Thomas Reuter*, Christof Hurschler
Comparison of biphasic material properties of equine articular cartilage from stress relaxation indentation tests with and without tension-compression nonlinearity

Abstract: The mechanical parameters of articular cartilage estimated from indentation tests depend on the constitutive model adopted to analyze the data. In this study, we present a 3D-FE-based method to determine the biomechanical properties of equine articular cartilage from stress relaxation indentation tests ($\varepsilon = 6\%$, $t = 1000$ s) whereby articular cartilage is modeled as a biphasic material without (BM) and with tension-compression nonlinearity (BMTCN). The FE-model computation was optimized by exploiting the axial symmetry and mesh resolution. Parameter identification was executed with the Levenberg-Marquardt-algorithm. The $R^2$ of the fit results varies between 0.695 and 0.930 for the BM-model and between 0.877 and 0.958 for the BMTCN-model. The differences of the $R^2$ occur from the more exact description of the initial stress relaxation behaviour by the fiber modulus from the BMTCN-model. The fiber modulus defines the collagen matrix of cartilage. Furthermore, for both models the determined values of Young’s modulus and permeability were in the same order of magnitude.

Keywords: Articular Cartilage, Stress Relaxation Indentation, FE-modelling, Parameter Identification, Biphasic Theory, Tension-Compression-Nonlinearity.

https://doi.org/10.1515/cdbme-2018-0116

1 Introduction

Articular cartilage can be investigated and characterized by biomechanical, biochemical and histological analysis. The mechanical parameters of articular cartilage estimated from experiments depend on the constitutive model adopted to analyze the data [1]. In this study, we modelled articular cartilage first as classic biphasic material (BM) and second as an extended BM-model with tension-compression nonlinearity (TCN) [2, 3]. The TCN describes the differing tension and compression properties of the solid matrix [4]. Additionally, an optimized biphasic 3D-FE-model-based approach was implemented to extract the BM-model parameters (Young’s modulus $E$, Poisson’s ratio $\nu$ and permeability $k$) and the BMTCN-model parameters (Young’s modulus $E$, fiber modulus $\xi$ and permeability $k$) directly by fitting the model to the measured stress relaxation indentation data [2-4].

Furthermore, a comparison between BM-model and BMTCN-model was performed.

2 Methods

In this section the stress relaxation indentation measurements to record the resulting indentation profile over time at a constant strain value and the needle indentations to measure
the cartilage thickness are explained. Also, the optimized FE-based parameter extraction model to identifying the material properties of articular cartilage is developed. In this case, articular cartilage is described as a biphasic material without and with TCN.

2.1 Experimental Setup

Equine medial anterior condyles were harvested from knee joints to carry out the stress relaxation indentation experiments (cadaver specimens from abattoir, n=5, age 5-6 years, no visible damage, stored fresh frozen at -20°C).

The indentation was performed by a flat-end cylindrical indenter with a tip diameter of 1 mm. The indenter was placed perpendicular to the tissue surface. A custom-built stress relaxation-indentation testing machine (fzmb GmbH, Bad Langensalza) was used to apply a strain of \( \varepsilon = 6\% \) with \( v = 0.02 \text{ mm/s} \) (Fig. 1). The samples were mounted in a culture dish filled with physiological saline solution. The resulting indentation force was measured with a force sensor (ME-Systeme©, Henningsdorf). The sensor was placed under the culture dish.

The following load profile was used: 1) Determination of the cartilage’s surface (move indenter until a load of 50 mN is reached); 2) Recovery from the preload of step 1 for 10 min; 3) Relaxation indentation at \( \varepsilon = 6\% \) for 1000 s [5].

The needle probe thickness measurement was performed first, followed by stress relaxation indentation at the location 1 mm adjacent to the location of the needle measurement (Fig. 1) [6]. All tests were performed at 37°C in 0.15 M PBS solution.

2.2 Computational Modelling

For the calculation of the biomechanical parameters of articular cartilage from stress relaxation indentation, an FE-based parameter extraction model was applied. The software Finite Elements for Biomechanics (FEBio©, Version 2.3) was used to set up the FE-model and the parameter identification [7]. The model was configured in accordance with the experimental setup to match the individual dimensions and boundary conditions of each specimen. Due to the long processing time, a pseudo-axisymmetric wedge model was implemented (Fig. 2). Additionally, the FE-mesh resolution was adjusted to minimize the model error and to get mesh independent model results [8].

