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

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Chapter 5

THREE-DIMENSIONAL FINITE ELEMENT ANALYSIS
OF THE CARTILAGINOUS STRUCTURES
IN THE HUMAN TEMPOROMANDIBULAR JOINT

Abstract- The three-dimensional distribution of loads and deformations in the cartilaginous structures of the human temporomandibular joint during statical joint loading tasks was examined by means of finite element simulations. A finite element model of the joint including a deformable cartilaginous disc and deformable cartilage layers on the articular bones was developed according to the geometry of the right joint of an embalmed male cadaver. Loading of the joint was simulated by prescribing a displacement to the mandibular condyle and was performed in four different habitual condylar positions. Furthermore, the influence of loading direction and material properties of the soft structures was examined. The results indicate that the disc is loaded in its entire intermediate zone when the condyle is located in the fossa of the temporal bone and that the loads are concentrated in the lateral part of this intermediate zone when the condyle is positioned on the eminence of the temporal bone. By adapting its shape to the geometry of the articular surfaces, the disc maximized the contact areas between these surfaces. This way, the presence of the disc prevented local peak loading of the cartilage layers. The load distribution capability of the disc appeared to be proportional to its elasticity. The soft fibrocartilage layers on the osseous structures in the joint strengthened the load distribution function of the disc by contributing to an increased congruency.
Introduction

The stresses and strains in the cartilaginous structures of the temporomandibular joint during loading are very difficult to obtain experimentally. They can be estimated by mechanical analysis, for which mathematical modeling has demonstrated to be a powerful tool. The governing mathematical equations describing the mechanics of these tissues are generally too complex to be solved analytically. Fortunately, the finite element method has proven to be a suitable numerical tool for obtaining adequate estimations for relevant loads and deformations (e.g., Huiskes and Chao, 1983; Mow et al., 1993; Koriath and Versluis, 1997; Van Eijden, 2000). This technique has already been applied to the temporomandibular joint (Chen and Xu, 1994; DeVocht et al., 1996; Chen et al., 1998). However, these models were two-dimensional, supposing that the strains in the direction perpendicular to the sagittal plane can be neglected and that the strains in the selected section are representative for the strains beyond this section. Recently, a three-dimensional finite element model of the temporomandibular joint disc has been developed (Beek et al., 2000a; see chapter 4). Results obtained with this model, indicate that the deformations of the disc are three-dimensionally distributed and that the assumptions applied for two-dimensional models did not hold.

In the model described in chapter 4 the articular surfaces of the mandibular condyle and the temporal bone were assumed to be undeformable. However, it has been suggested that deformable cartilage layers may play an important role in the load transmission in diarthrodial joints (Schreppers et al., 1990). Furthermore, the joint loads were analyzed only in a closed jaw configuration. During jaw opening, the mandibular condyle performs large translational and rotational movements with respect to the temporal bone. This means that the shapes of the contacting portions of the articular surfaces vary considerably. Presumably, this will influence the stress and strain distributions in the cartilaginous structures of the joint. This was confirmed by DeVocht et al. (1996) in a sagittal plane-strain analysis. Guided by our previous findings (Beek et al., 2000a), however, it remains uncertain to what extent their findings are applicable to the three-dimensional nature of the joint.
In the present study, a three-dimensional finite element model of the joint including a deformable cartilaginous disc and deformable cartilage layers on the articular bones was applied to elucidate the particular influence of each of the cartilaginous structures on the stress and strain distributions in the joint. Furthermore, the consequences of jaw displacement for three-dimensional joint load distribution were investigated.

Materials and methods

Model

A three-dimensional finite element model was created according to the right temporomandibular joint of an embalmed male cadaver (age: 77 years). This particular joint did not show any abnormalities. The shape of the various relevant surfaces was reconstructed by fitting polynomial surfaces through geometric measurements collected with a magnetic tracking device. For a detailed description of this procedure, see chapter 2 and 3 of this thesis. The reconstructed shapes of the disc and the cartilage layers were filled with tetrahedral elements using an automatic mesher (Menta 3.2, MARC Analysis Research Corporation, Palo Alto, USA). To be able to apply this mesher, the polynomial surfaces were overlayed with triangular patches. For a detailed description of the creation of the finite element mesh, see chapter 4. The thickness of the disc ranged from 1.04 mm at the intermediate zone to 3.29 mm at the posterior band region. The thickness of the cartilage layers could not be measured and was considered to have a constant value of 1.0 mm. Although this is somewhat thicker than the average found by Hansson et al. (1977), we applied this value for model stability reasons. The different articular structures were meshed separately. In order to merge them to a single joint model, remaining local incongruencies were reduced according to Beek et al. (2000a). The total mesh consisted of 35,717 tetrahedral elements (disc: 13,382, condylar cartilage: 10,368, temporal cartilage: 11,967) and had over 28,000 degrees of freedom. Figure 1 displays the finite element model in its initial state, when the jaw is closed.
Three-dimensional finite element model of a right human temporomandibular joint in the jaw-closed configuration. Orange: cartilage layer on the temporal bone (fossa plus eminence). Purple: temporomandibular joint disc. Red: cartilage layer on the mandibular condyle. On the right, the nomenclature used in the text.

