DEPTH DEPENDENT CREEP RESPONSE OF HUMAN ARTICULAR CARTILAGE DURING COMPRESSION: EXPERIMENTAL TESTING AND SIMULATION

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INTRODUCTION

The function of articular cartilage is to support and distribute loads and to provide lubrication in the diartroidal joint. Articular cartilage is composed of a charged solid matrix phase consisting of charged proteoglycan macromolecules and collagen fibers, an interstitial fluid phase, and an ion phase. Biomechanical models of increasing complexity have been developed to interpret the anisotropic and non linear properties in compression and tension, due to the heterogeneous and depth dependent composition and structure of articular cartilage. Experimental investigation of cartilage biomechanics is typically performed by using confined or unconfined compression, or indentation. The equilibrium response of articular cartilage is satisfactorily described, for small deformations, by the homogeneous isotropic elastic model. In particular, unconfined and confined compression are commonly used to evaluate the Young's modulus (E) and the aggregate modulus (H_A), respectively. Recently, video microscopy has been used during unconfined compression for the optical evaluation of the Poisson ratio's [1,2]. This parameter has also been calculated indirectly from the equilibrium compression data [3], or using the biphasic indentation technique [4]. The transient response of cartilage during compression has been described by the biphasic model, which introduces the hydraulic permeability as the parameter regulating the velocity of fluid exudation from the tissue.

In order to evaluate the variation of the biomechanical properties with depth, we performed confined and unconfined creep experiments on disks of human articular cartilage representing consecutive layers of the full cartilage thickness; furthermore, the experiments were simulated by implementing a poroelastic FE model of articular cartilage.

MATERIALS AND METHODS

Specimen Preparation

Cartilage samples were harvested from the lateral tibial plateau (LTP), the lateral condyle (LC), the medial condyle (MC), or the femoral head (FH) obtained by surgical cut from 3 patients who underwent total knee or hip prosthesis. Only portions of the knee or hip that visually were not involved in the osteoarthritis process were used to prepare the samples. We obtained cylindrical plugs, 9 mm in diameter, which were cut into three slices, approximately of the same thickness, ~ 650 μm, starting from the superficial zone. The samples were stored at –26°C prior to testing. On the day of testing, each sample was thawed at room temperature and equilibrated in PBS containing protease inhibitors for about 30 minutes. Each sample first underwent the confined compression creep test, and then, after re-equilibration, the unconfined compression creep test. Finally we evaluated the porosity, f, from the water content, determined by the difference between the weights of the hydrated and dried cartilage sample. Cartilage samples were dried in an oven at 80 °C for 24h.

Confined Compression Creep Test And Protocol

The specimen was placed in a confining chamber (9 mm in diameter) over a stainless steel porous filter (average pore size ~ 40 μm) allowing free fluid flow (Figure 1a). Compressive loading (15 kPa) was applied on the cartilage specimen by a non-porous 100g stainless steel indenter. The indenter was connected to an LVDT (resolution 1μm) which measured the indenter vertical displacement for a total time of approximately 10,000 seconds. Using the equilibrium stress-strain data we calculated the aggregate modulus H_A; we identified the permeability k, fitting our experimental creep curve to the analytical solution of the confined compression creep problem [5].

Unconfined Compression Creep Test And Protocol

The specimen was placed between two plexiglass platens (Figure 1b) and loaded with a 100 g weight resulting in 15 kPa compression for a total time of approximately 3,000 seconds. The superior platen was connected to an LVDT (resolution 1μm). Using the equilibrium stress-strain data we calculated the Young's modulus E. Furthermore, the top surface of the cartilage disk was observed using a stereo-microscope connected to a PC by a digital camera, and digital images were acquired during the testing to allow optical evaluation of the radial deformation and hence the Poisson's ratio. This parameter was

also calculated by comparing the equilibrium response data (E, H_A), as suggested in [3].

