

A FINITE ELEMENT MODEL OF TRANSPLANTED TISSUE-ENGINEERED CARTILAGE

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INTRODUCTION

Focal chondral defects are a very common and widespread disorder that result from traumatic injuries or imbalanced joint loading. Since these defects can induce osteoarthritic degenerative changes, their early detection and repair could prevent the onset of osteoarthritis and its associated burdens [1]. One promising repair strategy is the use of chondral and osteochondral tissue-engineered (TE) constructs which are composed of scaffolds seeded with cells. Determination of the stress and strain distributions that TE constructs experience when transplanted in human joints is of critical importance to their function and efficacy. Here, results from a finite element (FE) model of native articular cartilage (AC) are compared to a model of a transplanted TE construct surrounded by AC.

METHODS

In order to model fluid and solid phase interactions and also fixed charge densities (FCD) present in cartilage and TE construct, a biphasic swelling [2,3] model using FE package ABAQUS® (version 6.6) was developed. Two 2D axisymmetric models representing AC and AC with a transplanted TE construct were created. Models were 12mm in radius and 1mm in thickness and consisted of 1350 4-node bilinear displacement- pore pressure elements. In order to model transplanted TE construct, a 3mm region with 20% less FCD and 50% lower Young's modulus (Fig. 1) compared to native AC was created in the center [4]. In both models, the bottom surface was made impermeable, and zero pore pressure was prescribed on all other boundaries so that fluid could freely flow in and out. The displacements of the nodes on the bottom plane were confined in all direction representing cartilage attachment to the subchondral bone. The solid collagen-PG matrix was modeled as a Neo-Hookean material. Osmotic swelling pressure and chemical expansion stress (arising from repulsive forces between large negative groups on PGs) were included in the model in a UMAT subroutine. AC material properties comparable to those found in the literature and strain-dependent permeability were assigned to the model. For native AC, initial FCD and Young's modulus were set to 2×10^{-4} mmol/mm³ and 0.5 MPa respectively. Initial porosity was set to 75% for both models. At the start of the simulation, both models were in equilibrium with a physiological salt solution of 1.5×10^{-4} mmol/mm³. A rigid impermeable spherical indenter [5] with a radius of 20mm applied indentation to the AC. Short time creep response of the sample was investigated by applying a 1N load in a ramp fashion to the indenter over 0.5 sec and was then held constant for 30 sec.

RESULTS

At the end of simulation, we compared mechanical field variables of the model of uniform, native cartilage to the model that included the implanted TE construct.

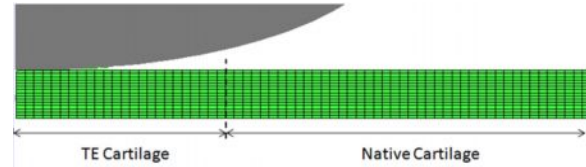


Fig.1. Simplified model of transplanted TE construct

In the TE model, maximum fluid velocity decreased by 42% and maximum pore pressure decreased by 17% (Fig. 2). Maximum principal strain increased 27% in the TE model (data not shown).

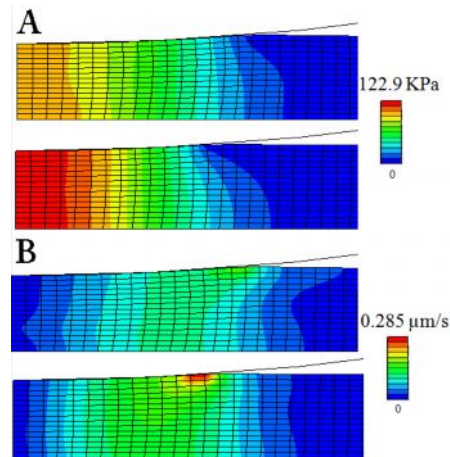


Fig.2. (A) Comparison of pore pressure and (B) fluid velocity distribution for TE model (top) and uniform AC model (bottom)

DISCUSSION

Using FE-based computer simulations, mechanical parameters in a TE cartilage construct were calculated. The altered fluid velocity and pore pressure in transplanted TE cartilage may have implications for the performance and biological response of the cell-seeded constructs. Future models will include a more detailed representation of AC structure and properties such as anisotropies and inhomogeneities, collagen fibers, chondrocytes and more complex geometries for transplant.

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