The present study was focussed on statical behavior. The mechanical properties of the disc as well as the cartilage layers were considered to be homogeneous and isotropic. For the Young's modulus $E_{\text{disc}}$ of the disc, a value of 6.80 MPa was applied which was intermediate between the values used by others (Korioth et al., 1992; Tanaka et al., 1994; Chen and Xu, 1994; Chin et al., 1996; DeVocht et al., 1996; Lai et al., 1998; Nagahara et al., 1999). For the cartilage layers, a value of 0.79 MPa was applied for the Young's modulus $E_{\text{cart}}$ (Tanaka et al., 1994). The Poisson's ratio applied was considered to be 0.4 for both tissues (Tanaka et al., 1994). The mandibular condylar bone and the temporal bone were assumed to be rigid. The temporal bone was fixed during the simulations. Friction inside the joint was assumed to be negligible (Nickel and McLachlan, 1994; Chen and Xu, 1994). Loading of the model was simulated by prescribing a displacement in a particular direction to the condyle. Because the stress and strain distributions inside the cartilaginous structures of the joint are unknown, the strains and stresses were
neutralized before each simulation. This means that all results presented should be interpreted relatively to the initial state of each simulation.

**Simulations**

The simulations were performed using the commercially available finite element software MARC K7.3 (MARC Analysis Research Corporation, Palo Alto, USA). This software was installed on the Cray C916 National Supercomputer at SARA in Amsterdam. In anticipation of large deformations, the stress and strain fields were calculated using the finite deformation theory. Basically, this theory employs the Cauchy stresses and true strains. Furthermore, an updated Lagrangian procedure was used which takes not the first but the current increment as a reference. The contact phenomenon was solved using the Direct Contact Method (MARC, 1996). All simulations were terminated before deformations became so large that the prerequisites for proper stress-strain analysis were exceeded. To allow comparison with our previous study, a reference simulation was performed according to Beek et al. (2000a). In a configuration concomitant with the jaw in a closed position, the condyle was displaced in a direction according to the estimated joint reaction force, perpendicular to the articular surfaces at the site of contact (Koolstra and Van Eijden, 1992). Stresses and strains were represented by the Von Mises stress and the equivalent elastic strain ($\varepsilon_{el} = \sqrt{2/3} \varepsilon_{ijkl} \varepsilon_{ijkl}$), respectively.

The possible influence of the presence of the disc inside the joint on the distribution of the deformations on the cartilage layers was elucidated by performing a similar simulation after the disc was removed. The sensitivity to loading direction was studied by performing simulations, in which the loading direction was varied in anterior, medial, lateral and posterior direction by 30° relative to the reference simulation. The sensitivity of the model for variations in the Young's modulus of the disc was investigated by performing simulations in which this parameter was given values of 0.068 MPa, 0.68 MPa, and 68.0 MPa. The value of 0.068 MPa was used for $E_{\text{disc}}$, because then the disc is softer than the cartilage layers, the value of 0.68 MPa was according to Chin et al. (1996), and the value of 68.0 MPa was according to Tanne et al. (1991). The influence of the position of the condyle relative to the temporal bone was investigated by performing simulations in which the condyle was positioned in four different positions. From its reference position in the fossa (A), the
condyle was rotated forwards over 20° (B), moved about 5 mm forwards and downwards along the eminence without rotation (C) or moved to this location combined with a 20° rotation (D), roughly according to a closed position, a (retrusive) terminal hinge position, a protrusive position and a habitual (wide) open position, respectively (Posselt, 1962). For each situation, the size of the contact areas between the cartilaginous structures was assessed roughly by adding the areas of the surface faces of individual elements which had a stress of over 0.005 MPa. The same procedure was repeated for a stress of 0.01 MPa to obtain a measure for the concentration of the contact stress. Furthermore, for each position the maximum values for the equivalent elastic strain and the Von Mises stress at the same compressive force of 1 N were determined.

