Biomechanical modeling of the human jaw joint
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Citation for published version (APA):

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Abstract- The hypothesis was tested that the dynamical behavior of the human temporomandibular joint disc can be adequately described by a poroelastic material model. This hypothesis was tested by comparing the results from model predictions with results obtained from cyclic indentation experiments that were performed with fresh discs. The relationship between the applied indentation and the resulting reaction force from both simulations and experiments showed remarkable similarities when the solid matrix was assumed to be hyperelastic. The maximum stress and the amount of energy dissipated in each subsequent cycle decreased both in the experiments and the simulations. Furthermore, a similar dependency on the indentation frequency and amplitude was found. It could be concluded that the poroelastic material model can be applied to describe the dynamical behavior of the temporomandibular joint disc adequately.
Introduction

The human temporomandibular joint contains a cartilaginous intra-articular disc. The influence of this disc on the mechanics of the joint has not been fully elucidated yet. Efforts have been made to investigate the mechanical behavior of the disc in the joint by means of finite element modeling (Chen and Xu, 1994; DeVocht et al., 1996; Chen et al., 1998; Nagahara et al., 1999; chapters 4 and 5). The results of these simulations show that the disc plays an important role in distributing and absorbing loads acting on the joint. The applied models were (quasi-) statical and the area of relevance of their results, therefore, is limited because the habitual functioning of the temporomandibular joint is highly dynamical (e.g., talking and chewing).

The mechanical behavior of the disc has also been investigated by means of experimental studies (e.g., Tanne et al., 1991; Teng et al., 1991; Chin et al., 1996; Lai et al., 1998; Tanaka et al., 1999). The results of these studies indicate that the mechanical behavior of the disc is non-linear, anisotropic and time-dependent. These studies also were of statical or quasi-statical nature. However, the behavior of cartilaginous tissues during statical loading differs from the behavior during dynamical loading (Eckstein et al., 2000). Recently, sinusoidal indentation experiments were performed to investigate the behavior of the disc under more physiologic loading conditions (Beek et al., 2001a: see chapter 6). It was found that the maximum reaction force and the amount of energy dissipated within the disc decreased in time (Fig. 1). Furthermore, these characteristics appeared to be dependent on location, strain amplitude, and excitation frequency. This specific behavior can probably be attributed to a mechanical interaction between the solid matrix (mainly collagen and proteoglycans) and the interstitial fluid (mainly water) of the cartilage (Mow and Wang, 1999).

In the last decades, various material models have been applied to describe the mechanical behavior of the human temporomandibular joint disc. A linear elastic material model is the most simple and can be applied in statical analyses (e.g., Beek et al., 2000a, chapter 4). Chen et al. (1998) applied a Mooney-Rivlin material model, developed to represent rubber-like material behavior. Such a Mooney-Rivlin model is
TMJ disc cartilage as poroelastic material

Figure 1.

A: Experimental set-up of the dynamical indentation experiments with human temporomandibular joint discs performed by Beek et al. (2001a). During the experiments, an intact disc was submerged in saline which was kept at a temperature of about 37°C.

B: Typical time series of the measurements. On top, the sinusoidal displacement of the top indenter. Below, the reaction force of the disc.

C: Typical stress-strain curve, calculated from the measurements obtained from 10 subsequent loading cycles.

capable of handling relatively large deformations often occurring in biological tissues. However, these two material models can not describe changes of the structure's mechanical behavior in time. Simple linear time-dependent behavior can be described using a viscoelastic material model (e.g., Fung, 1981). In order to assess the interaction of the solid matrix and the interstitial fluid, a biphasic theory has been developed (Mow et al., 1980), which showed that a large part of the loads acting on cartilaginous structures is carried by interstitial fluid pressurization (Soltz and Ateshian, 1998). Surprisingly, cartilage behaves in many aspects as soil for which a poroelastic model has been developed (e.g., Simon, 1992). It has been demonstrated that this material model is capable of adequately modeling the behavior of cartilaginous structures similar to the biphasic models (Prendergast et al., 1996; Wu et al., 1998). In contrast to the biphasic theory, the poroelastic theory is
implemented in most commercial finite element software. Consequently, this material model seems to be advantageous for application in dynamical finite element models involving cartilage in general and the temporomandibular joint disc in particular.

The purpose of the present study was to test the hypothesis that the poroelastic material model is applicable to describe the tissue behavior of the cartilaginous disc of the human temporomandibular joint disc. Therefore, a finite element model was developed and subjected to cyclic dynamical indentations similar to the ones with regard to human temporomandibular joint discs as described in chapter 6 of this thesis. The focus of the simulations was directed to the ability to adequately model the specific qualitative characteristics found in the experiments, namely the decrease in time of both the maximum force and amount of dissipation, and the dependency on both the strain amplitude and the excitation frequency.

