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Modeling of Articular Cartilage Growth Using COMSOL

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Abstract: The articular cartilage in the human knee joint functions in a highly demanding mechanical environment and its degeneration or defects have been, recently, treated with localized metal implants. The cartilage was found to grow onto the implant in the animal experiments. To simulate the cartilage growth, we have developed an axisymmetric human knee model with metal implant filling the assumed defect of the cartilage. To represent the biphasic behavior for cartilages and meniscus, we used poroelastic material model in COMSOL Multiphysics. The potential contacts among the respective interfaces were modeled with pore fluid continuity. As a preliminary attempt, the uniform growth was mimicked by thermal expansion. The results seem to be reasonable; however, more rigorous growth laws for the different constituents, which form the solid matrix of the cartilage, will be considered in future work.

Keywords: Articular cartilage, finite element modeling, poroelastic material, COMSOL models

1. Introduction

Articular cartilage is a multiphasic, fluid saturated, avascular connective soft tissue residing on the articulating ends of long bones in the diarthrodial joints and functions in a highly demanding mechanical environment. The degeneration or wear of the cartilage is a huge problem that affects millions of people every year. Localized resurfacing metal implants, replacing degenerated portions of the cartilage, are used to restore the healthy environment by reproducing the articulating surfaces in the joints, and have been showing good clinical outcomes (Custers et al., 2007, Manda et al., 2011). It has been shown that the local biomechanical factors directly affect the cartilage growth at the defect site (Duda et al., 2005, Klisch et al., 2008, Darling et al., 2003). From preliminary sheep experiments (Episurf Medical AB, Stockholm), the cartilage growth was observed around the implant. We believe that the healthy mechanical environment may have stimulated the growth of cartilage around the implant. In order to investigate this, there is need for developing an analytical articular cartilage growth model (ACGM).

The modeling of soft tissue structures, for many years, has been carried out using well established approaches; biphasic and poroelastic theories. In these approaches, tissue is modeled as a two phase immiscible mixture, consisting of incompressible solid and fluid phases. COMSOL Multiphysics, due to its expandable capabilities, have been gaining attention in recent years to model the complex biological processes in soft tissues. The long term objective of the present work is, using COMSOL, to develop an ACGM similar to the reported model from Ficklin et al. (2009) by including more realistic knee condyle geometry and dynamic loading situations. From this model, we want to investigate the effect of mechanical factors and role of metal implant, which is applied at the defect site to regenerate the articular surface, on the growth of cartilage around the implant.

2. Use of COMSOL Multiphysics

A simplified 2D axisymmetric representation of the human knee joint, including the articular cartilage layers, the meniscus and the underlying bones, was adopted from Wilson et al., (2003). The geometry was modified to include a defect sized (diameter 10 mm) metal implant and bones (Figure 1). The implant was placed slightly sunk (30% of cartilage thickness) into the cartilage with the surface of the implant perfectly matching the articulating surface. The models were solved with and without considering the defect sized implant.

2.1. Poroelastic model

The biphasic nature of the soft tissues can be represented by the poroelastic constitutive equations developed in the COMSOL software.