Results

Reference simulation

Figure 2.
The deformations (equivalent elastic strain $\varepsilon_{el}$) in the cartilaginous structures of the temporomandibular joint during loading in the direction perpendicular to the articular surfaces in the jaw-closed configuration in a frontal view. Color bar: amount of $\varepsilon_{el}$.

In order to displace the condyle over 0.19 mm from the unloaded initial state in the direction perpendicular to the joint surfaces, a force of 8.84 N was needed in case of the jaw-closed configuration. This force was calculated by summation of the nodal forces that were generated by the contact algorithm (MARC, 1996). In Fig. 2 the predicted equivalent elastic strain distributions in the condylar cartilage, disc and temporal cartilage are displayed in a frontal view. This figure shows that all three
structures were deformed to absorb the contact forces, generated by the condylar
displacement and that these deformations were largest in the cartilage layer on the
condyle, reaching values up to approximately 15% for the equivalent elastic strain $\varepsilon_{el}$.
The stresses in the various structures were distributed similarly to the strains. In the
disc larger stresses were generated than in the cartilage layers. The maximum
values for the Von Mises stress were 0.13 MPa, 0.30 MPa, and 0.11 MPa in the
condylar cartilage, the disc, and the temporal cartilage, respectively.

After removal of the disc, loading of the joint in the same direction as in the
reference simulation resulted in extremely concentrated strain fields in both cartilage
layers (Fig. 3). Removal of the disc resulted in a decrease of the contact areas on
both cartilage layers to about 50% (Table 1). The table also shows that without disc
the strains in the cartilage layers are much larger at the same load compared to the
case with disc.

![Figure 3.](image)

The deformations in the cartilage layers of the temporomandibular joint without the presence
of a disc during loading in the jaw-closed configuration in a frontal view. Color bar: amount of
$\varepsilon_{el}$.

**Influence of material properties**

The influence of the magnitude of the Young's modulus of the disc on the strain
distribution in the cartilaginous structures during joint loading in the jaw-closed
configuration is shown in Fig. 4. A lower value for the Young's modulus of the disc
resulted in more deformation but lower stress values in the disc. Increasing the
Young's modulus of the disc relative to the one of the cartilage layers, resulted in a
gradual shift of the deformations from the disc to the cartilage layers. The force
needed to displace the condyle over a similar distance (0.13 mm) was smallest when
the Young's modulus of the disc had a value of 0.68 MPa.
Chapter 5

Influence of loading direction

Variation of the loading direction had little influence on the strain distribution in the disc (data not shown). On the other hand, the strain distributions in both cartilage layers were slightly influenced by the loading direction. In both layers, the distributions showed a small shift in the direction concomitant with the loading direction.

Mandibular position

A rotation of the condyle by 20° to obtain a retruded jaw-open configuration led to higher strain values (Fig. 5B). Due to its rotated position relative to the disc, the strains in the condylar cartilage were located more posteriorly. The locations of the strains in the disc and in the temporal cartilage remained unaffected.

During the translation of the condyle from a closed-jaw position to a protrusive position, the disc moved along with the condyle in anterior direction. The region of the disc in contact shifted from the entire intermediate zone to the lateral part of the intermediate zone (Fig. 5C, D). The protrusion also resulted in a small displacement of the disc in medial direction. Furthermore, the shape of the disc became flatter during this translation. Due to changed location of the contact area, the deformations

Table 1.

Contact areas and the maximum values for the equivalent elastic strain (εeq) and the Von Mises stress (σVM) estimated at a condylar load of 1 N. The contact areas are given for two different stress values.