**Materials and methods**

**Model**

An axisymmetric finite element model of cartilaginous tissue (radius $r = 3.49$ mm) was created to simulate dynamical indentation experiments performed on fresh human temporomandibular joint discs (Fig. 1 and chapter 6). This model was loaded by undeformable indenters ($r = 1.97$ mm). The disc model was larger than the indenters, because the experiments had been performed with intact discs. The mesh consisted of 160 isoparametric quadrilateral elements (Fig. 2). The cartilaginous tissue was supposed to consist of a solid matrix and interstitial fluid, and was modeled using the poroelastic theory (Simon, 1992). Prendergast et al. (1996) compared results obtained with the poroelastic material model of several commercial finite element codes with an analytic solution of the biphasic material model. Initially, the values Prendergast et al. (1996) applied for the material parameters for the poroelastic material were also chosen in this study, i.e., a Young's modulus and a Poisson's ratio of the isotropic solid matrix of 0.4667 MPa and 0.1667, respectively; the biphasic permeability of the solid matrix was $7.5 \times 10^{-15}$ m$^4$/Ns. Additionally, the porosity of cartilage was assumed to be 0.7, according to a 70% fluid content of
cartilaginous structures (Mow et al., 1984; Fithian et al., 1990). The pore pressure at the boundary of the disc outside the indenters was assumed to be zero. This allowed for a free fluid flow across these boundaries in an aqueous environment. At the symmetry axis the radial displacement of the solid matrix and the radial velocity of the fluid were assumed to be zero. The indenters were modeled to be rigid and impermeable. It was assumed that the disc was directly connected to both indenters. The simulations were performed using MARC 7.3.2 (MSC.Software, Los Angeles, USA) on the Cray C916 computer at SARA, Amsterdam.

**Simulations**

The experiments were simulated by applying a sinusoidal displacement ($\Delta h$) in axial direction to the nodes in contact with the top indenter (Fig. 2). The strain value $\varepsilon$ as measured in the experiments (chapter 6) was calculated as the quotient of the axial displacement of the node on the symmetry axis and the unloaded thickness ($h_0$) of the tissue between the indenters ($\varepsilon=\Delta h/h_0$). The displacement of the nodes in contact with the bottom indenter was restrained. The reaction forces gathered at the node on the symmetry axis by means of kinematic tyings, enabled to obtain the total reaction force $F$ instantaneously. The applied stress ($\sigma=F/A$) was defined as the
reaction force divided by the area \((A=\pi r^2)\) of the indenters. After ten sinusoidal indentation cycles, relaxation was allowed for 200 s to verify the model's return to its original state. The ten indentation cycles were simulated in 1500 increments and the following 200 s of relaxation in 500 increments.

In contrast to the results of Prendergast et al. (1996), the results of our previous experiments (chapter 6) indicated that the behavior of the solid matrix was hyperelastically dependent on the strain. Therefore, a strain dependent Young's modulus \(E_{tg}\) was used to model this hyperelasticity of the solid matrix:

\[
E_{tg} = c_3\varepsilon_{eq}^3 + c_2\varepsilon_{eq}^2 + c_1\varepsilon_{eq} + c_0 \tag{1}
\]

in which:

\[
\varepsilon_{eq} = \sqrt{\varepsilon_{11}^2 + \varepsilon_{22}^2 + \varepsilon_{33}^2} \tag{2}
\]

This hyperelasticity was implemented in the poroelastic material model through a user subroutine. The applied values for \(c_3, c_2, c_1, \) and \(c_0\) were \(10^3, 10^3, 0,\) and 1, respectively. At present, there is no agreement on the Young's modulus of the temporomandibular joint disc (Tanne et al., 1991; Teng et al., 1991; Chin et al., 1996; Tanaka et al., 1999). Furthermore, reliable quantitative data concerning the permeability of cartilage and in particular of the temporomandibular joint disc is scarce (Mansour and Mow, 1976; Fithian et al., 1990; Prendergast et al., 1996). This parameter determines the ease of fluid flow through the solid matrix. Therefore, we performed sensitivity analyses involving the Young's modulus and the biphasic permeability to evaluate the influence of these parameters on the behavior of the poroelastic model.