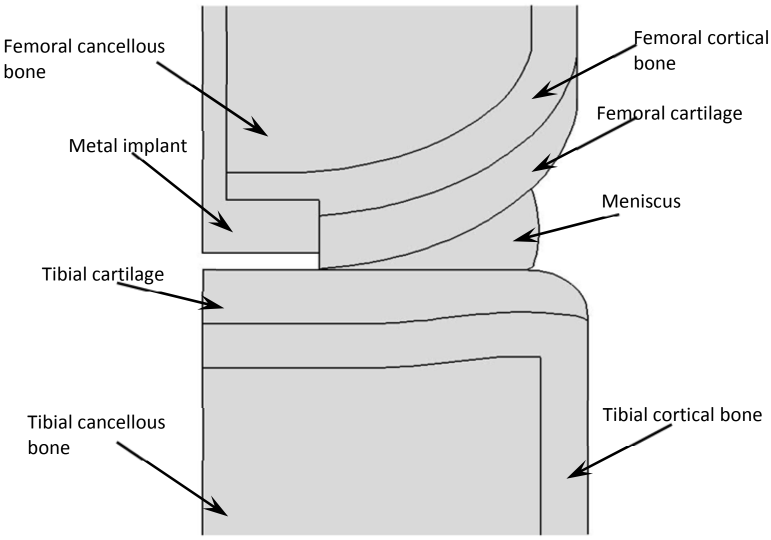

Figure 1. Simplified axisymmetric representation of the human knee condyle with the metal implant filling an assumed degenerated portion of the cartilage at center. The geometry was adopted from Wilson et al., (2003) and was modified to include metal implant, cancellous and cortical (2mm thick) parts of the bones. Material data was taken from Manda et al., (2011) and Wilson et al., (2003).

In the biphasic theory, articular cartilage is composed of a mixture of extracellular matrix as the porous solid phase (20 % of the total tissue mass by weight) and interstitial fluid as the fluid phase (80 %). Denoting the whole volume of mixture as V, the volume fraction of each phase is given by

$$\phi_i = \frac{dV_i}{dV}, i = s, f$$

with subscripts s and f are for solid and fluid phases, respectively. The saturation condition in the cartilage holds as

$$\phi_s + \phi_f = 1$$

Hence, the porosity of the model is equal to fluid volume fraction of the mixture ($\epsilon_p = \phi_f$).

The total stress acting at a point in the tissue is given by the sum of the solid and fluid stresses,

$$\sigma^{tot} = \sigma^{fluid} + \sigma^{solid}$$
$$\sigma^{tot} = -pI + C_{ijkl}\epsilon_{kl}$$

where σ^{solid} is the effective stress tensor due to elastic deformation of the solid phase and σ^{fluid} is the fluid stress, p is the hydrostatic fluid pressure from the fluid phase and I is the unit tensor, C_{ijkl} is the elasticity matrix of the drained solid phase and ϵ_{kl} is the strain matrix. The stress equilibrium equation is

$$\nabla \cdot \sigma^{tot} = 0$$

The continuity equation (law of conservation of mass) for the mixture is given by

$$\nabla \cdot \overrightarrow{v_s} - \nabla \cdot (\mathrm{k}\nabla\mathrm{p}) = 0$$

where, $\overrightarrow{v_s}$ is the velocity of the solid phase and k is the permeability, m^4/Ns

2.2. COMSOL modeling

We used the poroelastic model built into the COMSOL Multiphysics for the analysis. The cartilage was modeled as a biphasic fluid saturated porous medium and a uniform growth was simulated by a thermal expansion of the solid phase. The equivalence was established between the constituent growth laws and thermal expansion. The meniscus was also modeled as a saturated porous medium with transverse isotropy. Material properties for cartilage and meniscus were adopted from Wilson et al., (2003). The cancellous and cortical parts of the bones and the implant were assumed to be simple isotropic elastic materials, adopted from Manda et al., (2011).

The porosity, which is a constant of the COMSOL poroelastic model, has been modified to be strain dependent using the following relation

$$\epsilon_p = \frac{\epsilon_{p0} + e_{vol}}{1 + e_{vol}}$$

where ϵ_{p0} is an initial porosity and e_{vol} is volumetric strain of the solid phase.

The strain dependent permeability was also implemented using the variable porosity as

$$k = k_0 \left(\frac{1+e}{1+e_0}\right)^M , \qquad e = \frac{\phi_f}{\phi_s}$$

where k_0 is the initial permeability, e is void ratio, e_0 is initial void ratio, and M is the permeability coefficient, with values from Wilson et al., (2003).