<table>
<thead>
<tr>
<th>Contact area (mm²) with σVE &gt; 0.005 MPa</th>
<th>Condyle-Temporal</th>
<th>Condyle-Disc</th>
<th>Temporal-Disc</th>
<th>Condyle-Disc</th>
<th>Temporal-Disc</th>
</tr>
</thead>
<tbody>
<tr>
<td>Contact area (mm²) with σVE &gt; 0.010 MPa</td>
<td>Condyle-Temporal</td>
<td>Condyle-Disc</td>
<td>Temporal-Disc</td>
<td>Condyle-Disc</td>
<td>Temporal-Disc</td>
</tr>
<tr>
<td>Maximum εeq (%)</td>
<td>Condyle</td>
<td>Disc</td>
<td>Temporal</td>
<td>Condyle</td>
<td>Disc</td>
</tr>
<tr>
<td>Maximum σVM (MPa)</td>
<td>Disc</td>
<td>0.18</td>
<td>0.06</td>
<td>0.09</td>
<td>0.10</td>
</tr>
<tr>
<td></td>
<td>Temporal</td>
<td>0.12</td>
<td>0.03</td>
<td>0.04</td>
<td>0.07</td>
</tr>
</tbody>
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<tr>
<th>Contact area (mm²) with σVE &gt; 0.005 MPa</th>
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</tbody>
</table>
**Figure 4.**
Influence of the Young’s modulus of the temporomandibular joint disc on cartilaginous deformation in the jaw-closed position in a frontal view. Left panels: cartilage layer on the temporal bone. Middle panels: joint disc. Right panels: cartilage layer on the mandibular condyle. Color bar: amount of $\varepsilon_{el}$.

in all cartilaginous structures were biased to the lateral side when the condyle was located on the eminence.

Due to the changed location of the condyle, the loading direction in the protrusive position was oriented more vertically than in the jaw-closed and retruded jaw-open configuration. Compared to the jaw-closed configuration, the distribution of the
Figure 5.
Cartilaginous deformation in various joint configurations in a frontal view. A: jaw-closed configuration. B: retruded jaw-open configuration. C: protractive configuration. D: jaw-open configuration. 1st column: different configurations of the joint model in a medial view. 2nd column: cartilage layer on the temporal bone. 3rd column: joint disc. 4th column: cartilage layer on the mandibular condyle. Color bars: amount of $\varepsilon_{el}$. For clarity reasons, the colored strain range for the disc differs from that of the cartilage layers.

strains in the cartilage layer on the temporal bone shifted from the posterior slope of the eminence to its top (Fig. 5C).

An open-jaw configuration was obtained by a rotation of 20° of the condyle in the protractive position. The joint was loaded in the same direction as in the protractive configuration. The strain distributions in the temporal cartilage and the disc were unchanged compared to the protractive configuration, and the strains in the condylar cartilage were located more posteriorly (Fig. 5D).

The size of the contact area at the condyle-disc interface was larger or equal to the area at the temporal bone-disc interface (Table 1). Especially in the closed position there was a remarkable difference. When the condyle-disc complex was
located on the eminence the part of the contact area that underwent larger stresses was larger than when the condyle was located in the fossa. This is reflected by a smaller area with higher Von Mises stress and by smaller strains in both cartilage layers when the condyle-disc complex was located in the fossa.

Discussion

Model assumptions

In the temporomandibular joint the bones are covered with fibrocartilage (Moffett et al., 1964). This is contrary to many other diarthrodial joints and is reflected in the applied value for the elasticity of the cartilage layers, which is smaller than applied in other models (Schreppers et al., 1990). Previous finite element studies which included the temporomandibular joint and assumed the mechanical behavior of the disc to be linear elastic, mostly used mutually different values for its Young's modulus (Korioth et al., 1992; Tanaka et al., 1994; Chen and Xu, 1994; Chin et al., 1996; DeVocht et al., 1996; Lai et al., 1998; Nagahara et al., 1999). Recently, a large spread has been demonstrated in the mechanical characteristics of the disc of different individuals with sd's up to 90% (Beek et al., 2001a; chapter 6). While it is not known whether the elasticity of the articular cartilage varies proportional with that of the disc, only the latter was varied in a sensitivity analysis. The present results suggest that a value similar to the one of the cartilage layers is needed for the most evenly distribution of deformations, and that a higher value reduced the deformation of cartilaginous structures in the joint during loading.

There exists a large amount of biological variations concerning the geometry of the various structures in the temporomandibular joint and there is no agreement regarding the mechanical properties of its tissues. Probably, these parameters are subject to biological variation too. Therefore, the quantitative data from any mathematical model of this joint (and others) have to be considered with care. The present results, therefore, have to be interpreted expressly as qualitative.