The accuracy of the poroelastic material model was further investigated by determining the influence of the indentation amplitude and the indentation frequency, respectively. For comparison with experimental data (chapter 6), we applied 20%, 30% and 40% for the indentation amplitude at a frequency of 0.05 Hz, and 0.02 Hz, 0.05 Hz and 0.1 Hz for the indentation frequency at an amplitude of 30%, respectively.
Results

Using a linear poroelastic model (Prendergast et al., 1996), simulation of the cyclic indentation experiments (10 cycles) resulted in an almost linear viscous behavior (Fig. 3A). The only difference between poroelasticity and viscosity present in this simulation, was a shift of the stress-strain curve in the positive stress direction. The reaction force at 30% indentation was 4 N during the first cycle. The strains in the solid matrix were inhomogeneously distributed in axial direction. Simulating a relaxation of 200 s showed that all mechanical quantities returned to their original state. Introducing a strain dependency of the elasticity of the solid matrix according to Eq. (1) and (2), resulted in a stress-strain curve having about the same superproportional shape as found experimentally (Fig. 3B), although the decrease of the area enclosed by this curve after the second loading cycle was less.

\[
E_{eq} = c_3\varepsilon_{eq}^3 + c_2\varepsilon_{eq}^2 + c_1\varepsilon_{eq} + c_0,
\]

with \(c_3=10^3\), \(c_2=10^3\), \(c_1=0\), and \(c_0=1\).

Figure 3.
Simulation of cyclic dynamic indentation experiments. indentation frequency and amplitude were 0.05 Hz and 30%, respectively.
A: Using a linear poroelastic model according to Prendergast et al. (1996). B: Using a poroelastic model with a nonlinear solid matrix, according to \(E_{eq} = c_3\varepsilon_{eq}^3 + c_2\varepsilon_{eq}^2 + c_1\varepsilon_{eq} + c_0\), with \(c_3=10^3\), \(c_2=10^3\), \(c_1=0\), and \(c_0=1\).

Sensitivity analyses

The results of the sensitivity analyses are shown in Fig. 4. An increase of the elasticity of the solid matrix by a factor ten made the cartilage relatively stiff (Fig. 4A). The maximum force at equilibrium was ten times larger compared to the value in the reference simulation. This increase also had an influence on the ease of fluid flow through the collagen network. Although the area between the loading and unloading stress-strain curves was larger than during the reference simulation, the increase
was much less than the increase of the maximum force. A decrease in elasticity of the solid matrix had a reciprocal effect.

A ten-fold increase of the permeability resulted in a decrease of the maximum force (Fig. 4B). In the subsequent cycles there was hardly any decrease in resistance. The area between the loading and unloading stress-strain curves was decreased much more than the relatively small decrease of the maximum force. A decrease of the value of the permeability led to an increase in the maximum force in the first compression cycle. Also the amount of energy dissipation was larger. In each subsequent cycle the maximum force decreased asymptotically towards its value at equilibrium. While the maximum force at equilibrium was hardly influenced by the value of the permeability this decrease was relatively large.

**Figure 4.**

A: Influence of the Young's modulus $E$ on the stress-strain curves obtained from simulations of ten subsequent indentation cycles. The value of $E$ was varied with respect to $E_{ref}$=0.4667 MPa. Indentation frequency and amplitude were 0.05 Hz and 30%, respectively.

B: Influence of the permeability $k$ on the stress-strain curves obtained from simulations of ten subsequent indentation cycles. The value of $k$ was varied with respect to $k_{ref}$=7.5x10^{-15} m^{4}/Ns. Indentation frequency and amplitude were 0.05 Hz and 30%, respectively. For visibility reasons, only the first cycle is displayed.

**Dependency on indentation amplitude and frequency**

Figure 5 shows the results of the simulations in which the amplitude of the indentation with a frequency of 0.05 Hz was varied. Figure 5A shows that the reaction force is superproportionally dependent on the indentation amplitude. Figure 5B shows that the shape of the stress-strain curve is hardly influenced by the amplitude of the indentation. Unfortunately, the simulations with an indentation amplitude of 40% did not converge after six cycles. Therefore, only the first six cycles are depicted.
Figure 5.
Influence of the indentation amplitude. The values applied for the indentation amplitude were 20%, 30%, and 40%, respectively. The indentation frequency was 0.05 Hz.
A: Time series of the reaction force. B: Stress-strain curves.

Figure 6 shows the results of the simulations in which the frequency of the indentation with an amplitude of 30% was varied. Figure 6A shows that the maximum reaction force in the first indentation cycle is proportionally dependent on the frequency. However, the reaction force after ten cycles (approximation of equilibrium) is larger for an indentation frequency of 0.05 Hz than for a frequency of...

Figure 6.
Influence of the indentation frequency. The values applied for the indentation frequency were 0.02 Hz, 0.05 Hz, and 0.1 Hz, respectively. The indentation amplitude was 30%.
A: Time series of the reaction force. B: Stress-strain curves.
0.1 Hz. When the disc is indented at a higher frequency, the disc behaves stiffer during the loading phase of the indentation (Fig. 6B). The unloading phase is hardly affected by the frequency. Fig. 7 shows the maximum stress obtained at all combinations of indentation amplitude and frequency. This parameter appeared to be more dependent on the amplitude than on the frequency.

