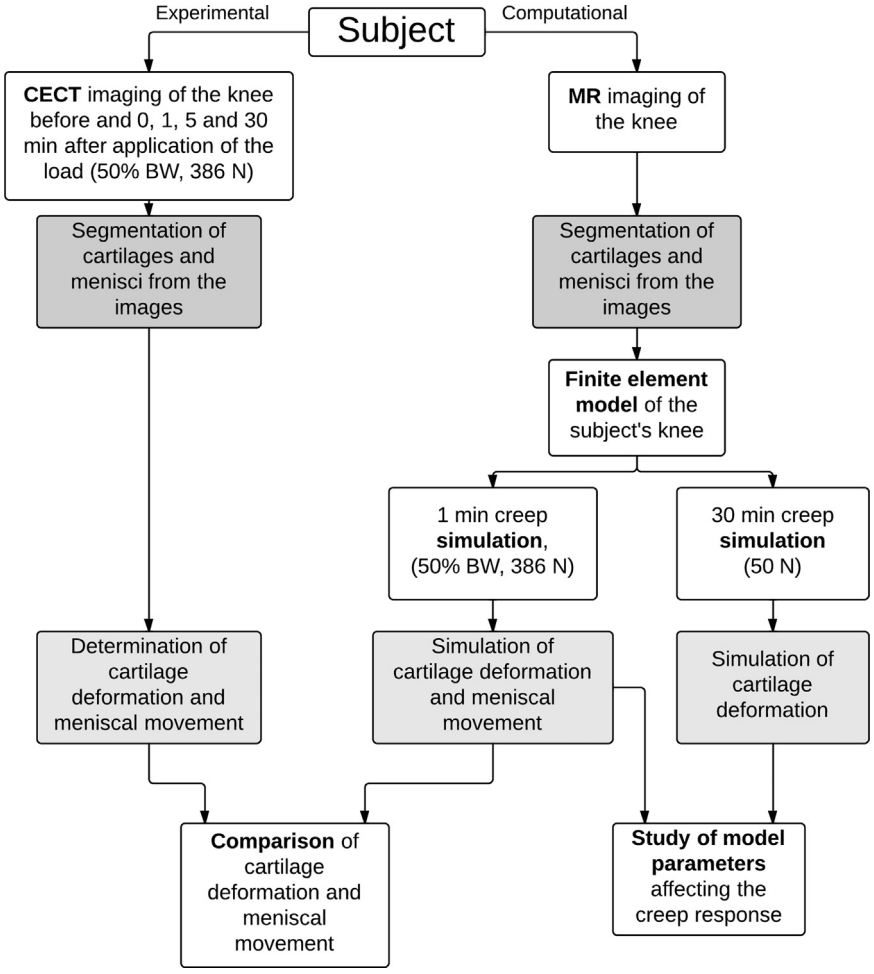

Fig. 1. Workflow chart of the experiment and FE simulations.

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