For the parameter identification, the parameter optimization module of FEBio© was used. This module iteratively identifies material parameters by solving an inverse finite element problem by coupling the FE-model to an optimization scheme that minimizes the error between the experimental results and the FE-predicted time-dependent stress relaxation curve. The Levenberg-Marquardt algorithm was used as optimization scheme [7, 11]. The optimization objective function that had to be minimized was defined as the sum of squares of the difference between experimental and simulated stress relaxation curves. The convergence tolerance and the scale factor for the objective function was set to 0.001. The coefficient of determination \( R^2 \) of the optimization was calculated for every fitting result for the evaluation of the parameter identification.

The FE-modelling and optimization process was conducted on a 64-Bit Quad-Core Workstation z800 HP®.

2.2.1 Biphasic material

First, the BM-model was used for the description of the mechanical behaviour of articular cartilage. According to the biphasic theory [9], the tissue was assumed to consist of an incompressible solid matrix, hydrated with an incompressible fluid. The total stress in the tissue was given by the sum of the solid and fluid stress (1)

\[
\sigma_{\text{tot}} = \sigma_s - pl
\]  

(1)

where \( \sigma_s \) is the effective stress tensor, \( p \) the hydrostatic fluid pressure, and \( I \) the unity tensor. The solid matrix is assumed to be hyperelastic and isotropic. The material representation used consisted of a solid matrix with hyperelastic compressible neo-Hookean properties, constant permeability and an inviscid fluid phase, which introduces three material parameters of Young’s modulus, Poisson’s ratio and permeability of the solid matrix [10].

The hyperelastic strain energy density was derived from the following function (2)
\[ \psi = \frac{\xi}{\beta} \left( I_1 - 3 \right) - \mu \ln J + \frac{\alpha}{\beta} \left( \ln J \right)^2 \]  

(2)

where \( \alpha \) and \( \mu \) are the Lame’ constants, \( I_1 \) is the first invariant of the left Cauchy green tensor and \( J \) is the determinant of the deformation gradient tensor [7].

### 2.2.2 Biphasic material with TCN

Second, the solid matrix of the BM-model was extended with tension-compression nonlinearity (TCN). The solid matrix is treated as a compressible isotropic neo-Hookean ground matrix reinforced with spherical fiber distribution. The fibers can only sustain tensile stress. Therefore, the compressive modulus of the material is defined to be the Young’s modulus of the neo-Hookean ground matrix. The strain energy density function of the fiber bundles is defined as

\[ \psi = \frac{\xi}{\beta} \left( e^{\alpha \left( I_1 - 1 \right)} - 1 \right) \]  

(3)

where \( \zeta \) is the fiber modulus, \( \alpha \) and \( \beta \) are constants and set to 0 and 2 to perform an almost linear stress-strain relationship at small strains [4, 12]. Poisson’s ratio of the neo-Hookean ground matrix is assumed to be 0 [13].

### 3 Results

Experimental stress relaxation curves of the five equine articular cartilage specimens represent typical viscoelastic relaxation behaviour of cartilage tissue (Fig. 3). No meaningful changes occur in the measured force after about \( t = 250 \) s.

Typically, the differences between simulation and experiment are model dependent (e.g. sample no. 4, Fig. 4). The calculated \( R^2 \) for the BM-model is 0.863 and 0.951 for the BMTCN-model. The BMTCN-model describes the stress relaxation indentation curves with higher accuracy than the BM-model. Particularly, the initial stress relaxation behaviour is better described by the fiber modulus from the BMTCN-model. However, the Young’s modulus of both models describes the equilibrium force of cartilage [4].

![Figure 4: Comparison between simulation and experiment.](image)

The \( R^2 \) varies between 0.695 and 0.930 for all samples for the BM-model. The estimated material parameters vary from 0.566 to 0.853 MPa for Young’s modulus \( E \) and 0.004 to 0.019 mm\(^4\)/Ns for permeability \( k \) (Tab. 1).