**Discussion**

In the present study an axisymmetric finite element model was used to simulate dynamical indentation experiments with human temporomandibular joint discs. The results of the previous experiments showed that the mechanical behavior of the disc was nonlinear and time-dependent (chapter 6). The fluid content in cartilaginous structures has been shown to comprise about 70% to 85% of the total mass (Mow et al., 1984; Fithian et al., 1990). The remaining part mainly consists of a collagen network and proteoglycans. Several studies have shown that this large amount of fluid plays an important role in the complex mechanical behavior of cartilage (Soltz and Ateshian, 1998; Bursac et al., 1999). Therefore, it is crucial to apply a material model that includes both fluid and solid constituents, enabling to distinguish between the mechanical functions of these different constituents. In the present study, the poroelastic theory was applied to investigate its capability to describe specific characteristics of the behavior of the temporomandibular joint disc as found in experiments.

The results of the simulations show various similarities with the results obtained from the dynamical indentation experiments. Both the reaction force and the amount of energy dissipation were decreased in subsequent cycles. In the simulations the maximum stress appeared to be superproportionally dependent on the strain amplitude and only marginally on the frequency (Fig. 7). The same characteristics have also been reported in the experimental study. Consequently, the results indicate that the poroelastic material model adequately describes the strain and frequency dependency of the temporomandibular joint discs as found experimentally.

In the indentation experiments described in chapter 6, intact temporomandibular joint discs were glued between the indenters. The applied model was also larger
Strain and frequency dependency of the maximum stress in the first cycle, for all combinations of indentation amplitude and frequency.

than the indenters. Apart from between the indenters and their near surroundings, the deformations in the tissue were negligible, ensuring that the influence of the limited measures of the model were negligible. Free fluid flow across the free boundaries of the disc was allowed. This boundary condition is in agreement with various studies which have shown that fluid flow across the boundaries does indeed occur, for example in confined compression experiments (review: Cohen et al., 1998).

The stress-strain curves from the indentation experiments (chapter 6) were determined for the whole tissue, i.e., the determined stresses did not only involve the stresses in the solid matrix but also the pressurization of the interstitial fluid. Because the pressurization of the fluid was not measured in the experiments, the exact stress-strain state of the solid matrix remained unknown. Therefore, the hyperelastic behavior of the solid matrix was modeled arbitrarily by a third-order polynomial. The applied constants can be adapted in order to obtain a shape similar to a specific experiment.

In the present study, the tissue of the disc was supposed to be homogeneous. However, various studies have shown that the internal organization of cartilaginous structures (e.g., direction and density of collagen fibers) varies with location (Arokoski, 1999) and, in particular, such regional variations in the internal structure
have also been reported to exist in the temporomandibular joint disc (Kopp, 1976; Piette, 1993; Minarelli et al., 1997). Furthermore, different proteoglycans have been reported with different functions in cartilaginous structures like the temporomandibular joint disc (Tanaka et al., 2000). These internal variations in the internal organization of the collagen fibers might be an explanation for observed regional variations in the mechanical behavior of cartilage (Gore et al., 1983; Athanasiou et al., 1995; Schinazi et al., 1997; Jurvelin et al., 2000) and the temporomandibular joint disc (Tanne et al., 1991; Teng et al., 1991; Lai et al., 1998; chapter 6). The sensitivity analyses described in this study show that such heterogeneous behavior can be modeled by assigning individual material parameters to relevant regions.

In the present simulations somewhat larger negative forces were predicted in the unloading phase of the indentation than in the experiments. This can be attributed to a better connection between disc and indenters in the simulations than in the experiments. Because fluid is expelled out of the cartilage, the volume of the tissue between the indenters decreases and tensile forces are needed to restore the original thickness (h0) of the tissue between the indenters. When the contact between the disc and the indenters is not perfect, which might be the case in the experiments, less tensile forces can be transmitted to the cartilage. Then, the original thickness would not be fully restored, which might also be an explanation for the lower elasticity found in all subsequent cycles compared to the first cycle.

The results of various finite element studies indicate that the temporomandibular joint disc plays an important role in the mechanics of the joint (Chen and Xu, 1994; DeVocht et al., 1996; Chen et al., 1998; Nagahara et al., 1999; chapters 4 and 5). The studies mentioned were of (quasi-) statical nature. However, in dynamical analysis the specific characteristics found experimentally (chapter 6) need to be implemented in the model in order to obtain valuable results. The results of the present study indicate that the poroelastic material model can be applied for this purpose.