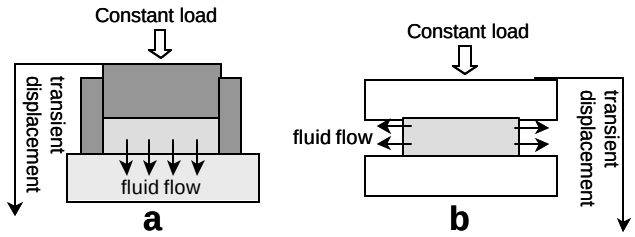

Figure 1. a) confined compression b) unconfined compression.

Poroelastic Model

A coupled pore fluid diffusion and stress analysis were performed using the commercial code ABAQUS (Hibbit Karlsson and Sorenses, Inc., Pawtucket, RI, USA) reproducing the creep-compression experimental tests both under confined and unconfined conditions. The poroelastic model describes the cartilage as a multiphase material constituted of an incompressible, isotropic, linear, elastic solid phase and an incompressible fluid phase. The model parameters are Young's modulus (E), Poisson's ratio (í), hydraulic permeability (k) and porosity (f) and were obtained as described above.

RESULTS

Results of the confined compression tests are shown in Table 1. Preliminary results of the unconfined compression tests for 2 samples of the femoral head representing a superficial and a deep layer show E values of 0.127 and 0.121 MPa respectively; indirectly evaluated ν values of 0.41 and 0.46 respectively; optically measured ν values of 0.39 and 0.45 respectively.

Table 1. Confined compression experiments results

Cartilage layer	thickness (μm)	H_A MPa	k $(m^4/N\ s) \times 10^{-15}$
LTP			
Deep	550 (n = 1)	0.598 (n = 1)	0.108 (n = 1)
Middle	687 (n = 3)	0.472 (n = 3)	0.266 (n = 3)
Superficial	700 (n = 2)	0.273 (n = 2)	0.788 (n = 2)
LC			
Deep	745 (n =1)	0.602 (n =1)	0.748 (n =1)
Middle	670 (n = 2)	0.476 (n = 2)	0.530 (n = 2)
Superficial	607 (n = 2)	0.258 (n = 2)	2.001 (n = 2)
MC			
Deep	850 (n = 1)	0.407 (n=1)	2.058 (n=1)
FH			
Deep	518 (n=2)	0.470 (n=2)	4.305 (n=2)
Middle	654 (n=5)	0.434 (n=5)	4.420 (n=5)
Superficial	632 (n=2)	0.364 (n=2)	4.634 (n=2)

Figure 2 shows one typical experimental confined compression displacement plot and the plot obtained by the poroelastic model with the parameters E, k, n, and f set to the measured ones for that particular tested sample.

CONCLUSIONS

Our results from the confined compression experiments show an increase in the aggregate modulus and, in most cases, a decrease in permeability moving from the superficial to the deep layer, reflecting the well known variability in tissue structure and composition with depth. The preliminary results derived from the unconfined compression experiments show a good agreement between the values of the Poisson's ratio calculated indirectly and those evaluated optically. The Poisson's ratio of the superficial layer is smaller than the one of the deep layer. An increase in the Poisson's ratio moving from the superficial to the deep layer has been both hypothesized [2] and measured [1,2].

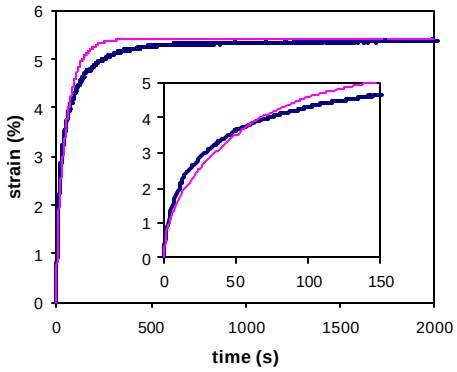

Figure 2. Comparison between an experimental confined compression displacement (thick line) and a simulated one.

The comparison between the simulated displacement plots and the experimental ones, shows that a constant value of k, as the one identified by confined compression experiments, is not able to precisely describe the transient response of articular cartilage during compression. Since permeability decreases with deformation [6], a constant value underestimates the true permeability for small deformations and conversely the permeability is overestimated for deformations approaching the equilibrium. Our experimental and computational method can be used in the future to identify the parameters that define the permeability dependence on deformation, as a function of depth from articular surface.

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