### Table 1: Calculated material parameters for BM-model.

<table>
<thead>
<tr>
<th>Sample no.</th>
<th>Young’s modulus ( E ) [MPa]</th>
<th>Poisson’s ratio ( \nu )</th>
<th>Permeability ( k ) [mm(^4)/Ns]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.853</td>
<td>0.00</td>
<td>0.004</td>
</tr>
<tr>
<td>2</td>
<td>0.689</td>
<td>0.00</td>
<td>0.006</td>
</tr>
<tr>
<td>3</td>
<td>0.603</td>
<td>0.00</td>
<td>0.006</td>
</tr>
<tr>
<td>4</td>
<td>0.566</td>
<td>0.00</td>
<td>0.019</td>
</tr>
<tr>
<td>5</td>
<td>0.720</td>
<td>0.00</td>
<td>0.007</td>
</tr>
<tr>
<td>Mean ± Std</td>
<td>0.686 ± 0.100</td>
<td>0.00 ± 0.00</td>
<td>0.008 ± 0.005</td>
</tr>
</tbody>
</table>

The determined values for the Poisson’s ratio for all samples are very close to 0.00 [13].

For the BMTCN-model, the \( R^2 \) varies between 0.877 and 0.958 for all samples. The estimated material parameters vary from 0.529 to 0.802 MPa for Young’s modulus \( E \), 0.007 to 0.111 MPa for fiber modulus \( \zeta \) and 0.004 to 0.014 mm\(^4\)/Ns for permeability \( k \) (Tab. 2).

![Figure 3: Experimental data from stress relaxation indentation.](image)
Table 2: Calculated material parameters for BMT-CN-model.

<table>
<thead>
<tr>
<th>Sample no.</th>
<th>Young’s modulus E [MPa]</th>
<th>Fiber modulus ξ [MPa]</th>
<th>Permeability k [mm²/Ns]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.802</td>
<td>0.089</td>
<td>0.004</td>
</tr>
<tr>
<td>2</td>
<td>0.679</td>
<td>0.007</td>
<td>0.006</td>
</tr>
<tr>
<td>3</td>
<td>0.539</td>
<td>0.111</td>
<td>0.004</td>
</tr>
<tr>
<td>4</td>
<td>0.529</td>
<td>0.060</td>
<td>0.014</td>
</tr>
<tr>
<td>5</td>
<td>0.683</td>
<td>0.053</td>
<td>0.006</td>
</tr>
<tr>
<td>Mean ± Std</td>
<td>0.646 ± 0.102</td>
<td>0.063 ± 0.035</td>
<td>0.007 ± 0.004</td>
</tr>
</tbody>
</table>

The determined values of Young’s modulus and permeability were in the same order of magnitude (the differences are not significant) for both models. The fiber modulus determined from the BMT-CN-model is 10-times lower than the Young’s modulus.

4 Conclusion

An optimized biphasic 3D-FE-based method was utilized to characterize the mechanical behaviour of articular cartilage from stress relaxation indentation data. Articular cartilage was modeled as a biphasic material without and with tension-compression nonlinearity. For the computational determination of the material parameters of the BM-model and BMT-CN-model, the optimization routine operated robustly and converged to unique parameter values within one hour. The BMT-CN model described the experimental stress relaxation indentation with higher accuracy ($R^2 = 0.92 ± 0.04$) than the BM-model ($R^2 = 0.84 ± 0.08$). Particularly, the initial stress relaxation behaviour is largely described by the fiber modulus from the BMT-CN-model. However, the Young’s modulus of both models described the equilibrium force of cartilage [4].

In this study, the determined parameters of both models were in the same order of magnitude, permeability values were particularly close. The mean value for the Poisson’s ratio estimated from BM-model is nearly zero. The Poisson’s ratio is rather insensitive and assumed to be zero for the BMT-CN-model [13]. The fiber modulus determined from the BMT-CN-model is 10-times lower than the Young’s modulus. Future works will focus on the characterization of cartilage constructs with the BMT-CN-model. Especially the maturing process of cartilage constructs will analyse with stress relaxation tests.

Author Statement

Research funding: The author state no funding involved.
Conflict of interest: Authors state no conflict of interest.
Informed consent: Informed consent is not applicable.
Ethical approval: The conducted research is not related to human or animals use.

References