The contacts between the respective contact surfaces (cartilages, femoral cartilage and meniscus, tibial cartilage and meniscus) have been modeled using the general structural mechanics contact models. The pore fluid at all the contact interfaces is assumed to be continuous, i.e., if the contact closes (src2dst in COMSOL), the pore fluid pressure on the destination surface should be equal to the fluid pressure on the corresponding source surface.
The tibial bottom plane was constrained in all directions and the dynamic loading, which a knee is subjected to during everyday life activities, was represented with an axial ramp compressive displacement (0.1mm) of 1 sec and then constant, applied to the model on the top plane of the femur. The displacement variables were discretized with quadratic triangular elements and fluid pressure, contact pressure variables were discretized with linear triangular elements. Adaptive mesh refinement in COMSOL was used with the time-dependent segregated solver, which refines the mesh whenever the mesh quality drops below a predefined value.

3. Results

The total pressure of the model without defect was showed in Figure 2. The contact pressure and total values increased with the defect filled with metal implant. The results obtained in the current work show that the cartilage was seen growing onto the implant with time (Figure 3). The mechanical loading also made the cartilage edges around the implant to radially move onto the implant, in addition to the biological growth.

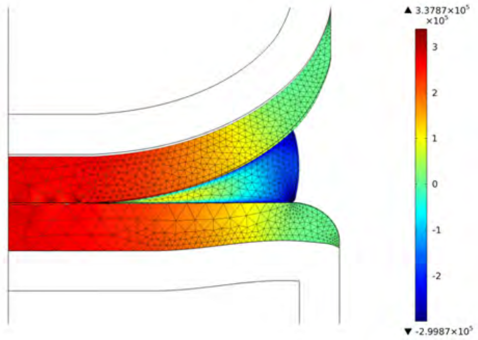

Figure 2. Total pressure (Pa) in the soft tissues, which is sum of fluid and solid pressures (scale=1) in a model without implant.

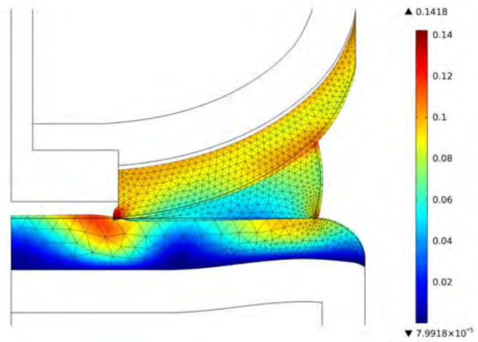

Figure 3. Displacements (mm) in the cartilages and meniscus (scale=2).

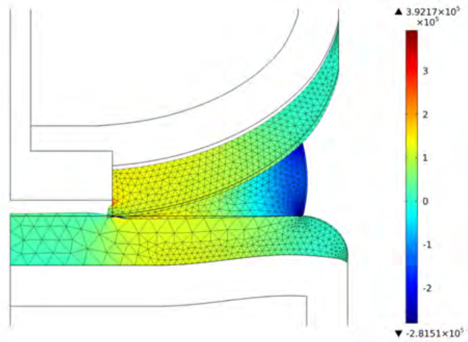

Figure 4. Total pressure (Pa) in the soft tissues, which is sum of fluid and solid pressures (scale=2).

A conclusion was that the deeper the implant was placed, the higher the cartilage grew onto the implant, and into the gap between the cartilages.

4. Discussion and Conclusions

The study mainly focused on developing a conceptual analytical model for cartilage growth. The various axial positionings of implant and dynamic loading cases were tested. The growth was, initially, implemented with a uniform thermal expansion. We faced many convergence difficulties when solving contact problem between the poroelastic surfaces, and while implementing pressure continuity at the contact interfaces. The models were working for the small displacement boundary conditions, but the solver was not converging for larger applied displacements/forces. We hope to solve these problems in near future.

Later, the actual growth of the individual constituents and compatibility among them will be implemented with the respective growth laws (Davol et al., 2008) in COMSOL. Results will demonstrate how the implant's position in the lining tissue and biomechanical factors affect the growth and behavior of cartilage surrounding the cartilage.

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