The simulations predicted relatively large deformations at small loads. This was confirmed experimentally for bovine cartilage (Soltz and Ateshian, 1998). These
authors also found equilibrium stresses similar to the ones predicted in the present study. The applied loads, however, are far below the ones estimated during habitual function (Koolstra and Van Eijden, 1992). Large loads will cause deformations far beyond the possibilities of the applicable finite element algorithms. However, we assume that the results of the present study are nonetheless representative, because the amount of load is supposed to have little influence on the distribution of deformations in the structures due to the application of a linear material model. In our statical analyses we applied material parameters obtained at equilibrium. However, equilibrium in cartilaginous structures is normally not reached within 2.5 hours (Soltz and Ateshian, 1998) and, therefore, is hardly ever reached under physiological conditions. The fluid content in cartilaginous structures is about 70% by weight, which supports 90% of the total stress for as long as 400 s (Soltz and Ateshian, 1998). This could be incorporated by adaptation of the disc's Young's modulus for simulating short-time, statical loading tasks or applying material models, that contain fluid support (e.g., biphasic or poroelastic models).

In the model the cartilaginous articular layers had a uniform thickness of 1 mm. This is somewhat thicker than averagely observed. As a consequence the influence of the articular cartilage has been slightly overestimated. Presumably, where this layer is relatively thin load bearing capacity will have to be transferred to the disc. For the upper articular surface the cartilage was thinnest in the roof of the fossa and for the condyle in the posterior part (Hansson et al., 1977). It was remarkable that at these sites hardly any loads were predicted by our model. Consequently, the overestimation of the influence of the articular cartilage can be considered negligible.

**Mechanical analysis**

From the results obtained from the simulation without disc, it can be concluded that the presence of a disc inside the temporomandibular joint prevents extreme and concentrated deformations of the cartilage layers. By decreasing the incongruency of the articular surfaces, the disc enabled an eight times larger loading of the joint. This suggests that the disc has a load distributing function in the joint. This suggestion is also supported by the fact that the disc became flatter when the condyle moved anteriorly in order to adapt to the changing geometry of the contacting surface of the temporal bone. Variation of the Young's modulus of the disc revealed that the load
bearing capability of the joint is proportional to this parameter. Although the cartilage layers themselves have limited potential of load bearing, they strengthen the load distribution function of the disc by contributing to an increased congruency. Apart from the functions elucidated by the results of the present study, the disc may play additional roles in the mechanics of the joint including for example the dissipation of energy during impact loading (Beek et al., 2001a; chapter 6).

Previously published finite element studies involving a freely movable temporomandibular joint disc (Chen and Xu, 1994; DeVocht et al., 1996; Chen et al., 1998) were two-dimensional and thus were unable to elucidate the lateral shift of the deformations occurring during anterior movement of the condyle found presently. This shift may be caused by the fact that the eminence is lower at the lateral side compared to the medial side. DeVocht et al. (1996) prescribed condylar displacements along the articular surfaces instead of perpendicular to these surfaces and, therefore, found relatively small strains. Chen and Xu (1994) and Chen et al. (1998) were able to simulate large stresses and large condylar forces but applied a much stiffer disc. The Von Mises stresses found in the present study were comparable with the ones found by DeVocht et al. (1996). Furthermore, these authors also predicted that the disc would move together with the condyle along the articular eminence of the temporal bone during jaw opening. Similar to our simulations, they found that ligaments and the lateral pterygoid muscle were not necessary to guide this movement and that the disc was mainly deformed in its intermediate zone. Chen et al. (1998) predicted stress values in the disc that were twice as large compared to the values in the cartilage layers, supporting our findings summarized in Table 1.

During loading in the jaw-closed configuration, the deformations in the disc were spread in its whole intermediate zone. This would suggest that clenching might lead to damage (perforations) in the whole intermediate zone. Translation of the condyle in forward direction to obtain a protrusive or open-jaw configuration, led to a concentration of the loading in the lateral part of the disc. This would suggest that during open-close movements the lateral part of the intermediate zone is primarily subjected to wear and friction. This is supported by Werner et al. (1991), who reported that wear leading to perforations of the temporomandibular joint disc was mainly located in the lateral part of its intermediate zone.
When the condyle was located at the articular eminence the contact areas were smaller than when the condyle was located in the fossa. In contrast, the relative area with $\sigma_{VM}>0.01$ was larger. This in agreement with the larger incongruency of the joint during protrusion or jaw opening. In the closed position is the relative contact area with $\sigma_{VM}>0.01$ on the lower side of the disc larger than on the upper side. This is in agreement with the difference in curvature between these surfaces.